Strategies Utilized while Minimizing Ankle Motion Bilaterally and Unilaterally during Level Ground Walking and Obstacle Clearance Tasks

by

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A thesis

presented to the University of Waterloo

in fulfillment of the

thesis requirement for the degree of

Master of Science

in

Kinesiology

Waterloo, Ontario, Canada, 2010

Author's Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

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Abstract

A great deal of research has been done on the adaptive strategies of individuals who have been affected by a gait altering ailment, but there is little research on the adaptive strategies to imposed restrictions in the healthy population. The role of the ankle in healthy gait is to generate a "push-off" force to create forward propulsion of the body (Winter, 2004). The purpose of this thesis was to identify adaptation patterns and compensation strategies in individuals while wearing and not wearing a device to reduce ankle motion(Ankle Motion Minimizer – AMM). Motion capture and force plate data were collected to determine the lower body kinematics and joint powers during both level ground walking and obstacle avoidance tasks. Repeated Measure ANOVAs with an alpha level of 0.05 determined that differences in the ankle angles and the ankle, knee, and hip powers existed between the various conditions. Results showed that participants had a decreased range of motion and power production at the ankle joint while wearing the AMM. Meanwhile, an increase in the power bursts from the ipsilateral knee were observed during the AMM conditions as well as small increases at the contralateral ankle and ipsilateral hip during the unilateral AMM condition. EMG analysis showed a distinct muscle activation pattern for each individual muscle during the different conditions. From this investigation, individuals who are unable to produce power through the ankle joint, were able to increase power propulsion predominately at the knee to compensate for the lack of propulsion provided by the ankle, therefore allowing ambulation to continue.

Acknowledgements:

I would like to thank my wife, Sarah, my parents, Patrick and Ursula, and my parents-in-law, Glenn and Barbara for their unwavering support. Dr. Stephen D. Prentice for his guidance as well as Jonathon Singer for all of his assistance throughout this project. Also for financial support from NSERC, Canada.

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1.0 Introduction:

Gait is learned at a very young age and becomes an ingrained muscle pattern early in life. The use of ambulation is paramount to a person's well-being and ability to complete daily tasks. A great deal of research has been done on the adaptive strategies of individuals who have been affected by a gait altering ailment, but there is little research on the adaptive strategies to imposed restrictions in the healthy population. The role of the ankle in healthy gait is to generate a "push-off" force to create forward propulsion of the body (Winter, 2004). This is done through causing angular momentum about the hip and then translating this momentum to the trunk in order to move forward.

The use of Ankle-foot orthotics (AFO) has been utilized in the aid of numerous ailments to prevent a low toe position during swing through either a lack of muscular activation of the tibialis anterior or a muscular contracture of the gastrocnemius and soleus muscles. The AFO is able to fix the toe position to achieve some level of toe clearance during normal walking and obstacle avoidance. An AFO consists of a molded plastic, which starts at the toes, runs under the length of the foot and up the back of the shank. It ends approximately half way up the calf by which it is attached using a Velcro strap. Once the orthotic is on and the foot/orthotic is in the shoe, the shoe will keep the orthotic in place during gait (Figure #1). To increase the restriction of plantarflexion, canvas straps were attached to the top of the AFO and connected to a velcro strap around the toes. The use of the AFO in this investigation is a means to restricting the ankle joint and not for the explicit use in which the AFO was generally meant for. The use of AFOs have not been used on a healthy population during gait and only the use of ankle bracing has been examined during quiet one-legged and two-legged standing on healthy individuals (Bennell & Goldie, 1994). Ankle braces typically restrict most medial and lateral movement of the joint to

prevent either an inversion or eversion strain of the ligaments. The restriction of movement in the anterior and posterior directions tends to be minimal, as this movement is needed to complete the tasks of walking, running and jumping, which occur during most sports in which the braces are utilized. The insight gained from the bracing of healthy individuals showed a decrease in the postural stability, where participants had larger oscillations of their centre of pressure tracings, while in individuals with ankle laxity there was an increase in postural stability (Bennell & Goldie, 1994).The use of a modified AFO on the healthy population will prevent a significant amount of the plantarflexion and some dorsiflexion that is occurring at the ankle, while the musculature is still intact to generate a moment. This moment will have a diminished effect as it will need to work in a more isometric fashion. The modified AFO or Ankle Motion Minimizer (AMM) by preventing movement will also inhibit some of the generation of power that could contribute to the gait cycle. However, the use of the ankle to create stiffness and allow muscle energy to be transferred will still be possible. This is done from a phenomalogical prespective, to determine what changes will occur by reducing the contributions of the ankle joint. Since the ankle joint powers will be diminished, the plasticity of the powers in the lower leg should allow for a redistribution of power to continue in order generate forward propulsion. By minimizing these contributions from the ankle joint, the resulting hip and knee joint moments and powers were able to be examined to determine the compensation mechanisms that may occur. The energetic cost of walking with an immobilized ankle has previously been reported to be no greater than that of a weight matched limb that is free to plantarflex as normal (Vanderpool, 2008). While the reduction in plantarflexion during push-off was obtained, a curved rocker bottom on the brace utlized was able to reduce the negative work at the ankle, therefore reducing the positive work needing to be performed to progress the body forward, also they were unable

to identify the exact source(s) of ability to achieve similar velocity of walking despite the loss of ankle power (Vanderpool, 2008).

The comparison of the hip and knee joints kinetic and kinematics can be analyzed and their values compared from the walking with the AMM against walking without the AMM condition. Any differences between the two conditions can then be attributed to the restriction of the ankle joint and its contributions, or lack thereof, to ambulation. According to Winter (1991), the three moments of the lower limb add up to the support moment to indicate how the limb as a whole will produce support for the trunk to allow ambulation; however, their relative contributions are not fixed. This plasticity in the joint moments can allow for one joint to compensate for another in their moment production (Winter, 1991). These insights can be used to theorize and create rehabilitation strategies which better account for the deficiencies of the individual (Abel, Juhl, Vaughan, & Damiano, 1998; Chen, Yeung, Wang, Chu, & Yeh, 1999; Lehmann, 1993; Wang, Lin, Lee, & Yang, 2007; Wang et al., 2005; Weiss, Brostrom, Stark, Wick, & Wretenberg, 2007).

Figure 1: Ankle-fixation orthotic

(Photo taken from http://www.neuromuscular-orthotics.com.au/images/PLS%20AFO.jpg)

2.0 Purposes:

The purposes of this research are to:

- Determine what alternative/ compensation strategies are used when ankle dorsiflexion/ plantar flexion is limited during the normal gait cycle both in level walking and during avoidance of a 30 cm high obstacle
- Determine the loss of contribution of the ankle joint to overall body propulsion during the gait cycle in level walking and while avoiding a 30 cm obstacle.
- Determine the loss of contribution of the ankle to overall body propulsion in level and obstacle walking while it is functioning in a limited capacity due to the wearing of an AMM.

The knowledge of how the body adapts to different situations, through both a muscular and kinematic/kinetic view is critical in learning how the body, be it neural, muscular, or both, is able to compensate for limitations that are being placed on it. The scope of this research will aid in defining the contributions of the joint powers through placing limitations on the ankle joint. This knowledge will be applicable to the plasticity in joint contributions to overall forward propulsion. These data will allow the contributions of the ankle joint powers and subsequently the knee and hip joints to be adequately accounted for in the understanding of human gait.

3.0 Hypotheses

The ankle joint in a fully functional individual has been previously shown to contribute approximately 80% of the forward propulsion to walking (Winter, 2004). This is stated for a fully functional ankle joint that is allowed to move freely through its full range of motion without any restrictions or dysfunctions being present. For patients that have some limitation, which does not allow the ankle to move through its full range of motion, they will by definition have a decrease in power production about this joint (Winter, 1991). This decrease in ankle power has to be compensated by either the knee or the hip or both, to allow the person to ambulate at the same rate. This alteration in joint moments as well as joint powers can be seen is certain populations. From the removal of the ankle joint and its contributions to the overall forward propulsion through the gait cycle we can examine the remaining joints and theorize the actual involvement in the progression of gait. It is therefore hypothesized **that the ankle joint will contribute less to the overall body propulsion when the ankle is functioning in a limited capacity then during normal walking.**

While the ankle is functioning at a limited capacity and in a fixed position, the muscle contraction and the moments of an able-bodied participant will have very limited mobility about the ankle joint and result in a minimal angular velocity. Since the power of a joint is calculated from its moment multiplied by the angular velocity, this would result in minimal joint power. To ambulate however, a particular support moment is required which is made up of the moment of the ankle, knee and, hip (Winter, 1991). Therefore it is hypothesized that **the hip and knee powers, specifically H3 in both regular and obstructed gait, and also K5 during obstacle avoidance, will increase to compensate for a loss of ankle power.** Also it is hypothesized that

the hip power will make up the majority of the lost power from the ankle through its contribution of the third power burst or pull off power.

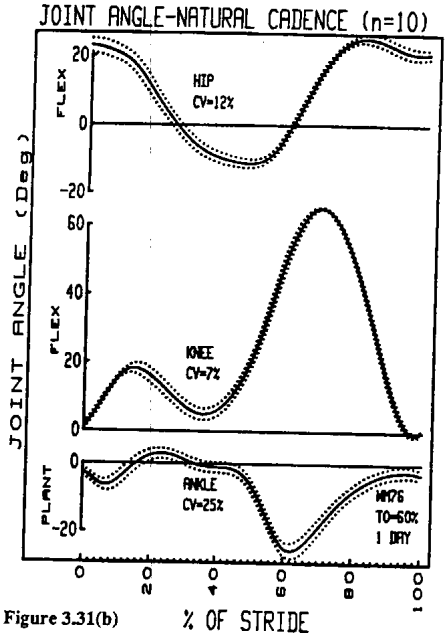
From a previous study by Landy, Singer, and Prentice (unpublished) based on stilt data, there was an increase in co-contraction of both the tibialis anterior muscle and the gastrocnemius muscle throughout the gait cycle. Since the human ankle joint was restrained in the stilts, this increase in muscle activity was unexpected. However, many studies (Anderson & Pandy, 2003; Crabtree & Higginson, 2009; Lehmann, 1993; Neptune, Kautz, & Zajac, 2001; Niang & McFadyen, 2004; Perry, 1992; Winter, 1983; Winter & Sienko, 1988; Zmitrewicz, Neptune, & Sasaki, 2007) showed that plantarflexors eccentrically contract to control and maintain a balance of the total body centre of mass. Also the plantarflexors and tibia restrains the forward progression over the foot. (Sadeghi et al., 2001) As the tibia is rotating over the ankle joint the knee flexes at half of this rate (Perry, 1992). This is due to the quadriceps counteracting this forward rotation to aid in the forward "free fall" of the body over the ankle. This leads to the hypothesis that **restraining the ankle joint during walking will lead to an increase in muscle activation, of the plantarflexors and the knee extensors**.

4.0 Literature Review

This section reviews the current literature including the existing theories of the lower leg musculature and its contributions towards propulsion in normal walking and special situations, the conditions that would compromise the ankle joint, and the alternatives to dorsiflexion and plantar flexion.

4.1 Normal walking/patterns – how the ankle is involved

The ankle joint has been strongly correlated with the forward propulsive forces or "push-off" power (the A2 power burst; see figure 2) required for gait. Three main theories identified with the ankle plantar flexor group consist of: 1) that the plantar flexors provide a deceleration of the tibia over the ankle and alter the function at the knee during the stance phase (A1 Power burst), 2) the plantar flexors provide forward progression just prior to toe-off in the gait cycle (Winter, 1983) and, 3) the plantar flexors accelerate the leg into the swing phase and this inertia energy is transferred to the trunk to provide forward progression through the H3 Power burst (Meinders, Gitter, & Czerniecki, 1998; Neptune et al., 2001). While these three theories are not exclusive to one another, the contribution of the ankle joint powers to a propulsive force has been controversial.



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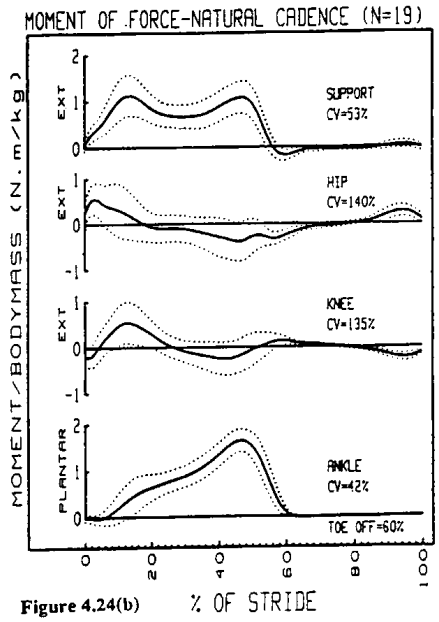

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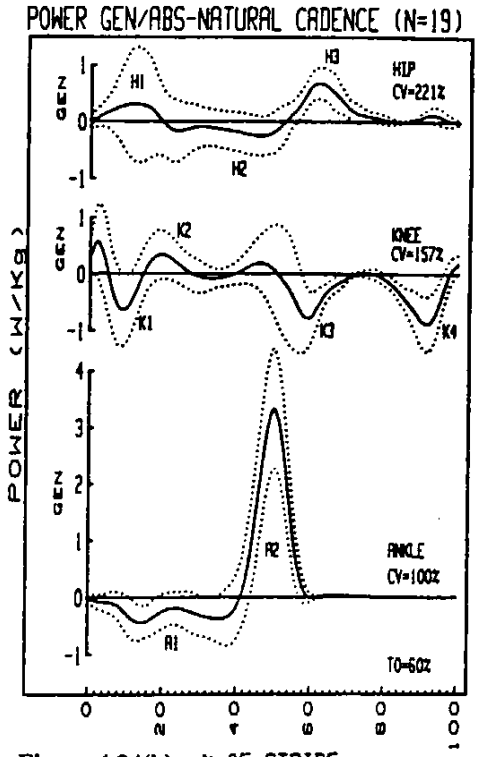

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Figure 2: Normal joint angle, moment of force, and generation of power for each of the ankle, knee and Hip joints (Winter, 1991).

The knee joint is proposed to have a stabilizing effect on the ankle and allow for energy transfer to occur between the ankle and the trunk to permit forward propulsion via the conservation of energy from the A2 power burst and the H3 power burst to initiate swing. The K2 power burst (see figure 2) is the only positive power burst from the knee and represents only approximately 15% of the propulsive forces in level gait. The actual power to swing the leg forward comes from a pendulum action and the hip flexor moments (Winter, 1991).

The hip power bursts of H1 and H2 are rather low and are mainly to maintain trunk balance during slow and normal cadences. During fast cadence the hip powers become more defined. H1 power burst is defined as the concentric action of hip extension, to pull the trunk forward (Winter, 1991). The H2 power burst is the absorption of energy by the hip flexors as the thigh is decelerated during its backwards rotation, usually stabilizing the trunk. The H3 (pull-off) burst is part of the pendulum effect to rotate the leg forward and prepare for the next heel contact (Winter, 1991) through concentric action of the hip flexors.

The power bursts have been well defined in the literature for particular muscle groups but not in relation for each particular muscle. As numerous muscles cross more than one joint, this allows for several actions of a particular muscle throughout the duration of the gait cycle. Of particular interest is the knee- hip synergy during the gait cycle, which allows a trade-off from the knee moment to the hip moment and vice versa as long as the support moment maintains a similar profile. The support moment is defined as the sum of the moments from the ankle, knee and hip normalized to one gait cycle. This synergy has been previously correlated and the mean hip moment versus the mean knee moment gives us a good negative correlation (-0.99) by which when the knee is extending the hip will be flexing and vice versa (Winter, 1991).

4.2 Contribution of the ankle during different situations

4.2.1 Changes in gait speed

The proportion of contribution from the ankle joint powers can vary during different tasks and walking patterns. Winter (1991) described the joint power contributions during varying speeds of gait. It was determined that at an increase in speed, the contribution of each of the joint powers changed to allow for a more dynamic flow of energy from one segment to another. In particular, the A1 power burst will decrease as speed increases to cause less braking in the initial part of the gait cycle. Due to this decrease of the plantar flexors, there is a decrease in undesirable foot slap. Therefore the level of contraction to prevent foot slap from occurring by the tibialis anterior muscle has decreased, which aids in keeping speed up in early stance. Also an increase in the A2 power burst will be applied as speed increases as this is where 80% of the propulsive forces originate from. The H3 power burst will also increase to counteract the backwards rotation of the leg and to initiate swing phase, from which most of the energy transfer occurs from the leg into the trunk and allow for a forward progression of the trunk.

4.2.2 Changes in gait during aging

The changes in gait during aging have been given considerable attention (Riley, DellaCroce, & Kerrigan, 2001; Schmitz, Silder, Heiderscheit, Mahoney, & Thelen, 2008; Winter, 1991) as there are many adaptations occurring during the aging process. Neurological, musculoskeletal and physiological changes have all been stated to contribute to the alteration in the characteristics of the gait cycle. As the body ages the neurological system exhibits a decrease in the transmission speeds along the neural pathways, there is an earlier onset of disease and, also a decrease in the total number of cells. The musculoskeletal system becomes altered as a decrease in strength is

observed. Stiffer joints become more apparent due to either an increase in co contraction of the musculature across the joint itself, a decrease in the laxity of the ligaments and/or, a degenerative disorder such as arthritis (Prince, 1997). Some of these disorders often lead to pain and discomfort during ambulation and movement in general, which leads to a more sedentary lifestyle and a decrease in the range of motion of the joints which are not being used (Prince, 1997). From a physiological point of view the contraction levels that can be obtained are much less effective than those of a younger able-bodied individual. A decrease in the number of motor units and a decrease in the number of muscle units limit the level of contraction that can be achieved (Prince, 1997). Due to the demands of walking up stairs, over obstacles and, navigating smaller spaces, to name a few situations, these tasks tend to become more arduous as the physical demand to complete the task is still the same, however, this will require a higher percentage of an elderly person's maximal strength to achieve.

Some of the general gait characteristics which have been reported to be altered due to aging include: decrease in stride length to increase the amount of double support time, increase in stride frequency, decrease in A2 power burst by approximately 25%, an increase in the H3 power burst (also have an increase in the hip extensors just before toe off), and an increase in the heel contact velocity (DeVita & Hortobagyi, 2000; Winter, 1991, 2004). This shift, from the A2 to the H3 power burst, of distal to proximal power production is noted to exist in active, healthy, elderly adults and also seems to be more pronounced at faster walking speeds (DeVita & Hortobagyi, 2000; Schmitz et al., 2008). This is coupled with the neurological activation pattern which includes a greater level of co contraction about the ankle joint to potentially increase ankle stiffness to accommodate any problems with balance (Schmitz et al., 2008). Through the same investigation, results showed that older adults increased their level of hamstring activity (hip

extensors) during loading and mid-stance at faster walking speeds. This increase in the hip extensors is facilitated by a forward trunk posture. This forward posture is theorized to allow the hip extensors to be further stretched and enable them to produce a larger amount of power when compared to a more erect posture (DeVita & Hortobagyi, 2000). This has been noted to contribute to the increased hip extensor power during early-stance, which is commonly observed in elderly adults as part of this distal to proximal shift. The support moment has been documented for both elderly and young subjects to be identical when normalized to the participant's body weight. The individual contributions of the ankle, knee and, hip however do not remain similar as the elderly show a redistribution of joint powers and torques to deemphasize the ankle plantar flexors and emphasize the hip extensors (DeVita & Hortobagyi, 2000).

4.2.3 Changes in gait with obstacle avoidance

Through everyday life and ambulation the avoidance of various obstacles is necessary to prevent tripping and/or falling. Insufficient limb elevation during the swing phase has been reported as a main contributor to tripping, whereas high foot contact velocity at heel contact with a small foot contact area (heel contact only vs. flat foot contact) with the ground can be the cause of a slip (Patla & Rietdyk, 1993). Approximately 52% of the elderly population experience a fall each year. From these falls 59% are due to trips or slips of some variety. These trips were most prevalent and occurred when the trailing foot strikes an obstacle during the swing phase of the gait cycle (Berg, Alessio, Mills, & Tong, 1997). These falls may result in fractures, complications of these fractures which can include hospital stays, loss of mobility and, death. The younger population is not void of such risks that obstacles employ, however, their complications tend to be of a less severe magnitude. The ability to negotiate an 8 inch (15 cm)

curb is a common obstacle that one faces on a daily occurrence. From the literature, there are two main theories about joint angles and moments when considering obstacle avoidance and clearance, hip hiking strategy and a knee strategy. Hip hiking has been previously described as a vertical translation of the hip to increase the toe clearance over an obstacle (Byrne & Prentice, 2003). This vertical translation of the hip can be achieved by biasing the normal swing phase trajectory to increase the distance between the toe and the obstacle (Patla & Rietdyk, 1993). This vertical translation has also been noted to be achieved by an increase in ankle plantar flexion of the stance leg (Perry, 1992). Therefore less knee flexion would need to be employed to achieve a safe toe clearance (Byrne & Prentice, 2003). In contrast, a knee flexion strategy can be utilized. This strategy is performed with an increase in active knee flexion and passive hip and ankle flexion (Byrne & Prentice, 2003). The knee flexors can also be shown to have an added effect of hip flexion due to the coupling action between the joints of the lower leg. A decrease in the K3 power burst (associated with knee absorption power right after toe off, an eccentric action of the knee extensors) as well as the H3 power burst gives way to an increase in the knee flexors, the new K5 power burst (See figure#3), which is used for increasing the lower limb elevation (McFadyen & Winter, 1991; Niang & McFadyen, 2004). This new K5 burst has been proposed and proven to increase the hip flexion during obstacle avoidance instead of the H3 burst more associated with "pull-off". This knee flexor power generates a rotational energy which is able to induce a movement-dependant flexor moment about the hip joint (McFadyen & Winter, 1991; Niang & McFadyen, 2004). These effects will cause an increase in the hip angle to elevate the distal end of the femur and therefore raise the leg, where as the increase in knee angle will elevate the distal end of the tibia/fibula and therefore increase the distance from the toes to the top of the obstacle (Patla & Rietdyk, 1993). This knee flexion strategy seems to be preferred to

the hip hiking strategy as it allows for greater dynamic stability. While observing toe clearance during normal walking and obstacle avoidance, Patla & Rietdyk, 1993, noted that there was a large increase in margin of safety during obstacle clearance (10 cm) versus normal walking (3.5 cm). This allows for a larger variation of limb control in that each joint has a greater range of error compared to regular walking before the toe is unable to clear the obstacle. Finally in both strategies, there is an increase in dorsiflexion to elevate the toes just a little higher to make sure toe clearance has been achieved.

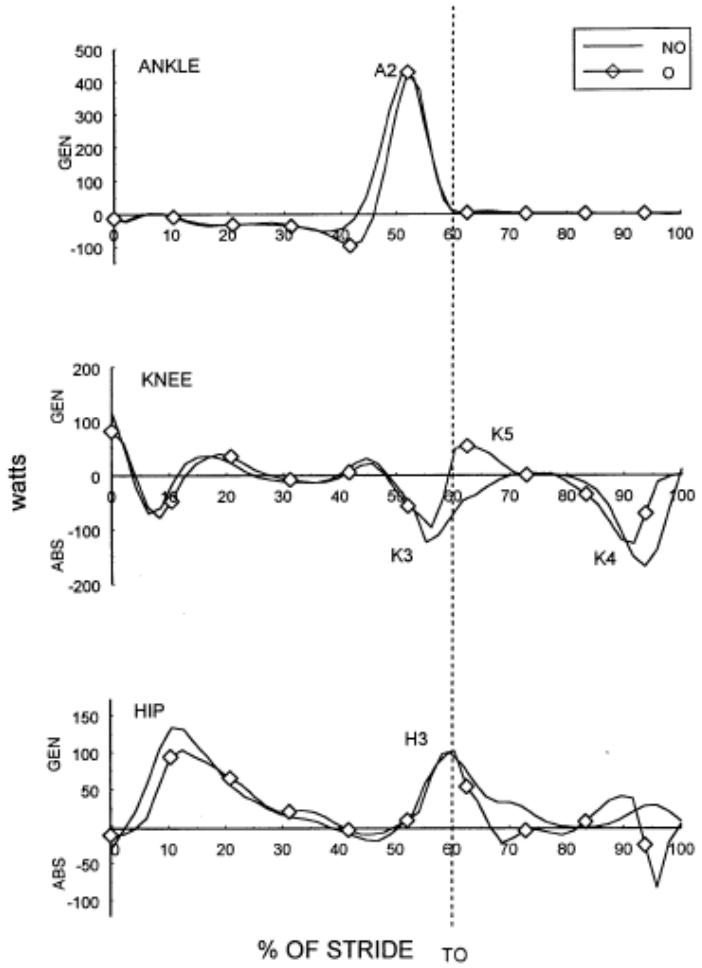

Figure 3: Angle, Knee and Hip joint powers during obstacle avoidance. The solid line represents the joint power during level ground walking while the diamonds represent the joint powers during obstacle walking (McFadyen & Carnahan, 1997)

4.2.4 Changes in gait due to prosthetic limbs

Amputees pose a unique problem when it comes to ambulation due to the lack of internal structure and stability as well as muscular contributions that are normally used in gait. Approximately 1 in 1500 people has an amputation of some degree and this number is expected to rise due to an increase in the mean age of the population (Perry, 1992). These amputations can be due to a variety of reasons including: trauma, infection, tumors, dysvascularity of the limb, and neurological disorders (Perry, 1992). For the scope of this project below knee amputations will be focused on as this has more relevance to the contributions of the ankle joint. Amputees are fitted with prosthetics to provide a stable and consistent environment to walking on due to the lack of the normal human foot, ankle, and part of the lower leg. Due to the loss of the ankle joint and the muscles that cross here, there is no plantar flexion occurring, which can play two major roles in the normal gait cycle. The gait of amputees has therefore been adapted to compensate using the hip and knee (Winter & Sienko, 1988). During the gait cycle, for the side with the prosthetic, the hip angle has been shown to stay close to normal and the knee angle is not as smooth when compared to able-bodied participants (Winter & Sienko, 1988). This leads to an asymmetrical gait pattern in amputees to contradict the symmetrical pattern that is known to occur in the able-bodied. This new asymmetrical gait has been suggested to be a new optimal pattern for an amputee within the confines of the already learned “normal” gait pattern and the new mechanics of this prosthesis (Winter & Sienko, 1988). The Gressinger foot (flex foot) has the ability to store energy in the mechanism of the prosthesis to apply later on during the normal toe-off phase. This has been previously reported that 30% of the energy that is stored in the ankle spring mechanism was returned at push-off (Winter & Sienko, 1988) and through a simulation method it was reported that the energy released by the prosthesis (12.0 J) was less than half of

that normally delivered by the soleus and gastrocnemius (28.3 J) from an able-bodied person (Zmitrewicz et al., 2007). This produces a small A2 burst which the S.A.C.H (Solid Ankle Cushioned Heel) foot cannot produce. Normally, there would be no A2 contraction from the amputee population as this joint and the muscles that are associated with it have been removed. This lack of plantarflexor activity is taken up by the hip flexor muscles through an increase in the H3 power burst where most of their forward propulsion originates due to the hip flexors pulling the leg forward in the sagittal plane in relation to the trunk. However, as previously mentioned the push-off power may not be directly contributing to forward trunk propulsion but by the deceleration of the mass of the swinging limb. With amputees, the mass of the lower limb has been radically changed. According to Dempster's Tables, the shank and foot segment of a male is 6.18% of the total body mass, therefore for a 75 kg (165 lbs.) male, the shank segment should weigh 4.64 kg (10.20 lbs) and a prosthesis has an average weight of 0.76 lbs (Ossur, 2009) to 1.08 lbs (OWWCO, 2009), for a flex foot and a S.A.C.H. foot respectively. From the moment equation of: moment = mass * velocity, as the mass has been reduced the overall moment of this segment is reduced and therefore this deceleration of the swinging limb will not provide enough energy for adequate forward progression of the trunk (Lehmann, 1993; Winter & Sienko, 1988). The knee moment of force for K1 and K2 power bursts during amputee gait is rather low and has been reported as being close to zero for the first half of stance, the K4 power burst was noted as being small or negligible as it was not necessary since the moment being created about the knee is smaller as the prosthesis mass is significantly less than the normal mass of the shank-foot segment. The K3 power burst is normal during the push-off and early swing phases (Winter & Sienko, 1988). This absence of the first two power bursts, which would normally produce an extensor moment, have been investigated through EMG of both the knee flexors and extensors

and determined that both sets of muscles were hyperactive which indicated that co contraction at the knee joint is occurring. This co-contraction produced via the rectus femoris, vastus lateralis, semitendinosus and biceps femoris, has been hypothesized to be a stabilizing adaptation of the knee joint during the weight acceptance phase of the gait cycle. This stabilization of the knee joint will allow for a transfer of energy either to or from the trunk to aid in forward propulsion. This knee collapse is also prevented by the strong hip extensor burst by restricting the forward rotation of the thigh in relation to the pelvis but this also activates the knee flexors as the can cross both joints. If unchecked this will result in knee flexion and therefore the knee extensors are required to activate to cancel out this moment (Winter, 1991; Winter & Sienko, 1988). This hip extensor strategy is also associated with a forward lean adaptation which acts as a counter to the extensor load being applied and has been theorized to further stretch the hip extensors to produce a larger amount of power as previously mentioned (DeVita & Hortobagyi, 2000). This type of activation pattern is noted to be found in other pathologies.

4.3 Conditions that compromise the ankle joint

While the ankle joint has a very prominent role in gait there are a number of conditions that either limit or completely disable its use. This leads to individuals having to develop alternate patterns and strategies to allow them to self-ambulate. Rheumatoid Arthritis is an inflammatory disease that attacks the cartilaginous tissue of the joints breaking the smooth surface down to leave a rough, abrased surface. This will lead to structural deformities in the joints that are affected. The inflamed and painful joints not only lead to a local restriction of motion but due to the altered uses and forces being applied can cause abnormalities in neighboring joints as well. The subtalar joint is a very common site where this problem occurs and is frequently treated by orthopedic interventions (Weiss et al., 2007). Hindfoot arthrodesis (surgical fixation of the ankle

joint (Perry, 1992))is a typical intervention used and has been shown to improve the range of motion of the ipsilateral knee joint due to the added stability and decrease in localized pain from the affected ankle joint (Weiss et al., 2007). This allowed for a closer to normal knee function throughout the gait cycle as well as an increase in hip extension and knee-flexion ROM after surgery. This increase in ROM allows for a statistically significant improvement of hip flexion, hip extension, and knee flexion moments. However, the arthrodesis did not statistically improve the ankle dorsiflexion or plantarflexion angles when compared to pre and post operation (Weiss et al., 2007). The increase in excessive dorsiflexion by fusing the ankle and by obstructing the normal plantarflexion that occurs leads to an increase in heel rocker action. This free fall of the foot now carries the tibia over the ankle joint resulting in a flexing of the knee at the same rate, rather than normal which is at half the speed (Perry, 1992). This requires a higher demand from the quadriceps to counteract this forward rotation. This is not unlike the muscular demands that are placed on a trans-tibial amputee, who uses the aid of a S.A.C.H. foot to ambulate (Perry, 1992). Hemiplegic gait (paralysis of either the right or left leg (Perry, 1992)) is commonly noted in post stroke patients and poses the challenge of relearning how to walking with impairments of static postural stability. This can include uneven weight distribution in quiet stance, with less weight being placed on the affected leg and an increase in sway when compared to healthy individuals (Chen et al., 1999; Wang et al., 2005). To correct for this deviation from the norm, AFOs are frequently prescribed and assist with ankle joint angle, an increase in walking speed and, a reduction in energy expenditure during gait (Chen et al., 1999). The mean body weight distributions between the two legs is not significant as the affected leg supported 44.21% while wearing an AFO and 41.72% while not wearing an AFO (Chen et al., 1999), while this asymmetry is small, it is causes a change in motor patterns which have become quite evident in

the overall gait cycle. The added benefit of the AFO is weight bearing has been hypothesized to be due to the increase in ankle stability by the increase in good alignment and the addition of external support (Chen et al., 1999; Wang et al., 2005). This increase in weight bearing has shown to be essential in gait as it will allow the contralateral leg to be moved and ultimately a step to be taken. However, since an AFO is semi-rigid this does not allow for a smooth translation of forces across the foot and causes a more rapid midstance period than reported in healthy individuals and a faster forward rotation of the tibia over the ankle joint (Wang et al., 2005). This rigidity may also cause a dampened foot slap causing the knee flexors to work harder, similarly to prosthetic gait.

Cerebral Palsy is commonly found in children as it is a non-progressive paralysis due to a brain injury near the time of birth (Perry, 1992). One of the effects of this disorder is spastic diplegia (a paralysis involving both lower limbs) in the lower extremities (Abel et al., 1998). This spasticity along with motor deficits can produce a gait pattern which is characterized by an excessive plantar flexion through the swing phase and at floor contact, an exaggerated stance phase knee flexion, and an increase in hip adduction and internal rotation (Abel et al., 1998; Perry, 1992). The aforementioned characteristics causes a high impact force in early stance, which has been shown to be reduced through the use of an AFO (Abel et al., 1998). The AFO also controls the amount of plantar flexion through the swing phase to allow adequate toe clearance instead of dropping the foot below horizontal and having toe drag occur. Without the adequate toe clearance in swing an increase in hip flexion or hip hiking is used to elevate the limb and the foot to achieve this clearance (Perry, 1992). This increase in plantarflexion throughout stance as well as the reduced plantarflexion moment and power generation result the elimination of heel strike, midstance dorsiflexion and metatarsal roll-off. This will effectively reduce the stride length of an

affected individual (Abel et al., 1998). The use of an AFO with these individuals allows for a reduction in the early stance ankle power burst and an increase in the late stance ankle moment. These two characteristics allow for a gait pattern which is closer to the normal population.

In a healthy population, ankle braces have been used to examine the effects on postural control during one-legged and two legged stance. The use of ankles braces, taping and, elastic bandage wraps have been investigated and the use of ankle braces has a significant detrimental effect on the control of posture during quiet standing (Bennell & Goldie, 1994). It was concluded that many supports not only limited the extremes of the ROM as designed to, but also limited the motion necessary for normal functional movement. Two possible explanations have been hypothesized: firstly, that the proprioceptive input that is normally induced by ankle motion and is important for postural control was decreased due to a result of the ankle supports and their restrictive properties (Bennell & Goldie, 1994). Secondly, that the braces by limiting the ankle joint movement, it has physically prevented a normal ankle strategy from being used in maintaining one-legged stance (Bennell & Goldie, 1994; Wang et al., 2005).

4.4 Alternatives to dorsiflexion

The use of AFO with the disabled population is rather large in comparison to the investigations which utilize the healthy population to obtain data on the occurrence and mechanisms that occur during gait. Dorsiflexion is a key mechanism in the gait cycle to allow for a controlled roll-off of the tibia over the ankle and to prevent the toe from coming in contact with the ground during the middle of the swing phase, which will cause a trip. During normal gait with healthy individuals this toe clearance is primarily obtained through either a hip hiking strategy, knee flexion strategy, or a combination of the two. This knee flexion strategy can be achieved by decreasing the K3 power burst, which normal prevents back kick through an eccentric contraction of the knee

flexors. This decrease will allow for the swing leg to become more flexed and allow for more toe clearance (Winter, 1991, 2004). During an increase in walking speed, it has been noted that all three Hip power bursts do increase, not only as a increase to the forward propulsion but to allow adequate toe clearance during swing, as the risk of tripping/falling is much greater at these speeds (Winter, 1991). Due to a lack of plantarflexion just prior to toe-off the elderly use an increase in the H3 power burst, hip flexion, to create the momentum needed in the limb so that it can be transferred to the trunk and create propulsion. This increase in the H3 power burst can also be used to elevate the limb to achieve a level of toe clearance. While the use of AFOs has primarily been used in a disabled population, one study (Bennell & Goldie, 1994), determined the contributions and limitations of ankle braces to a healthy population. They determined that the ankle braces had a significant detrimental effect on the postural stance of subjects. These effects consisted of having a larger total area of oscillations in the medial-lateral of the center of gravity. Also, in the one-legged condition, the frequency of non-support foot touchdowns was counted. These were significantly higher during the ankle bracing and ankle taping condition then during the condition with no ankle restraints. This study did not however, examine the effects of braces during gait. To this date, no studies have been completed as to the muscle patterns and altered contributions of proximal joints due to restrictions of the ankle joint.

5.0 Methodology

5.1 Subjects

Fifteen healthy participants (8 male and 7 female) were recruited from the University population, aged 18 – 30 years old to participate in the current study. All participants had the protocol, study procedure and methods explained to them and signed a consent form which was approved by the Office for Research Ethics of the University of Waterloo. All subjects were screened for lower body injuries and/or surgery in the past year. This was done verbally and the participant only continued once cleared. Participants were made aware that they could withdraw from the study at any time.

5.2 Subject preparation

Anthropometric measurements were obtained which included height, weight, age, the width between the right and left greater trochanter and depth of the pelvis (midpoint of the ASIS to the midpoint of the PSIS). The mean height, weight and, age of all participants were documented in Table 1.

Table 1: Mean height, weight, and age of all participants

Participant Group	Height (m)	Weight (kg)	Age (years)
Males	1.79 (± 0.07)	81.3 (± 8.4)	23 (± 4)
Female	1.65 (± 0.10)	59.2 (± 7.0)	24 (± 2)
All Participants	1.73 (± 0.11)	71.0 (± 13.6)	24 (± 3)

The AMM was securely fastened to the participant's ankle and was placed under the foot and inside the shoe. A Velcro strap that was fastened to the AMM was wrapped around the participant's shank. A second Velcro strap was wrapped around the participant's forefoot and

two canvas straps connected the top of the AMM to the Velcro strap around the toes to further limit plantar flexion (see Figure 4). All participants were required to wear their own running shoes for the experimental protocol. The AMM was attached to the right foot, left foot or both feet depending on the condition being performed. Each participant was given a practice period, which consisted of approximately 10-12 walking trails across the 6 m walkway at a self selected speed. This practice period was to allow the participants to become comfortable with the AMM(s) and their ability to use them.

Figure 4: Modified Ankle-fixation Orthotic

5.3 Motion capture

Each subject was instrumented with 24 Optotrak infrared sensors (Northern Digital Incorporated, Waterloo, ON) to allow for an accurate representation of the skeletal movements. Four sensors were attached in a cluster to each rigid plate. The rigid plates were located on the right side of the body for the shank and thigh, as well as the lower back (pelvis). These rigid plates consisted of tracking markers, which were calibrated to each anatomical segment, this allowed them to be positioned optimally so as not to interfere with the AMM or the EMG electrodes. Calibration markers were used to locate the following landmarks with respect to each segment: the fifth metatarsal, the first metatarsal, forefoot, lateral calcaneous, lateral malleolus, medial malleolus, lateral condyle of the femur, medial condyle of the femur, right and left greater trochanters, and the right and left iliac crests. An Optotrak 3020 system (Northern Digital Incorporated, Waterloo, ON) including 3 camera banks each consisting of three lenses was used for marker capture. These camera banks were located around the collection area (see figure #4) to insure that a volume of at least three full strides was collected, in which the middle stride incorporated the force plate and was the one being used for analysis. The capture volume was calibrated before each subject arrived and the origin was located on the right posterior of the force plate that is being used. Each participant was asked to complete a standing calibration trial before the walking trials began. The Optotrak data was collected at 64 Hz in order to be properly synchronized in time with the EMG and Force plate data. This also satisfied the Nyquist theorem of having a sampling rate of greater than two times the frequency of the signal.

5.4 EMG collection

Electromyography (EMG) was collected during each trial to determine the contributions of each muscle to the forward progression of walking and limb trajectory. Each area was shaved/ abrased

and cleaned with alcohol to increase the adherence of the electrodes to the skin. EMG surface electrodes consisted of pairs of 2.2 cm diameter, bipolar silver-silver chloride electrodes were placed on the muscle bellies at each of the corresponding muscle groups: tibialis anterior, medial head of the gastrocnemius, rectus femoris, biceps femoris, gluteus maximus, gluteus medius, the right rectus abdominus and, the right erector spinae. A ground electrode was placed on the medial condyle of the right knee. Each pair of electrodes were placed over the muscle belly of that particular muscle and had a spacing of 2.2 cm between centers (Zipp, 1982). A Bortec system was then used to collect the EMG data and the raw signals were amplified and bandpass filtered (10 – 1000 Hz), after which it was collected on computer via the OPTOTRAK Data Acquisition Unit (Northern Digital Incorporated, Waterloo, ON). EMG data were collected at 2048 Hz to eliminate an aliasing of the signal as well as to keep in accordance with the Nyquist theorem as the electrical muscle activity could occur in the range of 10 – 1000 Hz.

5.5 Force Plate Collection

An AMTI Force Plate 18" by 20" (45.72 cm by 50.8 cm) (AMTI, 2009) that was embedded and flush with the floor was used to calculate the ground reaction forces (Fx, Fy, and Fz)and moments (Mx, My, and Mz) that were being applied to the foot. These forces and moments were then used along with the kinetic/kinematic data to calculate the moment about each of the joints in the leg. These data were collected at 2048 Hz using the computer via the OPTOTRAK Data Acquisition Unit (Northern Digital Incorporated, Waterloo, ON).

5.6 Photographs or Video Recording

If the participant gave consent, photographs or video were taken during the study session. These images allow the researcher to recall how the task was performed as well as to verify movement data. Photographs and video can be very useful in teaching and presenting the study in a

scientific paper or a conference presentation. All images were focused on the participant's entire body with the face and/or eyes blacked out to conceal the identity of the participant.

5.7 Experimental set up

The three banks of Optotrak cameras were set up around the collection area in a manner that allowed the IRED markers to be seen by at least one camera at all times. This collection volume was centered about the force plate. The force plate was in the center of the 6 meter walkway which allowed the participants to reach a constant velocity and stride length before reaching the force plate and also allowed them to continue off the plate for another two strides without having to stop or deviate from their gait patterns. The same set up was also used for all conditions of normal and obstructed walking; each with and without an AMM.

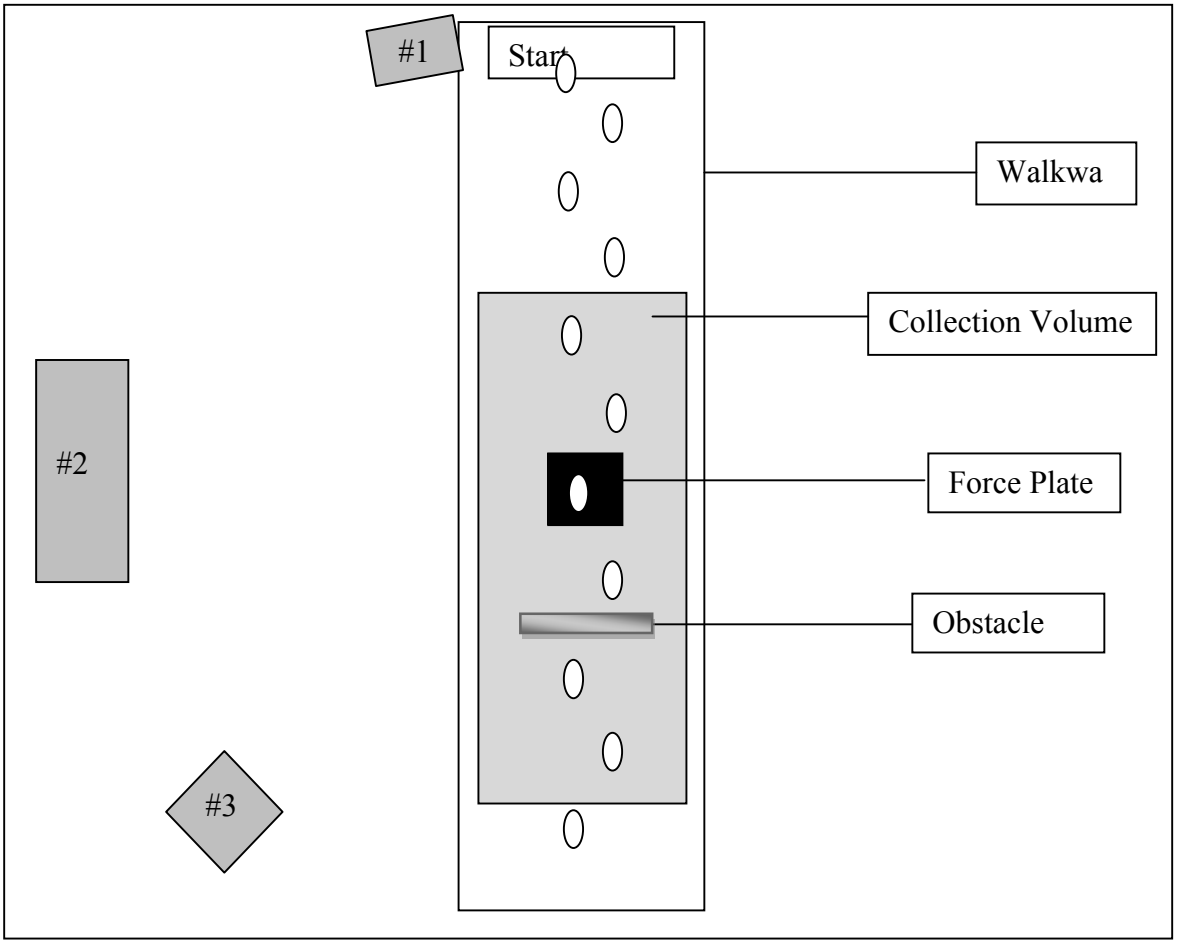

Figure 5: Laboratory Set-up for both obstacle and normal walking trials. For normal walking trials, the obstacle was removed to allow for a continuous gait

5.8 Testing Protocol

Once the participant was fitted with the AMM(s), IRED markers, and the EMG electrodes the participant was instructed to stand on the force plate for a calibration trial. A resting trial was also collected to determine and eliminate any existing bias from the muscular activity. After this trial, a starting point was determined to allow the participant, while walking normally, to land with their right foot on the force plate during a normal stride. This allowed for a precise

determination of heel strike as well as ground reaction forces and moments about the right foot which could then be determined in turn for the ankle, knee and the hip. The participant was then instructed to walk across the 6 meter walkway at a self selected speed, in all four conditions of wearing the AMM on the right foot, left foot and, both feet and not wearing the AMM. For half of these trials, a 30 cm obstacle was placed on the walkway in accordance with mid-swing for each participant. Trials were only included if the participant stepped on the force plate; and during obstacle trial successfully cleared the obstacle, until a total of at least 10 trials had been completed for each condition. The conditions were block randomized for each participant; in that all trials of a particular condition were completed at the same time before moving on to the next condition. The participant was allowed to rest between trials at their discretion. After each condition consisting of 10 trials had been completed, the starting point was again determined to account for any discrepancies in stride length between the conditions. The total testing time including set up and calibration was approximately 2.5 hours.

6.0 Data Analysis

All EMG, motion capture and, force plate data, were labeled and analyzed in Visual 3D (C-motion Inc, 2009). From the marker coordinates, a skeletal representation of the lower right side of the body including a full pelvis was constructed for each individual using their specific anthropometric data. This included the participant's height, weight, depth of the pelvis (measured from the midpoint of the ASIS to the midpoint of the PSIS) and, the width of the hips (measured as the distance between the right and left greater trochanters). From the skeletal reconstruction, the gait cycle was able to be monitored and the middle stride, determined as one full gait cycle (heel contact of the right foot to the next heel contact of the right foot) was used for the analysis and to normalize the time of the data. The EMG data had already been amplified and band passed filtered (10 – 1000Hz). The mean amplitude of the electrical activity was determined for each muscle and this bias level were removed from the signal. After this, the signal was full wave rectified and low pass filtered at 3 Hz using a second order Butterworth dualpass filter. Each muscle pattern was also normalized in time to a single gait cycle to allow for comparison between subjects. A two point normalization method was used to determine one full gait cycle of right heel contact to the subsequent right heel contact. The kinematic data allowed for a comparison between the conditions in relation to joint angle (ROM), moment and, powers.

6.1 Statistical analysis

The following values were determined for each trial collected:

- Ankle, knee, and hip peak angle
- Ankle, knee, and hip peak moment
- Ankle, knee, and hip peak powers
- Work done during each power burst for the ankle, knee and hip
- Muscle activation levels of: gastrocnemius, tibialis anterior, rectus femoris, biceps femoris, gluteus maximus, gluteus medius, right rectus abdominus, and right erector spinae throughout one complete gait cycle

The ankle, knee and hip joint angles and moments were calculated on a time series scale from heel contact of the right foot to the next heel contact of the right foot. The peak joint angles and peak joint moments were determined as the highest point during the gait cycle. Also the range of motion (ROM) was determined for the joint angles. The ankle, knee and, hip power bursts (A1, A2, K1-5, and H1-3) were determined and the peak value for each burst was recorded. This peak values was determined as the highest value on the curve during that particular power burst. Each power burst was determined by the zero crossing on either side as a start and an end point. The area under the power curve for each power burst at all three joints was also calculated to give the work done by that joint during a particular power burst phase. These values were repeatedly determined for all 10 trials per the 8 conditions. These conditions were: normal walking (no AMMs), AMM walking bilaterally, AMM walking unilaterally (right and left side), AMM obstacle walking bilaterally and unilaterally, and obstacle walking without an AMM.

From these values a one way repeated measures ANOVA was performed for each of the dependent variables. The dependent variables are located in Table 2. The repeated measures

factor was that of the trials in each of the condition. The ANOVAs allowed us to calculate a value for each mean joint angle, moment, and each power burst for the three joints. A comparison of each dependent variable across conditions was done to determine if any significant differences existed. An alpha level of 0.05 was used and if significance was found a Tukey HSD test was performed to determine a significant difference between the pair wise comparisons.

6.2 Time Series Data

All time series data, which includes EMG, joint angles, joint moments and, joint powers were normalized as a percentage of the total gait cycle. Cycle time was defined as the right heel contact to the consequent right heel contact, where right heel contact was determined when the marker placed on the lateral calcaneous reached the lowest point of the cycle, therefore incorporating one complete stride. Toe-off tends to occur at 60% of this gait cycle in accordance with Winter (1991). The mean peak joint angle, mean peak joint moment, mean peak joint powers, for the ankle, knee, and hip during level walking, level walking with the AMM, obstacle walking and, obstacle walking with the AMM were calculated. This generated 29 measures per participant. These 29 measures were averaged across all participants to determine the overall group averages of each of these particular measures. These 29 group measures were used in the statistical analysis mentioned previously. The individual measures that were calculated are presented in Table 2.

Table 2: Specific values that were calculated throughout the gait cycle and used in the statistical analysis of the data.

Ankle	Mean peak angle	Mean peak moment	Mean Peak A1 Power burst	Mean Peak A2 Power Burst	Mean Work Done A1 Power Burst	Mean Work Done A2 Power Burst	
Knee	Mean Peak Angle	Mean Peak Moments	Mean Peak K1 Power Burst	Mean Peak K2 Power Burst	Mean Peak K3 Power Burst	Mean Peak K4 Power Burst	Mean Peak K5 Power Burst
	Mean Work Done K0 Power burst	Mean Work Done K1 Power Burst	Mean Work Done K2 Power Burst	Mean Work Done K3 Power Burst	Mean Work Done K4 Power Burst	Mean Work Done K5 Power Burst	
Hip	Mean Peak Angle	Mean Peak Moments	Mean Peak H1 Power Burst	Mean Peak H2 Power Burst	Mean Peak H3 Power Burst		
	Mean Work Done H1 Power Burst	Mean Work Done H2 Power Burst	Mean Work Done H3 Power Burst				
General Variables	Walking Velocity	Maximum Toe Height					

The joint angles, moments, and powers of the ankle, knee and hip were plotted for each condition for one gait cycle. The normal walking with the AMM condition were plotted and compared to the normal walking without the AMM. Two standard deviations were included from the normal walking condition (without AMM) to show any difference between the two conditions. Also, the obstacle condition was plotted and compared to the obstacle condition with the AMM. Again, two standard deviations were included from the obstacle walking (without AMM) to show any difference between the two conditions. If a deviation from the normal joint angle, moment or, power occurred; the phase of the gait cycle that it had occurred, the duration, and the amplitude of the activation patterns were noted. A significant difference occurred when the average muscle activation deviated outside of the range of two standard deviations, which was determined during the normal walking condition. A deviation outside of the normal variance

was recorded when the mean muscle activation strayed from the normal variance for more than four consecutive data points.

6.3 EMG Data

Electromyography data were amplitude normalized to the maximal electrical activation during the normal walking condition for each muscle individually. The muscle activation patterns were averaged over the ten trials per condition to produce 4 different muscle activation patterns per muscle for each individual participant. Therefore, a total of 16 different plots were produced of activation data. The muscle activations during the AMM conditions were plotted against the normal condition (±SD) for each individual participant to examine any changes in muscle patterns during the gait cycle. The muscle activations during the obstacle condition (±SD) were plotted against the obstacle condition while wearing the AMM for each participant individually to examine any changes in the muscle patterns. If a deviation from the normal muscle pattern occurs, the phase of the gait cycle that it has occurred, the duration, and the amplitude of the activation patterns were noted. A significant difference were determined if the average muscle activation deviated outside of the one standard deviation range, which were determined during the normal walking condition.

7.0 Results

The findings of the study were presented by each joint: Ankle, Knee and, Hip. Twenty-nine repeated measures ANOVAs were run on the variables in Table 2. As walking velocity would affect all trials, it is presented first and the maximum toe height is presented at the beginning of the obstacle section as this is where it has relevance.

7.0.1 Walking Velocity

During the walking trials, a change in the average walking velocity was noted between conditions. The average walking velocity was calculated as the mean progression of the right iliac crest marker during the full gait cycle that was being analyzed.

Table 3: Changes in Mean walking velocity between trial conditions. Values denoted by an "*" are significantly different from the normal condition.

AMM Placement	Mean Walking Velocity (m/s)	Percent Change
Normal walking (None)	1.35	
Ipsilateral Walking	1.30	Decrease 3.08 %
Contralateral Walking	1.32	Decrease 2.06 %
Bilateral Walking	1.29	Decrease 3.95 %
Normal Obstacle (None)	1.22	
Ipsilateral Obstacle	1.16	Decrease 4.92 %
Contralateral Obstacle	1.17	Decrease 4.10 %
Bilateral Obstacle	1.13*	Decrease 7.38 %*

A repeated Measures ANOVA was run on the mean walking velocity of all participants during the trials collected through all the different conditions and a significant difference was detected, therefore a post hoc Tukey HSD was employed and no significant differences across both the level ground walking ($F(3, 42) = 0.1707$, p-value 0.9156) were observed whereas for the obstacle walking ($F(3, 42) = 6.623$, p-value <0.0009) did have a significant difference between conditions (Figure 6). Winter defines the "natural" cadence over level ground as 105 ± 6

steps/min and having a stride length of 1.51m (step length of 0.75 m), this would result in a walking velocity of 1.33 m/s. The slow cadence being approximately 20 steps/min less than natural (85 steps/min) with a stride length of 1.38 m (step length of 0.69 m) would result in a slow walking velocity of 0.99 m/s, well below the velocities observed during the level walking trials. Therefore it is noted that a small decrease in speed may have affected the overall outcome but these changes were not significant and cannot be entirely responsible for the lack in power generation across the different conditions.

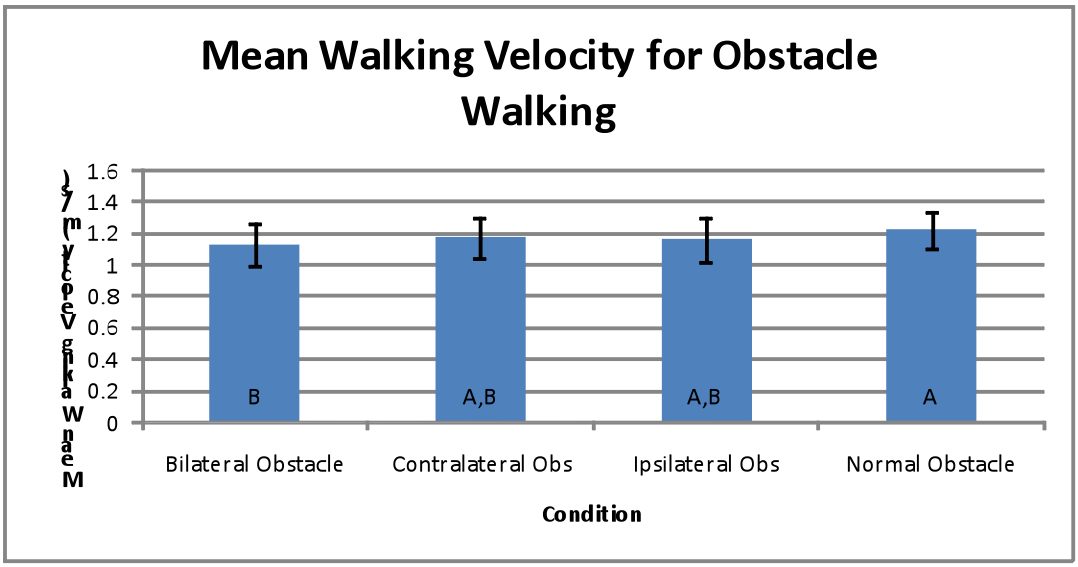

Figure 6: Mean walking velocity across Obstacle walking conditions. Levels with the same letter are not significantly different from each other.

7.1 Level Walking

The results will display the power curves for each joint. These power curves are a product of the moment and the angular velocity that are also displayed. The angular velocity can be determined as the slope of the line in the angle graphs. The amount of work done at each joint was determined as the area beneath the curve for that particular power burst. Since the A2 power burst occurs just prior to the swing phase to help generate the push-off force, the other power bursts that occur here, mainly the K3 and H3 power bursts were of the greatest interest as they would be the most likely candidates to compensate for the ankle. Therefore, the A1, A2, K3 and H3 power bursts are displayed below in their respective sections.

7.1.1 Ankle:

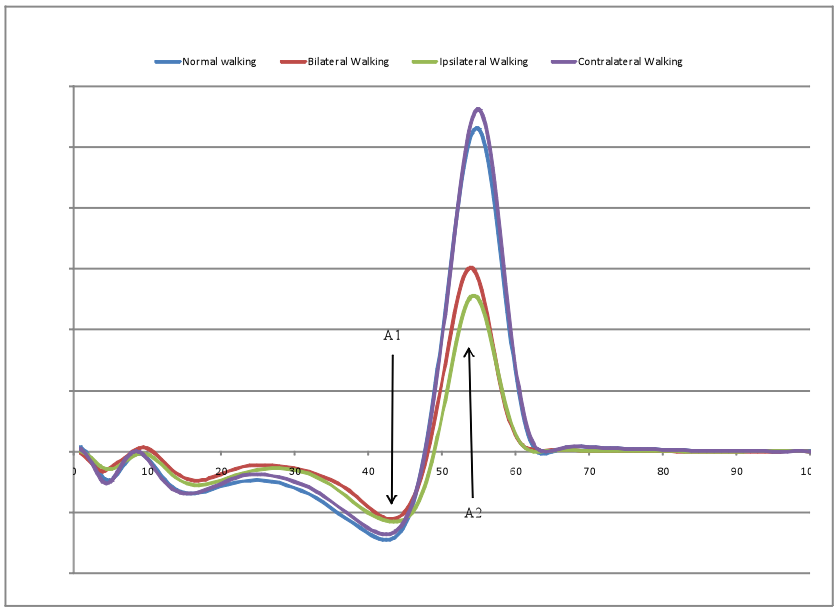

Figure 7: Mean Ankle Power Time series curves for the walking on level ground condition. A generation of power is represented by positive values and power absorptions are represented by negative values.

The peak power during the A1 power phase at the ankle showed a non-significant decrease during the AMM affected conditions in comparison to the normal walking condition (F (3, 42) =

0.3221, p-value =0.8014). The normal walking condition had an average value of -0.92 ± 0.28 watts/kg, while the bilateral AMM condition recorded -0.71 ± 0.15 watts/kg, the contralateral side of the unilateral condition was -0.91 ± 0.26 watts/kg and the ipsilateral side was -0.73 ± 0.18 watts/kg.

The average amount of work performed during the A1 power burst was calculated and determined to be significantly different while wearing vs. not wearing an AMM (F (3, 42) = 17.4512, p-value <0.0001). While not wearing the AMM the amount of work done was -0.18 ± 0.05 J/kg, while wearing the AMM bilaterally the amount of work done was -0.12 ± 0.04 J/kg, the contralateral side of the unilateral walking was -0.17 ± 0.05 J/kg and the ipsilateral side was -0.13 ± 0.03 J/kg (Figure 8).

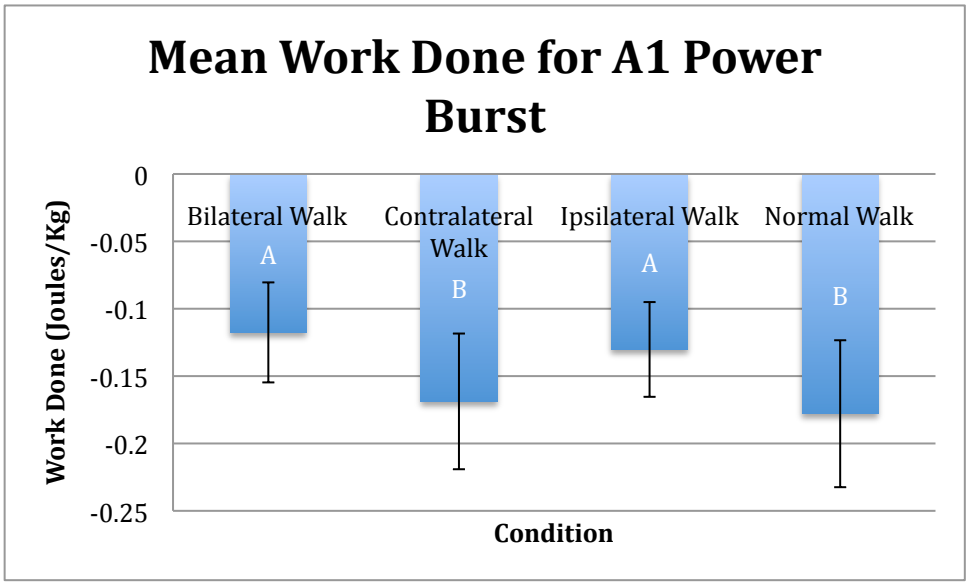

Figure 8: Mean work done during the A1 power burst across the 4 walking conditions. Levels not separated by the same letter are significantly different from each other.

The generation of peak power during the A2 power phase at the ankle showed a nonsignificant decrease during the AMM affected conditions in comparison to the normal walking condition (F (3, 42) = 0.3062, p-value = 0.8208). The normal walking condition had an average value of 3.08 ± 0.75 watts/kg, while the bilateral AMM condition recorded 1.75 ± 0.69 watts/kg, the

contralateral side of the unilateral condition was 3.10 ± 0.69 watts/kg and the ipsilateral side was 1.43 ± 0.37 watts/kg.

The average amount of work performed during the A2 power burst was calculated and determined to be significantly different while wearing vs. not wearing an AMM ($F(3, 42) = 69.485$, p-value <0.0001). While not wearing the AMM the amount of work done was 0.24 ± 0.05 J/kg, while wearing the AMM bilaterally the amount of work done was 0.13 ± 0.05 J/kg, the contralateral side of the unilateral walking was 0.25 ± 0.06 J/kg and the ipsilateral side was 0.10 ± 0.03 J/kg (Figure 9).

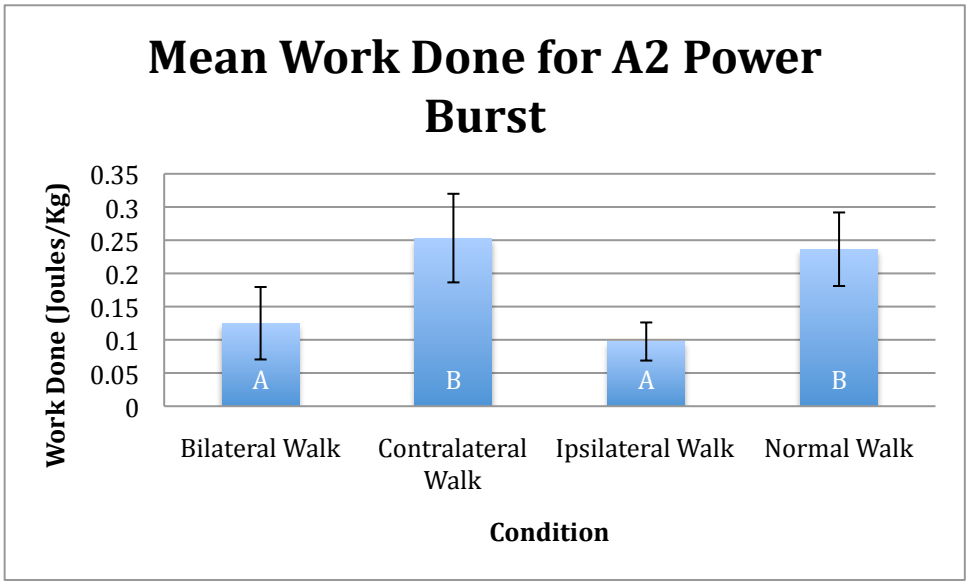

Figure 9: Mean Work Done during the A2 power burst across the 4 walking conditions. Levels not separated by the same letter are significantly different from each other.

The average ankle moment (Figure 10) did not significantly change across conditions. The peak ankle moments were: 1.64 ± 0.16 N*m/kg, 1.59 ± 0.21 N*m/kg, 1.60 ± 0.18 N*m/kg, and 1.67 ± 0.14 N*m/kg respectively for the Normal, AMM Bilateral, ipsilateral side and the contralateral side of the unilateral conditions.

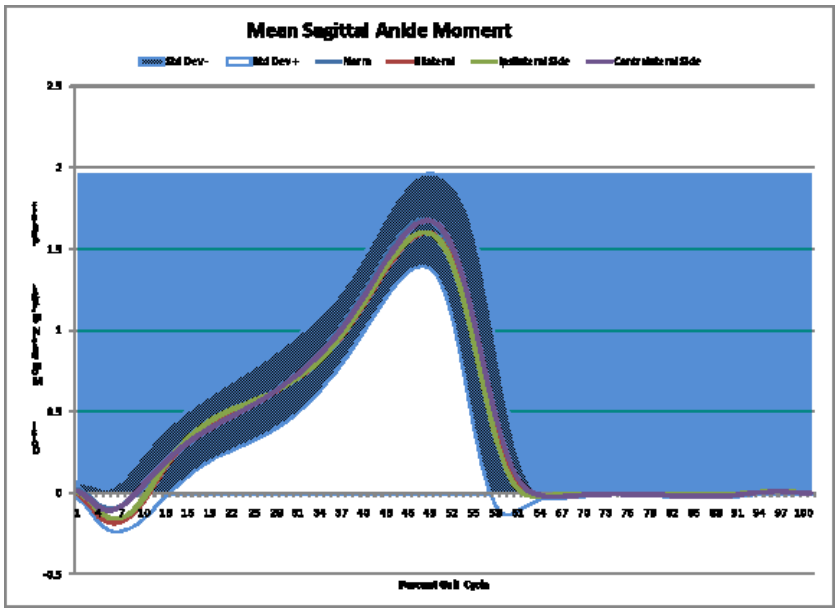

Figure 10: Mean Ankle Moment Time series curves for the walking on level ground condition. The shaded blue area represents two standard deviations of the normal (no AMM) condition. Where a plantarflexion moment is represented by a positive number.

The range of motion that the ankle joint was able to go through (Figure 11) was limited for the conditions including an AMM. The normal condition had an average peak ankle angle of 39.8 ± 6.4^0. The angle of contralateral side during unilateral AMM walking was 39.4 ± 7.0^0, the ipsilateral side during unilateral walking recorded an angle of 20.4 ± 6.2^0 and the bilateral AMM condition was 21.7 ± 5.9^0.

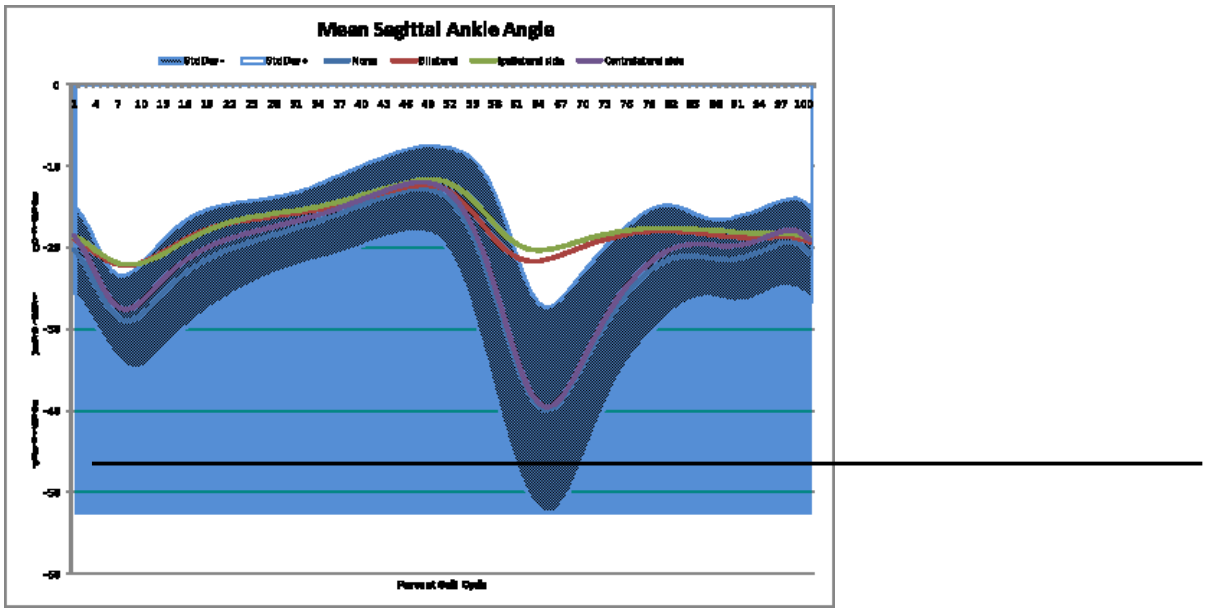

Figure 11: Ankle Angle Time series curves for the walking on level ground condition. The shaded blue area represents two standard deviations of the normal (no AMM) condition. The black line represents the ankle angle during the quiet standing position.

7.1.2 Knee:

The knee power bursts are all present during the four different conditions, while the magnitude of each varies. Since the K3 power burst is during the transition from stance to swing, this will be the focus for generating forward propulsion during the AMM conditions.

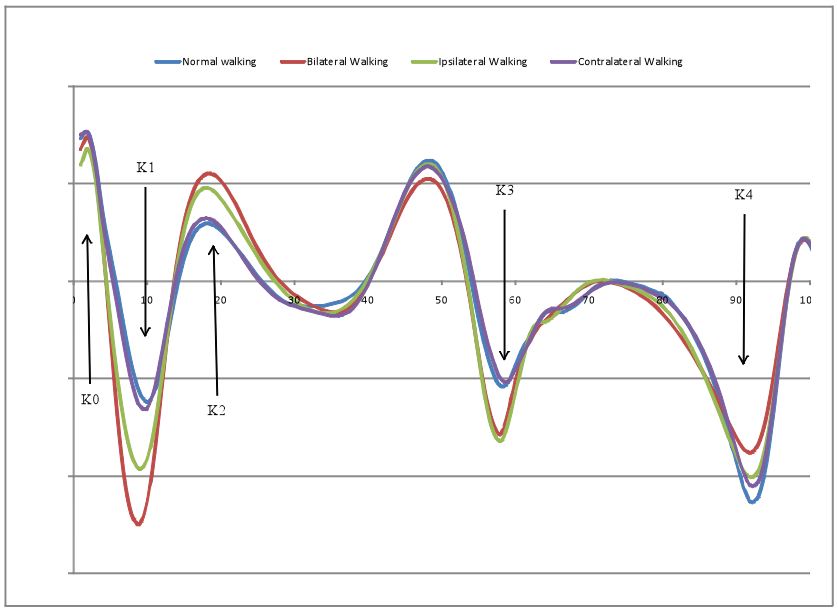

Figure 12: Mean Knee Power Time series curves for the walking on level ground condition.

Table 4: Mean work done for knee power bursts across conditions. Levels not separated by the same letter are significantly different.

			Means (Joules/kg)	STD Dev (Joules/kg)
K0-S Work	Bilateral Walk	B	0.021	0.010
	Contralateral Walk	A	0.030	0.015
	Ipsilateral Walk	B	0.020	0.007
	Normal Walk	A	0.028	0.012
K1-S Work	Bilateral Walk	B	-0.077	0.073
	Contralateral Walk	A,B	-0.041	0.056
	Ipsilateral Walk	A,B	-0.061	0.048

	Normal Walk	A	-0.038	0.054
K2-S Work	Bilateral Walk	A	0.054	0.051
	Contralateral Walk	A,B	0.028	0.048
	Ipsilateral Walk	A,B	0.044	0.041
	Normal Walk	B	0.025	0.050
K3-S Work	Bilateral Walk	A,B	-0.070	0.027
	Contralateral Walk	B,C	-0.055	0.027
	Ipsilateral Walk	A	-0.071	0.032
	Normal Walk	C	-0.054	0.019
K4-S Work	Bilateral Walk	A	-0.111	0.025
	Contralateral Walk	A	-0.111	0.015
	Ipsilateral Walk	A	-0.117	0.030
	Normal Walk	A	-0.114	0.018

Analysis of the conditions throughout the gait cycle showed that each power bursts was significantly different between the four different conditions except for the K4 power burst (Table 4). The average amount of work performed during each knee power burst was calculated and determined to be significantly different between numerous conditions K0 ($F(3, 42) = 8.9119$, p-value <0.0001), K1 ($F(3, 42) = 6.1043$, p-value $=0.0015$), K2 ($F(3, 42) = 3.4798$, p-value $=0.0241$), K3 ($F(3, 42) = 4.9844$, p-value $=0.0048$).

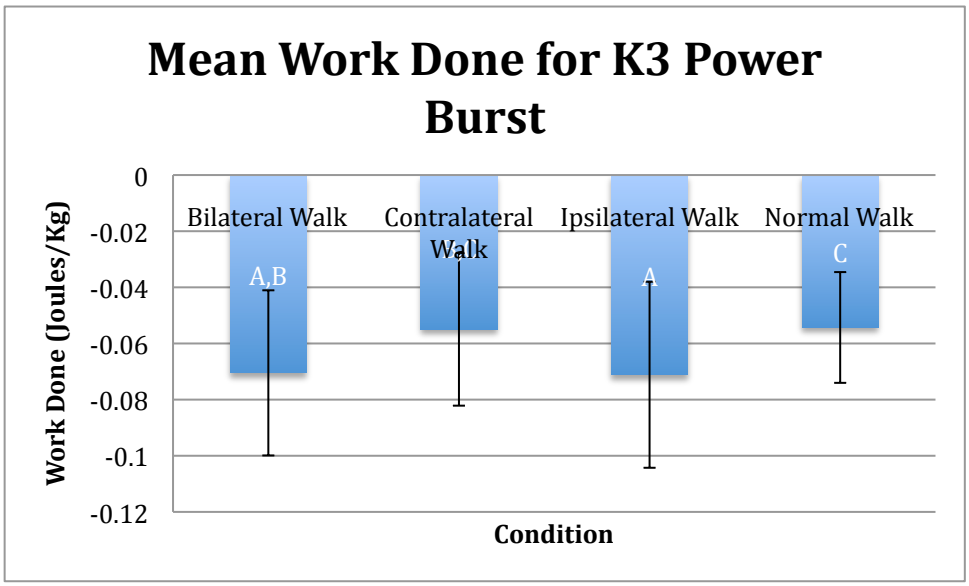

Figure 13: Average work done for the K3 Power burst across the 4 walking conditions. Levels not separated by the same letter are significantly different from each other.

Table 5: Mean Peak Powers for knee power bursts across conditions. Levels not separated by the same letter are significantly different.

			Means (Watts/kg)	STD Dev (Watts/kg)
K1-S Peak	Bilateral Walk	A	-1.26	1.19
	Contralateral Walk	A	-0.78	0.99
	Ipsilateral Walk	A	-0.99	0.74
	Normal Walk	A	-0.73	1.00
K2-S Peak	Bilateral Walk	A	0.60	0.53
	Contralateral Walk	A	0.35	0.56
	Ipsilateral Walk	A	0.50	0.38
	Normal Walk	A	0.34	0.59
K3-S Peak	Bilateral Walk	A	-1.00	0.38
	Contralateral Walk	A	-0.72	0.30
	Ipsilateral Walk	A	-0.99	0.31
	Normal Walk	A	-0.74	0.24
K4-S Peak	Bilateral Walk	A	-0.98	0.28
	Contralateral Walk	A	-1.22	0.21
	Ipsilateral Walk	A	-1.09	0.31
	Normal Walk	A	-1.28	0.21

Analysis of the conditions throughout the gait cycle showed that each peak power bursts was not significantly different between the four different conditions (Table 5).

The average knee moment did not significantly change across conditions (Figure 14), even though there was an increase in amplitude during the AMM trials. The peak knee moments were: 0.36 ± 0.38 N*m/kg, 0.68 ± 0.40 N*m/kg, 0.62 ± 0.32 N*m/kg, and 0.39 ± 0.39 N*m/kg respectively for the Normal, Bilateral AMM condition, ipsilateral side of the unilateral condition and the contralateral side of the unilateral condition.

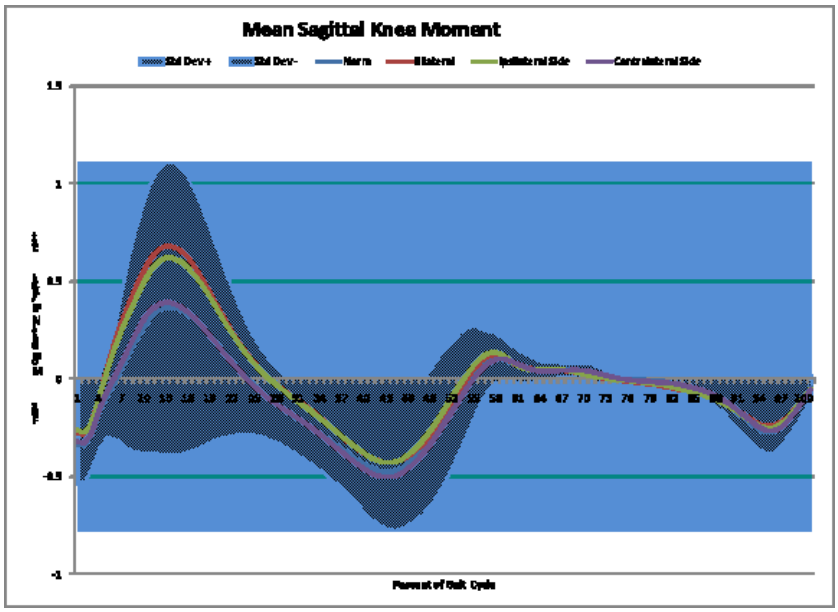

Figure 14: Mean Knee Moment Time series curves for the walking on level ground condition. The shaded blue area represents two standard deviations of the normal (no AMM) condition. Where knee extension moments are represented by a positive number.

The range of motion that was completed by the knee joint was very similar regardless of condition imposed upon the participant (Figure 15). The normal condition had an average peak ankle angle of 64.9 ± 4.0^0. The angle of the contralateral side during unilateral AMM walking was 64.3 ± 4.6^0, the ipsilateral side during unilateral walking recorded an angle of 69.6 ± 7.7^0 and the bilateral AMM condition was 67.6 ± 6.5^0.

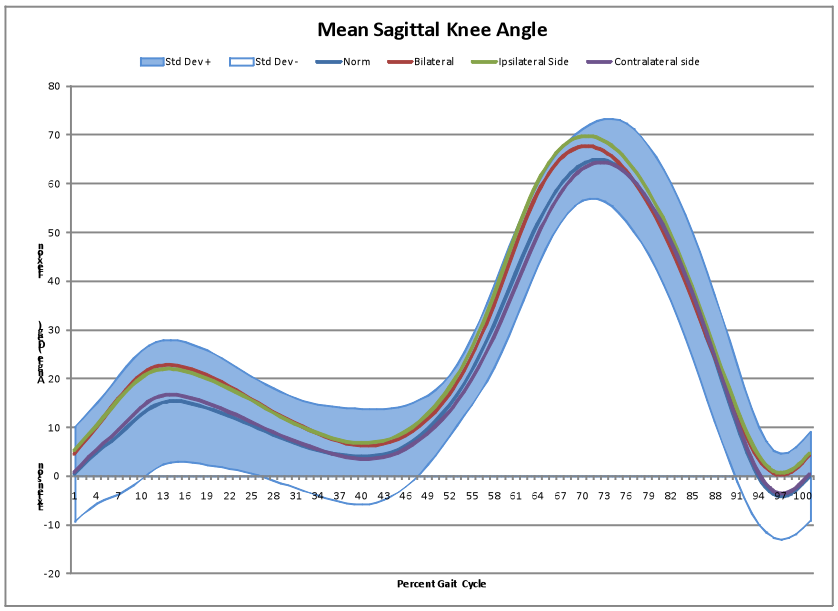

Figure 15: Knee Angle Time series curves for the walking on level ground condition. The shaded blue area represents two standard deviations of the normal (no AMM) condition.

7.1.3 Hip:

Only the amount of work done at the generating power bursts showed a significant difference across the conditions. The H3 power burst is shown below as this power bursts is expressed during the stance to swing transition and is the most efficient way for generating forward propulsion during the AMM conditions.

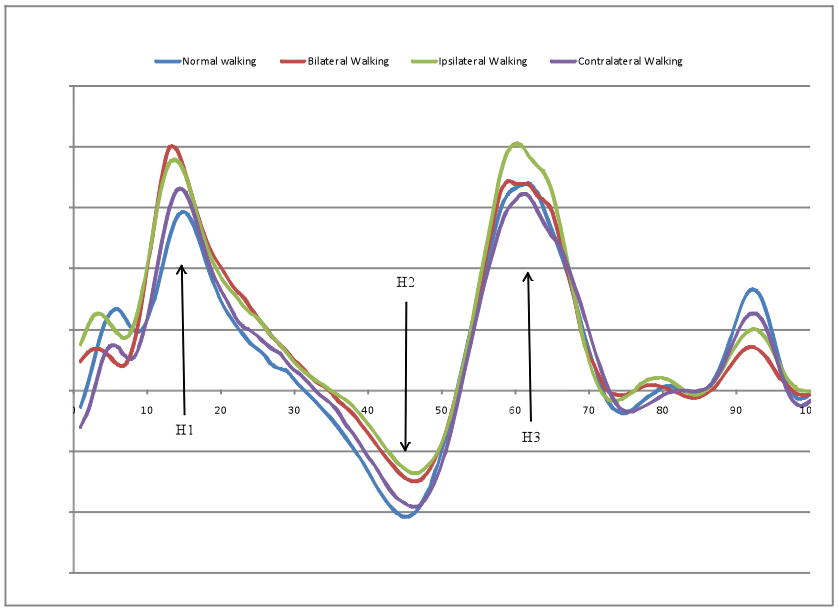

Figure 16: Mean Hip Power Time series curves for the walking on level ground condition.

Table 6: Mean Work for hip power bursts across conditions. Levels not separated by the same letter are significantly different.

			Means (Joules/kg)	STD Dev (Joules/kg)
H1-S Work	Bilateral Walk	A	0.13	0.06
	Contralateral Walk	A,B	0.11	0.05
	Ipsilateral Walk	A,B	0.12	0.05
	Normal Walk	B	0.10	0.05
H2-S Work	Bilateral Walk	A	-0.04	0.04
	Contralateral Walk	A	-0.06	0.05
	Ipsilateral Walk	A	-0.04	0.03
	Normal Walk	A	-0.06	0.06
H3-S Work	Bilateral Walk	A,B	0.10	0.03
	Contralateral Walk	B	0.09	0.03
	Ipsilateral Walk	A	0.11	0.04
	Normal Walk	B	0.09	0.03

Table 7: Mean Peak Powers for hip power bursts across conditions. Levels not separated by the same letter are significantly different.

			Means (Watts/kg)	STD Dev (Watts/kg)
H1-S Peak	Bilateral Walk	A	0.87	0.35
	Contralateral Walk	A	0.79	0.26
	Ipsilateral Walk	A	0.85	0.34
	Normal Walk	A	0.73	0.26
H2-S Peak	Bilateral Walk	A	-0.36	0.26
	Contralateral Walk	A	-0.45	0.37
	Ipsilateral Walk	A	-0.33	0.17
	Normal Walk	A	-0.48	0.37
H3-S Peak	Bilateral Walk	A	0.82	0.25
	Contralateral Walk	A	0.77	0.18
	Ipsilateral Walk	A	0.91	0.30
	Normal Walk	A	0.83	0.19

Analysis of the conditions throughout the gait cycle showed that the work done during the H1 and H3 power bursts was significantly different between the four different conditions (Table 6) .

The average amount of work performed during each hip power burst was calculated and determined to be significantly different between numerous conditions H1 ($F(3, 42) = 3.7348$, p-value <0.0182) and H3 ($F(3, 42) = 5.7455$, p-value $=0.0022$) (Figure 17). For the hip peak powers, no significant differences were found over any of the power bursts across the 4 different conditions (Table 7).

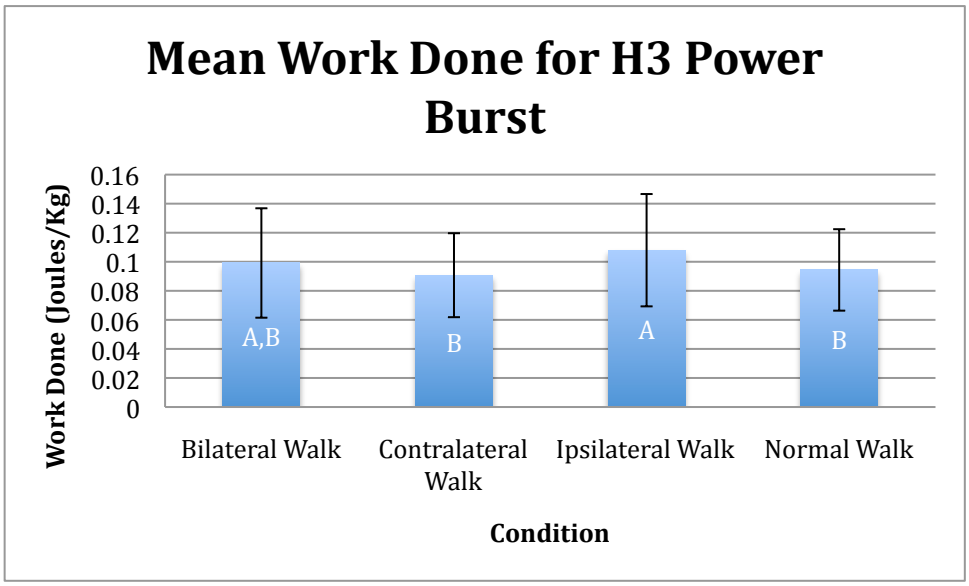

Figure 17: Average work done for the H3 Power burst across the 4 walking conditions. Levels not separated by the same letter are significantly different from each other.

The Mean hip moment did not significantly change across conditions (Figure 18). The maximum hip moments were: 0.69 ± 0.19 N*m/kg, 0.73 ± 0.23 N*m/kg, 0.70 ± 0.18 N*m/kg, and 0.72 ± 0.21 N*m/kg respectively for the Normal, bilateral condition, ipsilateral side and the contralateral side of the unilateral condition. While the minimum hip moments were: -0.54 ± 0.24 N*m/kg, - 0.47 ± 0.17 N*m/kg, -0.47 ± 0.13 N*m/kg, and -0.52 ± 0.25 N*m/kg respectively for order mentioned above.

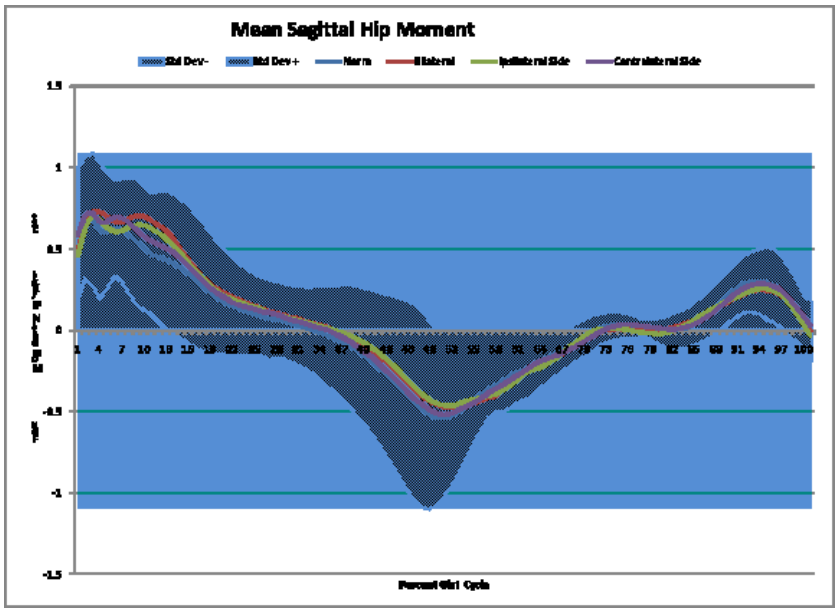

Figure 18: Mean Hip moment Time series curves for the walking on level ground condition. The light shaded blue area represents one standard deviation of the normal (no AMM) condition. A positive number indicates extensor moment.

The range of motion of the hip was a more accurate measure as the hip joint does not go through multiple peak phases like the other two joints (Figure 19). The normal condition had a ROM of 41.4 ± 9.1^0, the ROM of the contralateral ankle during unilateral AMM walking was 39.5 ± 8.3^0, the ipsilateral side during unilateral walking recorded an angle of 41.5 ± 8.3^0 and the bilateral AMM condition was 40.1 ± 8.9^0. These angles and ROM were all very similar across conditions.

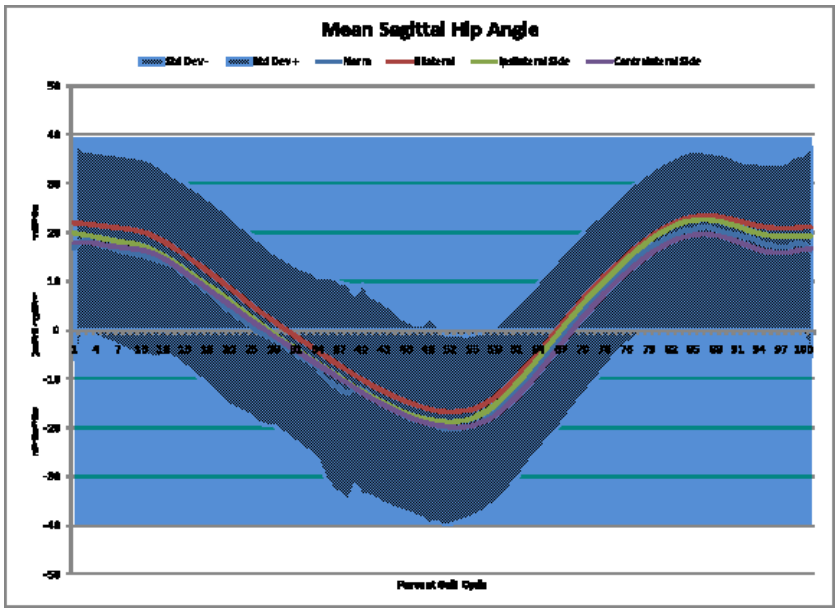

Figure 19: Mean Hip angle Time series curves for the walking on level ground condition. The light shaded blue area represents one standard deviation of the normal (no AMM) condition. A positive number indicates hip flexion.

7.2 Individual Participant's values

Each participant's temporal measures were calculated and the values were collapsed across trial for each condition within each participant (Table 8). Participants showed one of two trends during the bilateral condition in comparison to the normal walking condition. Six of the participants displayed an increase in cadence and a decrease in stride length during the bilateral condition, while having the walking velocity stay at a relatively constant value. These participants also showed an increase in the stance ratio and decrease in swing ratio.

Seven participants showed an opposing trend of a decrease in cadence and an increase in stride length from the bilateral condition in comparison to the normal walking condition. These participants tended to decrease the stance ratio and increase the swing ratio during the bilateral

condition. One participant (Female #1) showed a decrease in cadence while maintaining stride length, however, the walking velocity slightly increased between the two conditions.

Table 8: Mean temporal measures per participant collapsed over the individual level ground walking trials.

		Walking Velocity (m/s)	S.D.	Cadence (steps/min)	S.D.	Stride Length (m)	S.D.	Stance Ratio	S.D.	Swing Ratio	S.D.
Female #1	Normal Walk	1.33	0.08	104.81	2.03	1.61	0.04	0.57	0.01	0.43	0.01
	Bilateral Walk	1.34	0.05	80.68	4.94	1.59	0.04	0.43	0.02	0.57	0.02
	Ipsilateral Side	1.31	0.04	96.73	3.63	1.53	0.04	0.48	0.01	0.52	0.01
	Contralateral Side	1.26	0.03	106.42	2.38	1.37	0.02	0.61	0.01	0.39	0.01
Female #2	Normal Walk	1.20	0.03	86.27	2.70	1.87	0.04	0.47	0.01	0.53	0.01
	Bilateral Walk	1.21	0.03	114.44	4.26	1.41	0.05	0.58	0.01	0.42	0.01
	Ipsilateral Side	1.24	0.03	117.13	2.66	1.39	0.04	0.58	0.01	0.42	0.01
	Contralateral Side	1.21	0.04	84.00	2.41	1.56	0.03	0.46	0.01	0.54	0.01
Female #3	Normal Walk	1.28	0.02	108.39	2.76	1.80	0.05	0.58	0.01	0.42	0.01
	Bilateral Walk	1.17	0.05	77.14	1.69	1.58	0.05	0.42	0.01	0.58	0.01
	Ipsilateral Side	1.26	0.02	86.69	3.24	1.53	0.03	0.51	0.01	0.49	0.01
	Contralateral Side	1.27	0.03	103.21	3.73	1.35	0.03	0.59	0.01	0.41	0.01
Female #5	Normal Walk	1.37	0.05	85.22	2.69	1.49	0.03	0.45	0.01	0.55	0.01
	Bilateral Walk	1.41	0.04	112.33	2.47	1.37	0.02	0.60	0.01	0.40	0.01
	Ipsilateral Side	1.46	0.04	109.61	2.49	1.47	0.05	0.58	0.01	0.42	0.01
	Contralateral Side	1.35	0.04	80.67	4.87	1.58	0.04	0.45	0.02	0.55	0.02
Female #6	Normal Walk	1.43	0.05	112.52	3.14	1.32	0.02	0.58	0.01	0.42	0.01
	Bilateral Walk	1.35	0.06	79.40	3.58	1.58	0.03	0.44	0.02	0.56	0.02
	Ipsilateral Side	1.35	0.08	91.17	2.67	1.53	0.06	0.51	0.01	0.49	0.01
	Contralateral Side	1.35	0.03	108.02	0.86	1.40	0.03	0.59	0.01	0.41	0.01
Female #7	Normal Walk	1.64	0.05	96.01	3.67	1.42	0.03	0.47	0.01	0.53	0.01
	Bilateral Walk	1.53	0.04	113.65	2.11	1.39	0.03	0.59	0.00	0.41	0.00
	Ipsilateral Side	1.41	0.05	108.97	2.10	1.48	0.03	0.60	0.01	0.40	0.01
	Contralateral Side	1.56	0.04	83.62	2.24	1.56	0.02	0.45	0.01	0.55	0.01
Female #8	Normal Walk	1.32	0.03	119.51	3.05	1.32	0.04	0.58	0.01	0.42	0.01
	Bilateral Walk	1.23	0.04	81.41	5.14	1.69	0.03	0.43	0.02	0.57	0.02
	Ipsilateral Side	1.25	0.04	88.94	3.90	1.58	0.03	0.49	0.01	0.51	0.01
	Contralateral Side	1.30	0.05	109.57	1.37	1.38	0.02	0.58	0.01	0.42	0.01
Male #1	Normal Walk	1.19	0.03	89.25	2.32	1.47	0.03	0.46	0.01	0.54	0.01
	Bilateral Walk	1.01	0.04	113.68	2.88	1.42	0.04	0.60	0.01	0.40	0.01
	Ipsilateral Side	1.11	0.04	110.71	2.28	1.47	0.03	0.58	0.01	0.42	0.01
	Contralateral Side	1.13	0.03	85.43	5.13	1.53	0.03	0.45	0.02	0.55	0.02
Male #2	Normal Walk	1.23	0.03	111.99	2.09	1.34	0.02	0.58	0.01	0.42	0.01
	Bilateral Walk	1.29	0.03	88.72	1.85	1.58	0.04	0.49	0.01	0.51	0.01
	Ipsilateral Side	1.35	0.02	91.83	2.73	1.59	0.04	0.51	0.01	0.49	0.01
	Contralateral Side	1.33	0.02	110.05	1.63	1.39	0.01	0.58	0.01	0.42	0.01
Male #3	Normal Walk	1.32	0.04	98.81	3.13	1.41	0.03	0.47	0.01	0.53	0.01
	Bilateral Walk	1.28	0.05	103.11	1.76	1.40	0.03	0.58	0.01	0.42	0.01
	Ipsilateral Side	1.27	0.04	115.01	2.13	1.49	0.04	0.59	0.01	0.41	0.01
	Contralateral Side	1.22	0.04	100.32	2.49	1.51	0.03	0.47	0.02	0.53	0.02
Male #4	Normal Walk	1.38	0.05	119.65	1.72	1.33	0.02	0.57	0.01	0.43	0.01
	Bilateral Walk	1.38	0.06	83.75	2.82	1.50	0.05	0.50	0.01	0.50	0.01
	Ipsilateral Side	1.43	0.04	83.26	2.25	1.92	0.04	0.46	0.01	0.54	0.01
	Contralateral Side	1.39	0.04	117.10	1.91	1.43	0.03	0.56	0.01	0.44	0.01
Male #5	Normal Walk	1.32	0.05	69.36	1.95	1.58	0.03	0.47	0.02	0.53	0.02
	Bilateral Walk	1.12	0.05	105.09	1.70	1.39	0.03	0.60	0.01	0.40	0.01
	Ipsilateral Side	1.22	0.02	102.71	2.10	1.77	0.03	0.57	0.01	0.43	0.01
	Contralateral Side	1.27	0.02	100.19	2.45	1.52	0.02	0.48	0.01	0.52	0.01
Male #6	Normal Walk	1.69	0.03	91.81	2.34	1.31	0.03	0.61	0.01	0.39	0.01
	Bilateral Walk	1.61	0.05	88.76	2.77	1.61	0.03	0.49	0.01	0.51	0.01
	Ipsilateral Side	1.52	0.05	82.99	2.20	1.89	0.04	0.47	0.01	0.53	0.01
	Contralateral Side	1.63	0.04	115.93	3.48	1.39	0.03	0.58	0.01	0.42	0.01
Male #7	Normal Walk	1.21	0.03	82.99	2.24	1.62	0.01	0.51	0.01	0.49	0.01
	Bilateral Walk	1.15	0.04	105.67	1.82	1.40	0.02	0.59	0.01	0.41	0.01
	Ipsilateral Side	1.12	0.05	103.17	2.96	1.77	0.03	0.58	0.01	0.42	0.01
	Contralateral Side	1.24	0.02	103.55	2.18	1.53	0.03	0.47	0.01	0.53	0.01
Male #8	Normal Walk	1.26	0.01	100.94	1.46	1.36	0.03	0.63	0.01	0.37	0.01
	Bilateral Walk	1.29	0.05	84.17	3.61	1.53	0.02	0.48	0.01	0.52	0.01
	Ipsilateral Side	1.26	0.03	84.45	2.19	1.97	0.04	0.47	0.01	0.53	0.01
	Contralateral Side	1.26	0.05	119.83	1.38	1.44	0.03	0.57	0.01	0.43	0.01

7.3 EMG

Electromyography was collected from the following muscles: tibialis anterior, medial gastrocnemius, rectus femoris, biceps femoris, gluteus maximus, gluteus medius, rectus abdominus, and erector spinae. There were no remarkable changes outside of the normal

deviation that were consistent between the participants. Each participant did have their own adaptations between the conditions to manage with the constraints of the AMMs. An average of all of the participant's data was noted to "wash out" any of the individual variability. One representative sample of the muscle patterns was shown in the results section while two more samples for each muscle have been included in the appendix.

For the Tibialis anterior muscle (Figure 20), there was very little deviation outside of the normal variation. In general, the ipsilateral AMM condition shows a decrease in muscle recruitment during mid stance and into mid swing, whereas the contralateral side and bilateral conditions showed no deviation until late swing phase where there activation levels dropped just below.

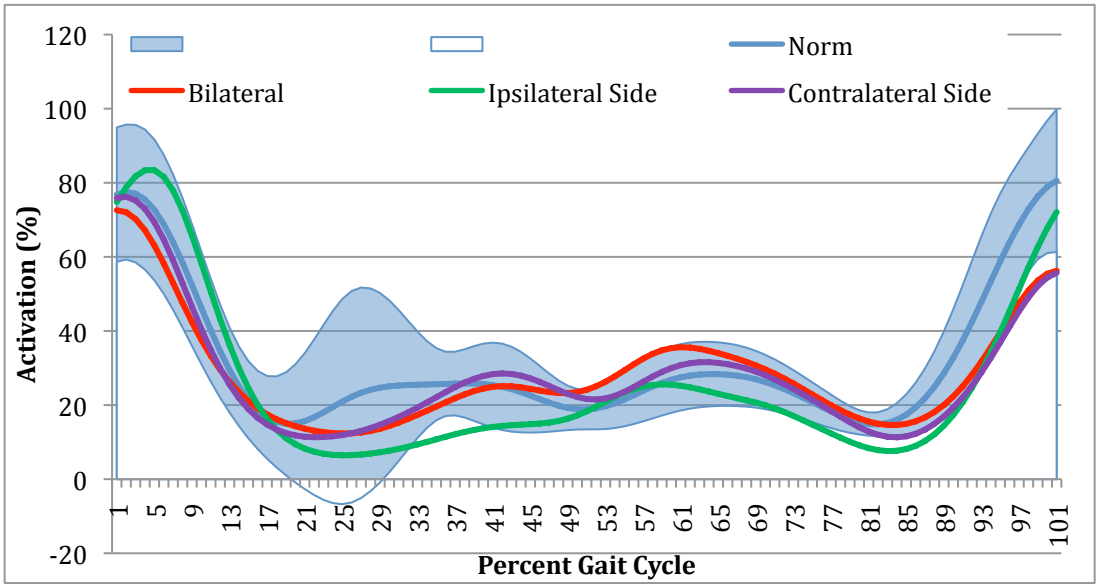

Figure 20: Mean Tibialis Anterior Time series curves for the walking on level ground condition. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

The gastrocnemius muscle (Figure 21) showed a decrease in activation for both the bilateral and ipsilateral side of the unilateral conditions throughout the entire stance phase of the gait cycle. While this decrease was not below the normal variation of the normal walking, it is still important to note as this was the specific use that the AMM was utilized for.

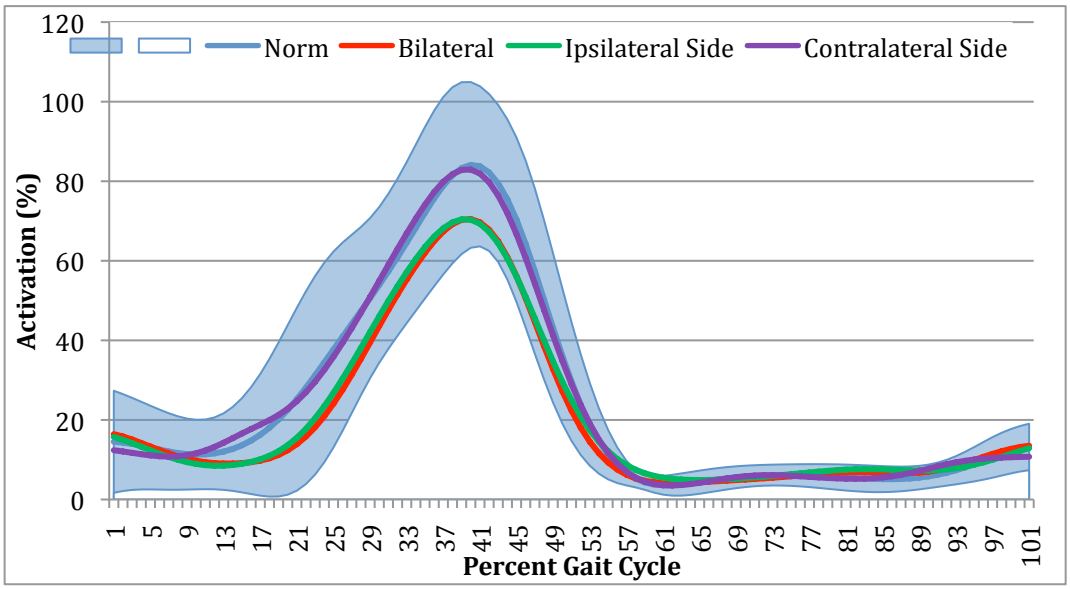

Figure 21: Mean medial Gastrocnemius Time series curves for the walking on level ground condition. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

The Rectus Femoris muscle (Figure 22) was noted to have an increase in activation for the leg in which the AMM was placed on, therefore bilaterally and the ipsilateral side. As seen below this increase in activation during toe off is just on the inside of the normal variation of the normal ground walking condition. During early stance, all three AMM conditions have a delay in deactivation and in reaching the steady level of mid-stance phase. Only the condition where an AMM was placed bilaterally does a deviation occur beyond the variation of the normal walking condition.

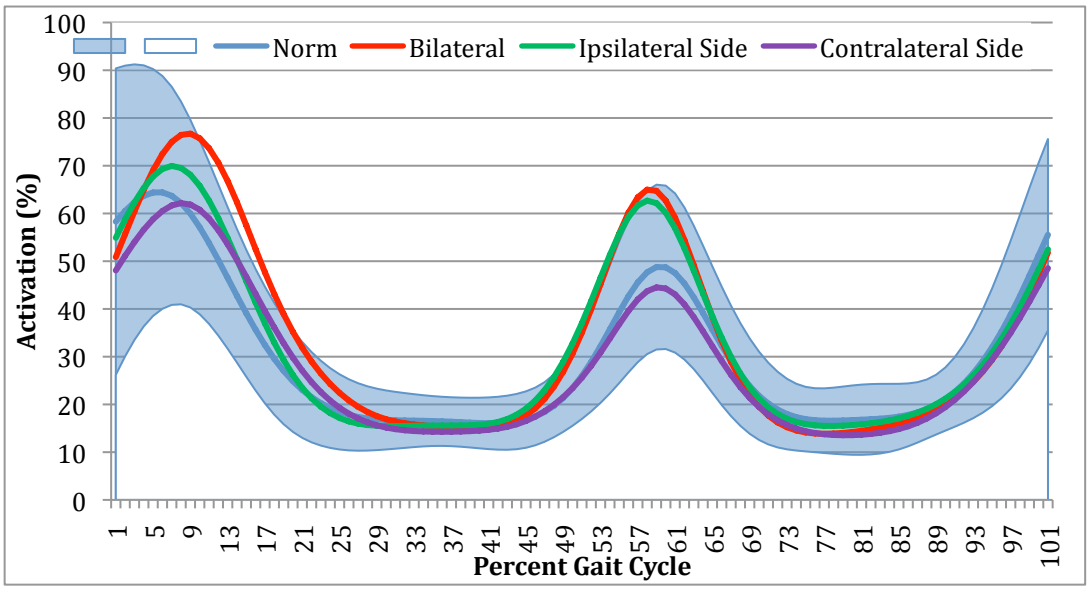

Figure 22: Mean Rectus Femoris Time series curves for the walking on level ground condition. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

The Biceps Femoris muscle did not deviate from the normal walking condition until swing phase (Figure 23). During swing phase, the contralateral leg to the AMM placement, the biceps femoris muscle actually had an increase in activation, which deviated just outside the normal variation during early swing and then again after the peak activation in late swing. The bilateral AMM condition and the ipsilateral leg seemed to have very little difference to the activation of this muscle throughout the gait cycle.

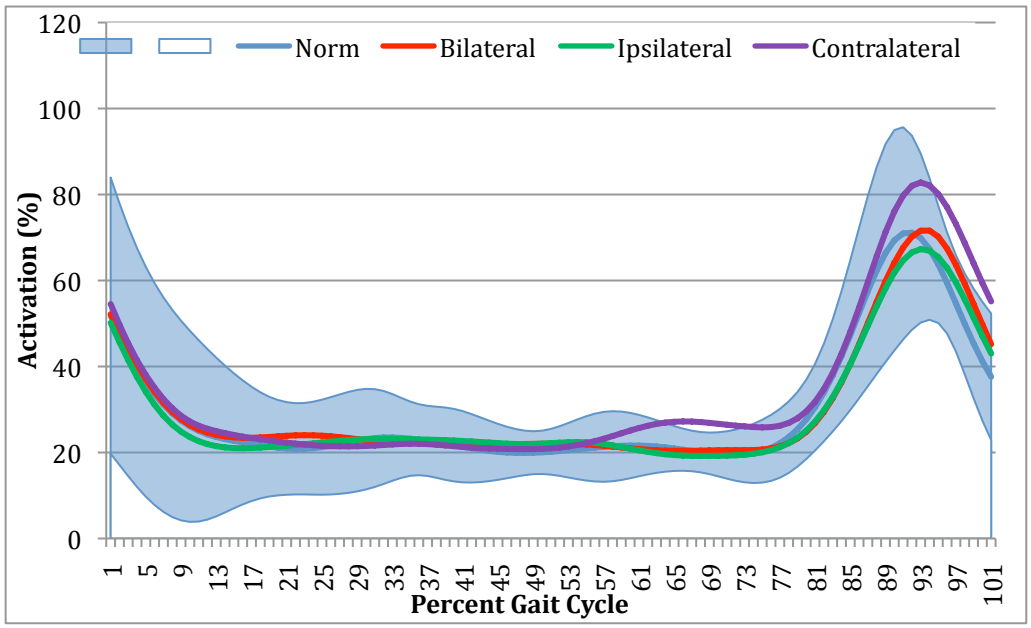

Figure 23: Mean Biceps Femoris Time series curves for the walking on level ground condition. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

The Gluteus Maximum muscle (Figure 24) held a fairly consistent pattern with an increase is activation occurring during early stance and around toe off during the gait cycle. The deviation of any of the conditions from the normal variation was limited to mid stance where the bilateral AMM and the ipsilateral conditions had the activation patterns rising above the normal activation levels at this time.

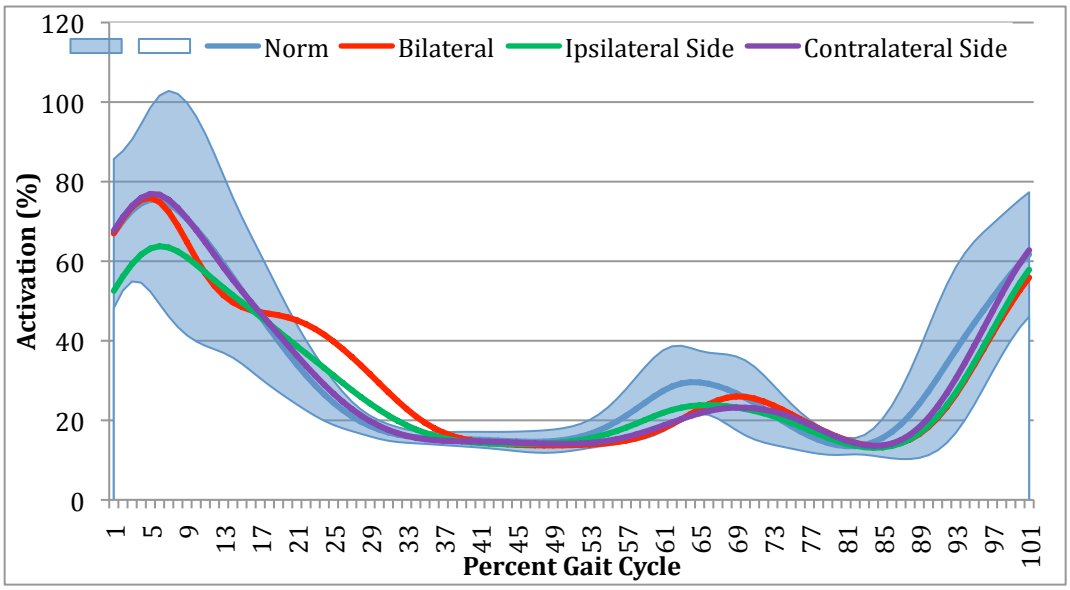

Figure 24: Mean Gluteus Maximus Time series curves for the walking on level ground condition. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

The Gluteus Medius muscle also showed relatively little changes across the conditions of AMM placements (Figure 25). A lower activation level during the beginning of early stance phase from the ipsilateral AMM condition was noted, however, this activation level did not drop below the normal range of variation which includes the normal walking trials. In mid swing, both the contralateral and both AMM conditions display a slight increase as well as a short duration in activation above the normal range.

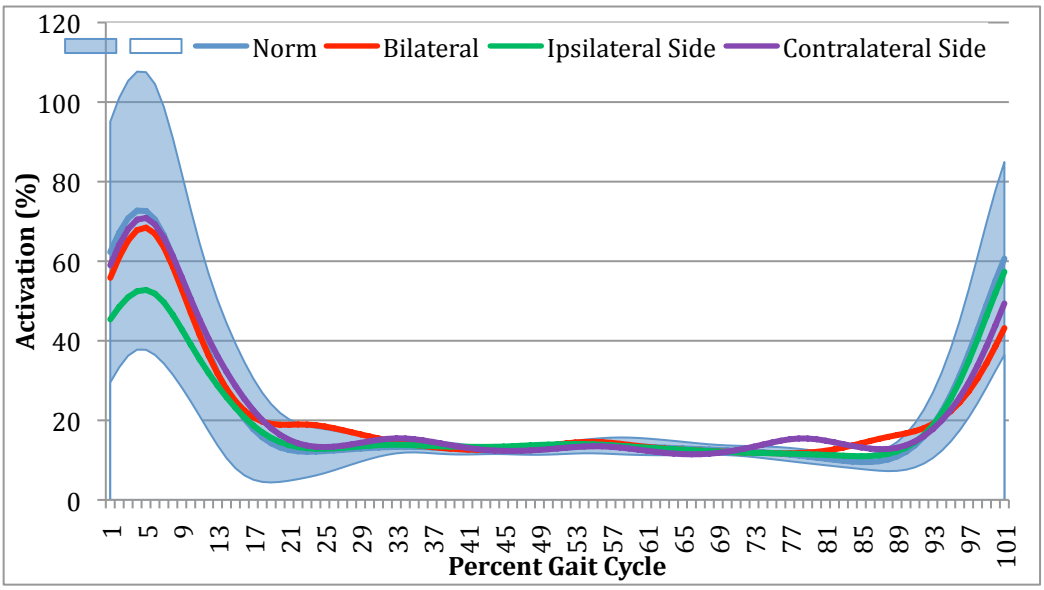

Figure 25: Mean Gluteus Medius Time series curves for the walking on level ground condition. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

The Rectus Abdominus muscles (Figure 26) showed no variation between conditions, but just maintained a steady level of activation. Each participant showed a different level of activation for this muscle with a different amount of normal variation possible but the overall strategy was the same in that a steady level of activation was maintained during the condition. When deviations did occur outside of the normal variation that is plotted, its duration and amplitude tended to be brief and small respectively.

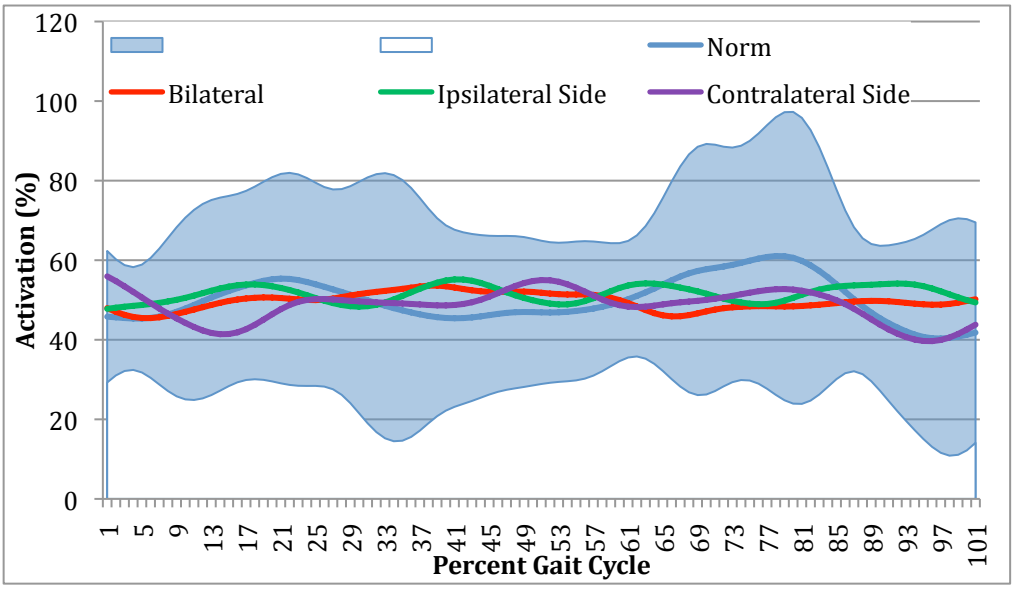

Figure 26: Mean Rectus Abdominus Time series curves for the walking on level ground condition. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

The erector spinae muscle exhibited a similar activation pattern across the different conditions (Figure 27). Only in the bilateral AMM condition did the average stray outside of the normal variation shown. This occurred during early stance phase in which an increase in activation was noted during the bilateral AMM condition. At toe off, a decrease across all conditions was observed, however, this deviation was well inside the normal variance.

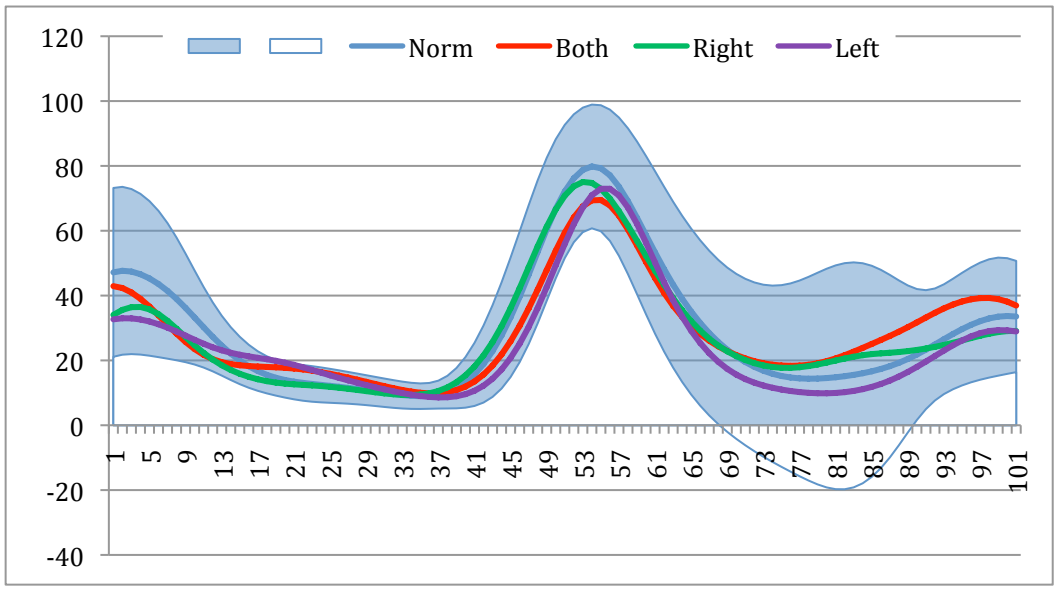

Figure 27: Mean Erector Spinae Time series curves for the walking on level ground condition. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

7.4 Obstacle

While the level walking conditions caused a restriction on the ankle joint, which caused the knee and the hip joints to alter the magnitudes of some of their power bursts, obstacle walking places even larger constraints on the lower body, especially the knee. While the power generation from the knee is more taxing, due to the extra K5 burst, this will hinder the amount of plasticity that the lower limb as a whole will have. This restriction of plasticity from the knee will place a larger demand on the hip joint in order to compensate for the loss of propulsion about the ankle.

7.4.1 Ankle:

Figure 28 below displays the ankle power bursts and the variations that exist between the different conditions that were place on participants. While a decrease in both peak power and the work done is seen, the timing of the burst is noted to be earlier than in level walking, in accordance with McFayden and Carnaghan. 1997.

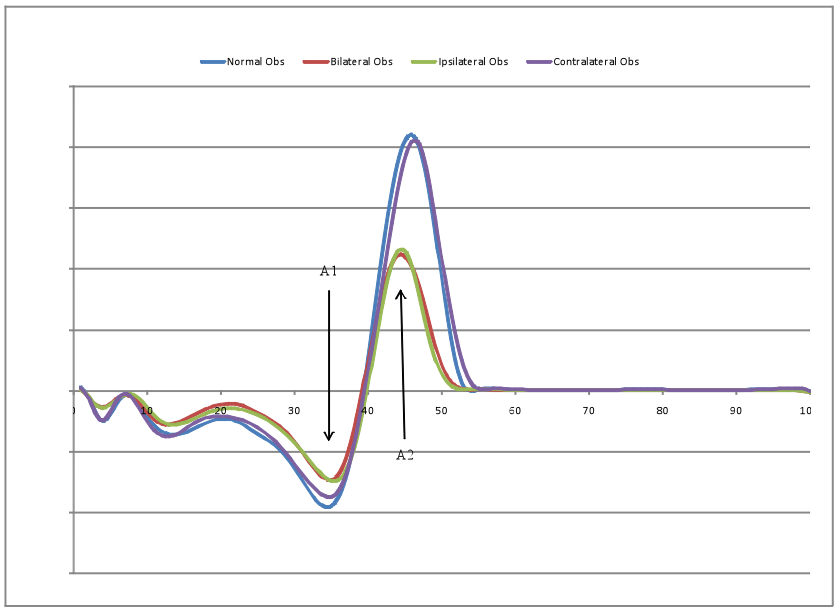

Figure 28: Mean Ankle Power Time series curves for the obstacle conditions. A generation of power is represented by positive values and power absorptions are represented by negative values.

The peak power during the A1 power phase at the ankle showed a significant decrease during the AMM affected conditions in comparison to the normal obstacle condition ($F(3, 42) = 13.391$, p-value <0.0001). The normal obstacle condition had an average value of -1.12 ± 0.28 watts/kg, while the bilateral AMM condition recorded -0.85 ± 0.22 watts/kg, the ipsilateral side of the unilateral condition was -1.16 ± 0.37 watts/kg and the contralateral side was-0.84 ± 0.19 watts/kg.

The average amount of work performed during the A1 power burst was calculated and determined to be significantly different while wearing vs. not wearing an AMM ($F(3, 42) = 29.6093$, p-value <0.0001). While not wearing any AMM, the amount of work done was -0.22 ± 0.05 J/kg, while wearing the AMM bilaterally the amount of work done was -0.16 ± 0.031 J/kg, the contralateral side of the unilateral walking was -0.28 ± 0.054 J/kg and the ipsilateral side was -0.16 ± 0.036 J/kg (Figure 29).

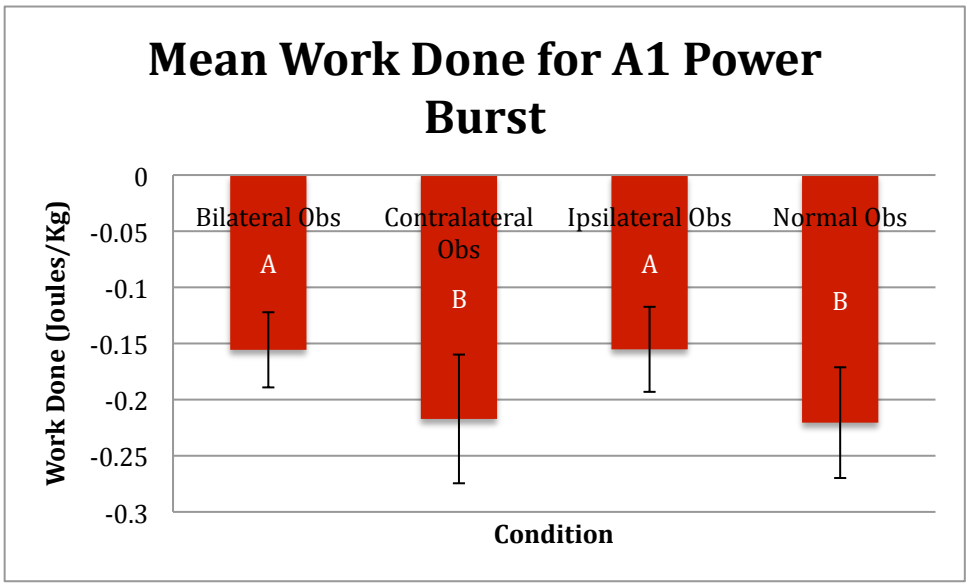

Figure 29: Mean work done during the A1 power burst across the 4 obstacle conditions. Levels not separated by the same letter are significantly different from each other.

The generation of peak power during the A2 power phase at the ankle showed a decrease during the AMM affected conditions in comparison to the normal obstacle condition ($F(3, 42) = 59.019$, p-value <0.0001). The normal walking condition had an average value of 3.65 ± 0.77 watts/kg, while the bilateral AMM condition recorded 1.96 ± 0.69 watts/kg, the contralateral side of the unilateral condition was 3.57 ± 0.80 watts/kg and the ipsilateral side was 1.72 ± 0.33 watts/kg.

The average amount of work performed during the A2 power burst was calculated and determined to be significantly different while wearing vs. not wearing an AMM ($F(3, 42) = 49.68$, p-value <0.0001). While not wearing the AMM the amount of work done was 0.25 ± 0.068 J/kg, while wearing the AMM bilaterally the amount of work done was 0.13 ± 0.048 J/kg, the contralateral side of the unilateral walking was 0.25 ± 0.068 J/kg and the ipsilateral side was 0.11 ± 0.024 J/kg (Figure 30).

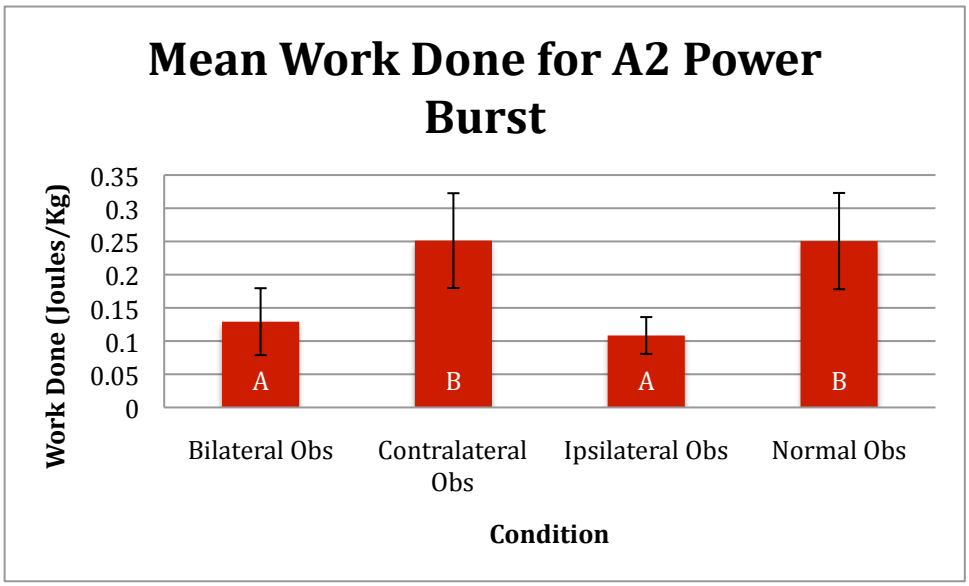

Figure 30: Mean Work Done during the A2 power burst across the 4 obstacle conditions. Levels not separated by the same letter are significantly different from each other.

The average peak ankle moment did not significantly change across conditions (Figure 31). The peak ankle moments were: 1.61 ± 0.19 N*m/kg, 1.53 ± 0.18 N*m/kg, 1.52 ± 0.16 N*m/kg, and 1.55 ± 0.23 N*m/kg respectively for the Normal, bilateral condition, ipsilateral side and the contralateral side of the unilateral condition.

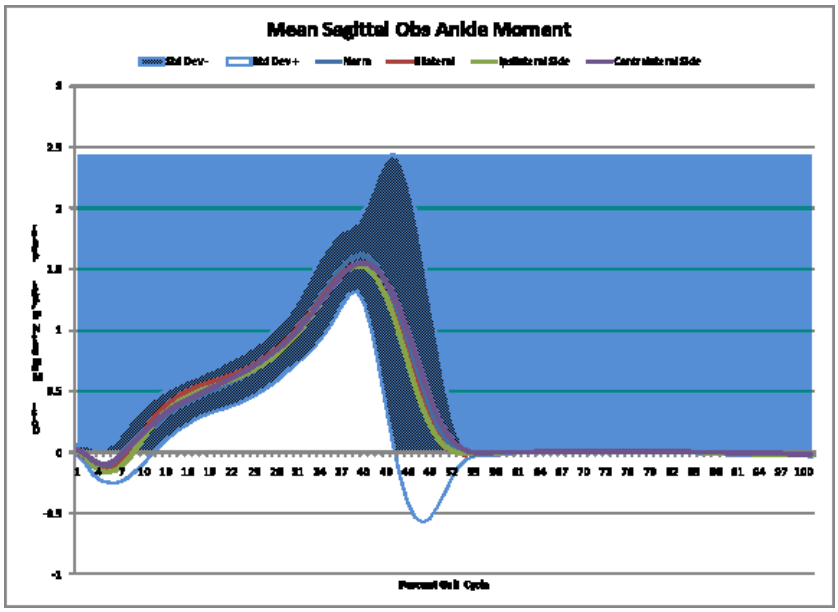

Figure 31: Mean Ankle Moment Time series curves for the obstacle walking conditions. The shaded blue area represents two standard deviations of the normal (no AMM) condition. Where a plantarflexion moment is represented by a positive number.

The range of motion that the ankle joint was able to go through was limited for the conditions including an AMM (Figure 32). The normal condition had a minimum peak ankle angle of -30.3 ± 7.3^0. The angle of the contralateral ankle during unilateral AMM walking was -29.3 ± 7.8^0, the ipsilateral side during unilateral walking recorded an angle of -18.4 ± 6.0^0 and the bilateral AMM condition was -19.2 ± 3.7^0. These angles were determined to be roughly around 50% of the gait cycle.

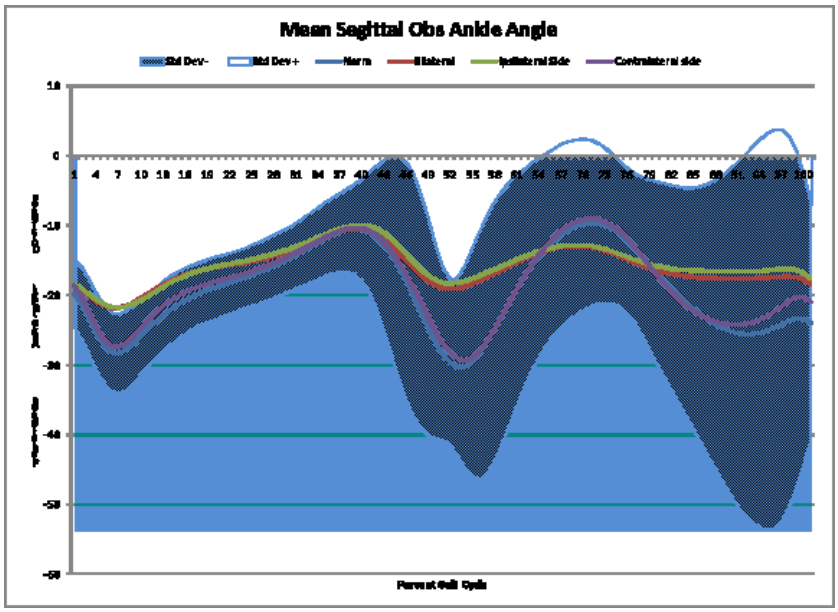

Figure 32: Ankle Angle Time series curves for the obstacle walking conditions. The shaded blue area represents two standard deviations of the normal (no AMM) condition. The black line represents the ankle angle during the quiet standing position.

7.4.2 Knee:

While the knee joint shows a similar pattern to the level walking condition, the extra K5 power burst is noted. This K5 power is utlizied in aiding knee elevation in order to achieve successful clearance of the obstacle during avoidance tasks.Again the K3 power burst is displayed as it is displays the transition from stance to swing.

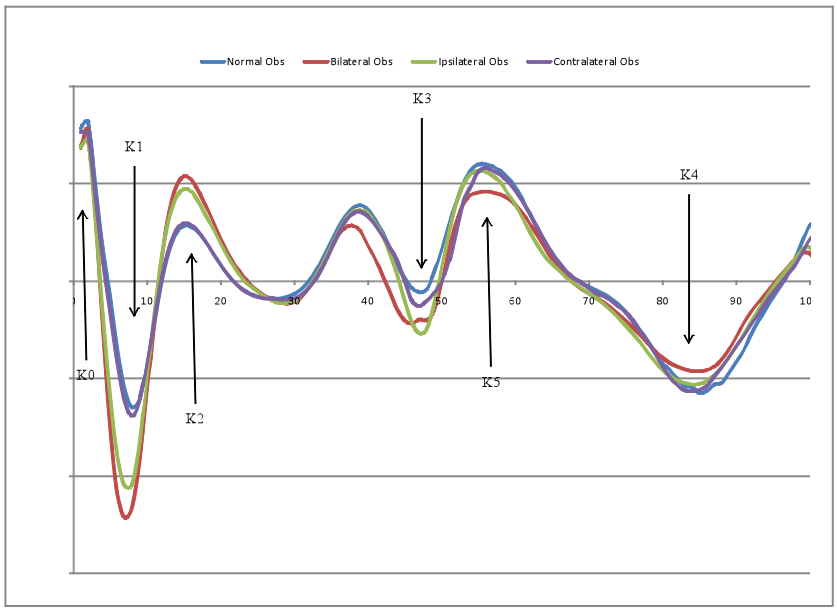

Figure 33: Mean Knee Power Time series curves for the obstacle walking conditions.

Table 9: Mean work done for knee power bursts across obstacle conditions. Levels not separated by the same letter are significantly different.

			Means (Joules/kg)	STD Dev (Joules/kg)
K0-S Work	Bilateral Obs	A	0.02	0.01
	Ipsilateral Obs	A	0.03	0.02
	Contralateral Obs	B	0.02	0.01
	Normal Obs	B	0.03	0.01
K1-S Work	Bilateral Obs	A	-0.08	0.08
	Ipsilateral Obs	A.B	-0.05	0.07
	Contralateral Obs	B,C	-0.07	0.06
	Normal Obs	C	-0.04	0.05
K2-S Work	Bilateral Obs	A	0.05	0.05
	Ipsilateral Obs	A,B	0.03	0.05
	Contralateral Obs	B	0.05	0.04
	Normal Obs	B	0.03	0.04
K3-S Work	Bilateral Obs	A	-0.05	0.04
	Ipsilateral Obs	A,B	-0.04	0.04
	Contralateral Obs	A,B	-0.05	0.03
	Normal Obs	B	-0.03	0.02
K4-S Work	Bilateral Obs	A	-0.11	0.05
	Ipsilateral Obs	A	-0.12	0.05
	Contralateral Obs	A	-0.12	0.05
	Normal Obs	A	-0.12	0.05
K5-S Work	Bilateral Obs	A	0.09	0.02
	Ipsilateral Obs	A	0.09	0.03
	Contralateral Obs	A	0.10	0.03
	Normal Obs	A	0.10	0.04

Analysis of the conditions throughout the gait cycle (Table 9) showed that the work done by each power burst was significantly different between the four different conditions for the K0, K1, K2 and K3 power bursts. These power bursts are represented below with the levels of significance. The average amount of work performed during the knee power bursts were calculated and determined to be significantly different between numerous conditions K0 ($F(3, 42) = 9.042$, p-

value <0.0001), K1 (F (3, 42) = 7.3198, p-value =0.0005), K2 (F (3, 42) = 5.4536, p-value =0.0029), K3 (F (3, 42) = 2.9773, p-value =0.00422) (Figure 34).

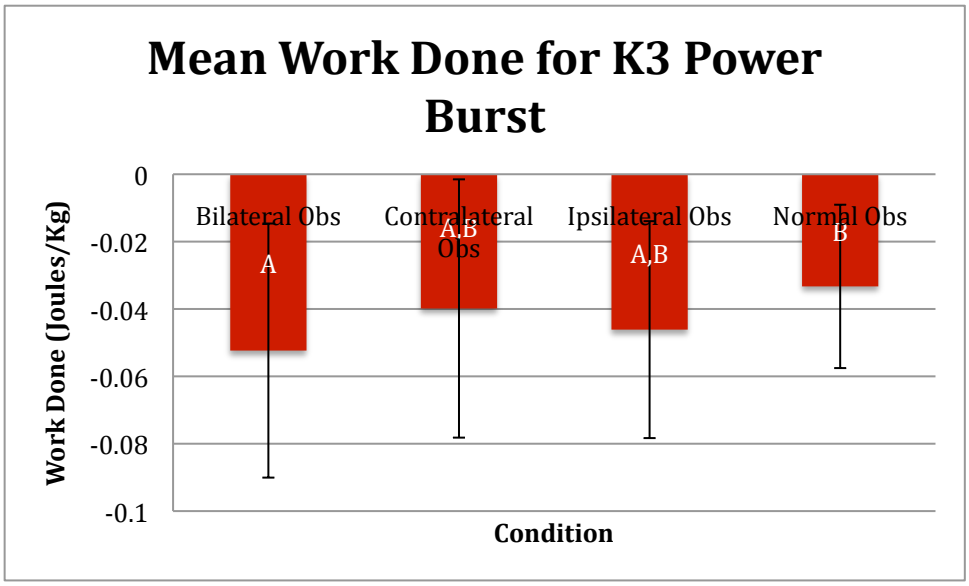

Figure 34: Average work done for the K3 Power burst across the 4 obstacle conditions. Levels not separated by the same letter are significantly different from each other.

Table 10: Mean Peak Powers for knee power bursts across conditions. Levels not separated by the same letter are significantly different. The K0 power was not included as this power burst is already decreasing as the gait cycle begins. Therefore the peak would be the beginning value of the gait cycle and not a true representation of the actual power burst.

			Means (Watts/kg)	STD Dev (Watts/kg)
K1-S Peak	Bilateral Obs	A	-1.30	1.25
	Ipsilateral Obs	A,B	-0.84	1.09
	Contralateral Obs	B	-1.10	0.83
	Normal Obs	B	-0.77	0.93
K2-S Peak	Bilateral Obs	A	0.57	0.49
	Ipsilateral Obs	A,B	0.33	0.51
	Contralateral Obs	B	0.51	0.38
	Normal Obs	B	0.33	0.51
K3-S Peak	Bilateral Obs	A	-1.04	0.54
	Ipsilateral Obs	A,B	-0.81	0.55
	Contralateral Obs	A,B	-0.97	0.46
	Normal Obs	B	-0.77	0.40
K4-S Peak	Bilateral Obs	A	-0.63	0.34
	Ipsilateral Obs	A,B	-0.75	0.39
	Contralateral Obs	A,B	-0.72	0.44
	Normal Obs	B	-0.80	0.44
K5-S Peak	Bilateral Obs	A	0.65	0.21
	Ipsilateral Obs	A,B	0.77	0.24
	Contralateral Obs	A,B	0.73	0.29
	Normal Obs	B	0.82	0.26

Analysis of the conditions throughout the gait cycle showed that the peak power by each power burst (Table 10) was significantly different between the four different conditions for all the knee power bursts. These power bursts are represented below with the levels of significance. The average peak power performed during the knee power bursts were calculated and determined to be significantly different between numerous conditions K1 ($F(3, 42) = 6.0542$, p-value= 0.0016), K2 ($F(3, 42) = 4.4832$, p-value =0.0081), K3 ($F(3, 42) = 3.6578$, p-value =0.0198), K4 ($F(3, 42) = 3.5757$, p-value =0.0216) and K5 ($F(3, 42) = 3.2884$, p-value =0.0298).

The average knee moment did not significantly change across conditions (Figure 35), even though there was an increase in amplitude during the AMM trials. The peak knee moments were: 0.41 ± 0.33 N*m/kg, 0.70 ± 0.39 N*m/kg, 0.67 ± 0.32 N*m/kg and 0.41 ± 0.38 N*m/kg respectively for the Normal, bilateral condition, ipsilateral side and the contralateral side of the unilateral condition.

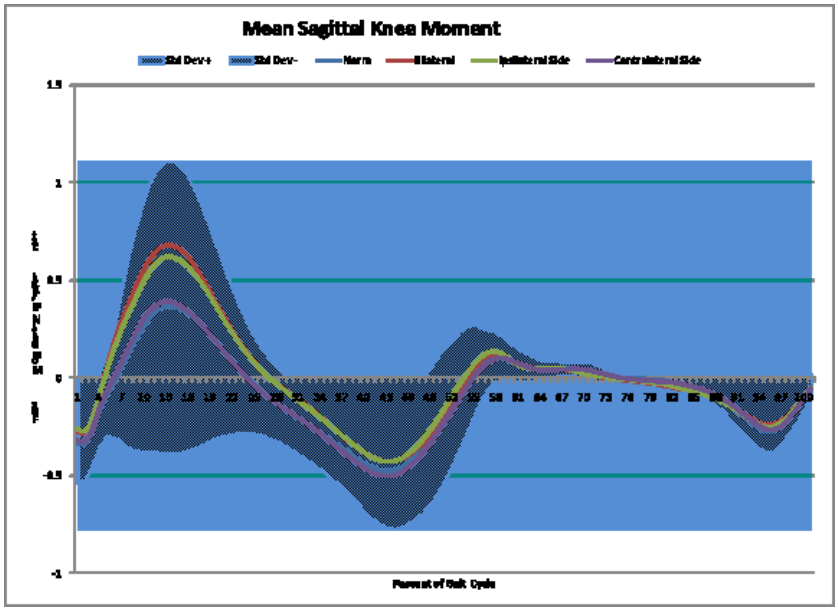

Figure 35: Mean Knee Moment Time series curves for the obstacle walking condition. The shaded blue area represents two standard deviations of the normal (no AMM) condition. Knee extension moments are represented by a positive number.

The range of motion that was completed by the knee joint was very similar regardless of condition imposed upon the participant (Figure 36). The normal condition had an average peak knee angle of 104.9 ± 7.1^0. The angle of the contralateral knee during unilateral AMM walking was 102.0 ± 8.5^0, the ipsilateral side during unilateral walking recorded an angle of 108.2 ± 8.2^0 and the bilateral AMM condition was 105.9 ± 6.4^0.

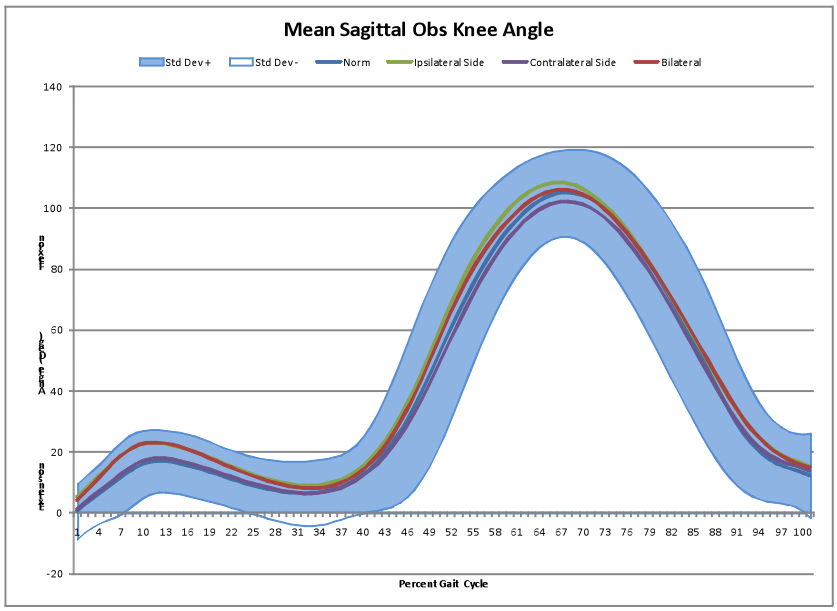

Figure 36: Knee Angle Time series curves for the obstacle walking conditions. The shaded blue area represents two standard deviations of the normal (no AMM) condition.

7.4.3 Hip:

Only the amount of work done at the generating power bursts (H1 and H3) showed a significant difference across the conditions. The H3 power burst is shown below as this power bursts is expressed during the stance to swing transition and is the most efficient way for generating forward propulsion during the AMM conditions.

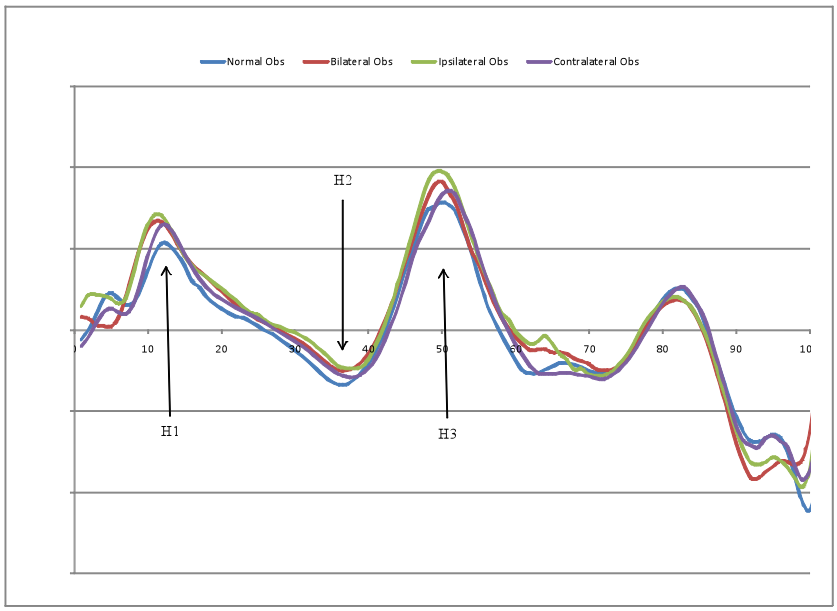

Figure 37: Mean Hip Power Time series curves for the obstacle walking conditions.

Table 11: Mean work done for Hip power bursts across obstacle conditions. Levels not separated by the same letter are significantly different.

			Means (Joules/kg)	STD Dev (Joules/kg)
H1-S Work	Bilateral Obs	A	0.12	0.05
	Ipsilateral Obs	A,B	0.10	0.05
	Contralateral Obs	A,B	0.13	0.06
	Normal Obs	B	0.09	0.06
H2-S Work	Bilateral Obs	A	-0.05	0.03
	Ipsilateral Obs	A	-0.05	0.05
	Contralateral Obs	A	-0.04	0.03
	Normal Obs	A	-0.06	0.05
H3-S Work	Bilateral Obs	A	0.14	0.05
	Ipsilateral Obs	A,B	0.13	0.04
	Contralateral Obs	A,B	0.15	0.07
	Normal Obs	B	0.12	0.04

Analysis of the conditions throughout the gait cycle showed that the work done by the H1 and H3 power bursts were significantly different between the four different conditions (Table 11). These power bursts are represented below with the levels of significance. The average amount of work performed during the hip power bursts were calculated and determined to be significantly different between numerous conditions H1 ($F(3, 42) = 3.8442$, p-value= 0.0161) and H3 ($F(3, 42) = 3.2866$, p-value =0.0298).

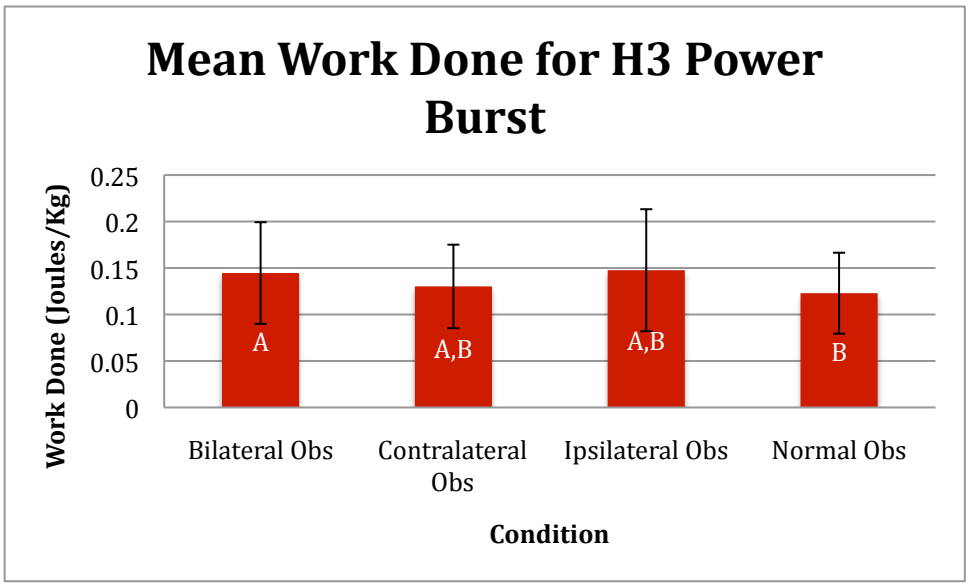

Figure 38: Average work done for the H3 Power burst across the 4 obstacle conditions. Levels not separated by the same letter are significantly different from each other.

Table 12: Mean Peak Powers for hip power bursts across obstacle conditions. Levels not separated by the same letter are significantly different.

			Means (Watts/kg)	STD Dev (Watts/kg)
H1-S Peak	Bilateral Obs	A	0.85	0.37
	Ipsilateral Obs	A	0.76	0.30
	Contralateral Obs	A,B	0.88	0.35
	Normal Obs	B	0.69	0.30
H2-S Peak	Bilateral Obs	A	-0.42	0.30
	Ipsilateral Obs	A	-0.44	0.34
	Contralateral Obs	A	-0.36	0.22
	Normal Obs	A	-0.48	0.30
H3-S Peak	Bilateral Obs	A	1.21	0.50
	Ipsilateral Obs	A	1.18	0.43
	Contralateral Obs	A	1.26	0.56
	Normal Obs	A	1.15	0.40

Analysis of the conditions throughout the gait cycle showed that the peak power during the H1 power burst were significantly different between the four different conditions. This power burst is represented below with the levels of significance. The average peak power produced during the hip power burst were calculated and determined to be significantly different between numerous conditions H1 ($F(3, 42) = 4.636$, p-value= 0.0069) (Table 12).

The Mean hip moment did not significantly change across conditions (Figure 39). The maximum hip moments were: 0.7 ± 0.17 N*m/kg, 0.72 ± 0.25 N*m/kg, 0.70 ± 0.22 N*m/kg and 0.70 ± 0.17 N*m/kg respectively for the Normal, bilateral condition, ipsilateral side and the contralateral side of the unilateral condition. While the minimum hip moments were: -0.89 ± 0.19 N*m/kg, - 0.89 ± 0.19 N*m/kg, -0.92 ± 0.21 N*m/kg, and -0.91 ± 0.23 N*m/kg respectively for order mentioned above. The extensor moment that occurs during the middle of the gait cycle or just prior to toe off before the obstacle was also recorded. These peaks were: -0.49 ± 0.21 N*m/kg, - 0.49 ± 0.15 N*m/kg, -0.45 ± 0.17 N*m/kg, and -0.49 ± 0.24 N*m/kg respectively.

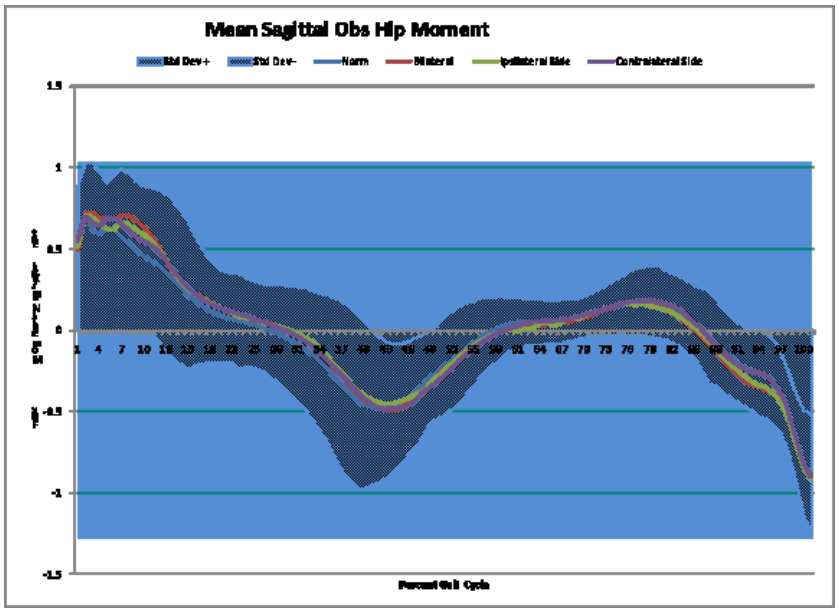

Figure 39: Mean Hip moment Time series curves for the obstacle walking conditions. The light shaded blue area represents one standard deviation of the normal (no AMM) condition. Where a positive number indicates an extensor moment.

The range of motion of the hip was recorded as: normal condition 89.7 ± 10.1^0, the contralateral hip during unilateral AMM walking was 86.6 ± 16.0^0, the ipsilateral hip during unilateral walking recorded an angle of 88.0 ± 19.2^0 and the bilateral AMM condition was 88.0 ±18.1^0. These angles and ROM were all very similar across conditions (Figure 40).

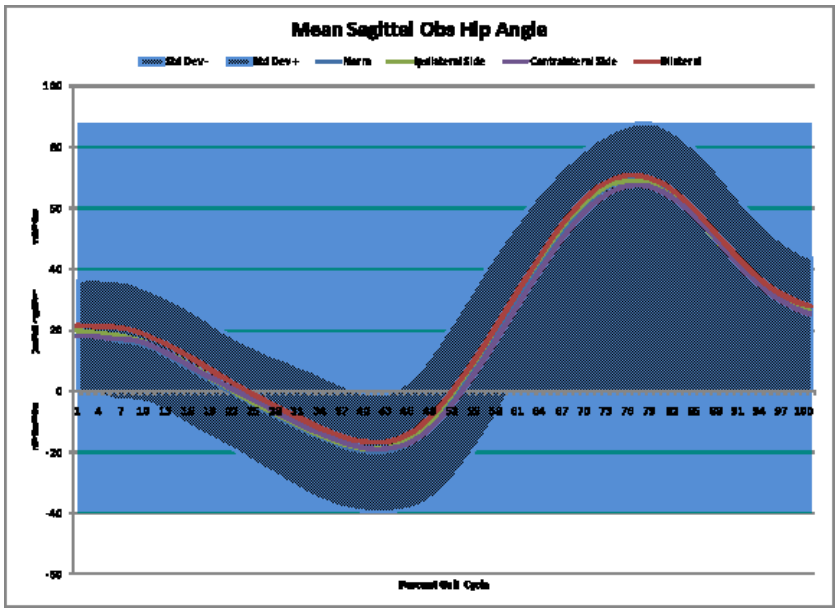

Figure 40: Mean Hip angle Time series curves for the obstacle walking conditions. The light shaded blue area represents one standard deviation of the normal (no AMM) condition. Where a positive number indicates hip flexion.

7.5 Individual Measures:

Throughout the obstacle conditions, there was a number of methods participants used in order to deal with the obstacle as well as the AMM. Of the fifteen subjeccts, six were able to maintain their walking velocity while walking over the obstacle in the bilateral conditions when compared with the normal obstacle condition. Of these six subjects, five of them maintained their walking velocity by decreasing cadence and increasing the stride length. The last subject decreased cadence but did not noticibly increase stride length.

Eight of the subjects displayed a decrease in walking velocity of the bilateral condition compared to the normal obstacle condition. Of these eight subjects, four of them decreased cadence and increased stride length while the other four increased cadence and decreased their stride length.

One subject subject actually increased walking velocity during the bilateral obstacle condition, this was accomplished by increasing cadence by a larger amount than they decreased their stride length.

Table 13: Mean temporal measures per participant collapsed over the individual obstacle walking trials.

		Walking Velocity (m/s)	S.D.	Cadence (steps/min)	S.D.	Stride Length (m)	S.D.	Stance Ratio	S.D.	Swing Ratio	S.D.
Female #1	Normal Obs	1.18	0.08	99.54	2.45	1.36	0.02	0.62	0.01	0.38	0.01
	Bilateral Obs	1.08	0.07	73.39	1.65	1.62	0.03	0.46	0.02	0.54	0.02
	Ipsilateral Side	1.06	0.05	89.97	1.91	1.72	0.02	0.53	0.01	0.47	0.01
	Contralateral Side	1.02	0.04	98.45	3.55	1.50	0.04	0.62	0.01	0.38	0.01
Female #2	Normal	1.06	0.05	82.27	1.91	1.54	0.04	0.51	0.01	0.49	0.01
	Bilateral	1.13	0.05	100.56	1.96	1.32	0.03	0.60	0.01	0.40	0.01
	Ipsilateral Side	1.15	0.05	102.42	1.45	1.47	0.02	0.60	0.01	0.40	0.01
	Contralateral Side	1.01	0.03	72.95	3.60	1.88	0.06	0.42	0.02	0.58	0.02
Female #3	Normal	1.08	0.06	103.80	1.49	1.42	0.02	0.63	0.01	0.37	0.01
	Bilateral	1.07	0.04	82.66	2.62	1.63	0.02	0.49	0.02	0.51	0.02
	Ipsilateral Side	1.07	0.04	85.04	2.82	1.61	0.04	0.50	0.01	0.50	0.01
	Contralateral Side	1.06	0.07	99.45	3.25	1.55	0.05	0.59	0.01	0.41	0.01
Female #5	Normal	1.27	0.04	79.17	2.67	1.56	0.03	0.48	0.01	0.52	0.01
	Bilateral	1.28	0.03	105.39	2.18	1.36	0.02	0.62	0.01	0.38	0.01
	Ipsilateral Side	1.35	0.02	100.53	0.68	1.47	0.03	0.60	0.01	0.40	0.01
	Contralateral Side	1.29	0.03	80.31	2.31	1.82	0.03	0.45	0.02	0.55	0.02
Female #6	Normal	1.27	0.06	99.42	2.72	1.32	0.03	0.61	0.01	0.39	0.01
	Bilateral	1.11	0.05	83.50	1.48	1.71	0.03	0.50	0.01	0.50	0.01
	Ipsilateral Side	1.23	0.08	89.74	1.23	1.71	0.02	0.51	0.01	0.49	0.01
	Contralateral Side	1.19	0.04	104.84	2.79	1.53	0.04	0.60	0.01	0.40	0.01
Female #7	Normal	1.49	0.07	84.82	2.11	1.59	0.06	0.50	0.01	0.50	0.01
	Bilateral	1.43	0.04	98.23	1.54	1.59	0.04	0.59	0.01	0.41	0.01
	Ipsilateral Side	1.42	0.04	105.54	2.22	1.48	0.03	0.60	0.01	0.40	0.01
	Contralateral Side	1.43	0.04	85.07	4.46	1.62	0.03	0.48	0.01	0.52	0.01
Female #8	Normal	1.28	0.06	103.94	1.38	1.37	0.02	0.62	0.01	0.38	0.01
	Bilateral	1.08	0.07	84.81	1.91	1.71	0.04	0.51	0.01	0.49	0.01
	Ipsilateral Side	1.11	0.04	86.86	2.79	1.67	0.03	0.44	0.02	0.56	0.02
	Contralateral Side	1.18	0.06	106.72	4.73	1.59	0.03	0.59	0.01	0.41	0.01
Male #1	Normal	1.12	0.03	88.87	1.56	1.70	0.02	0.51	0.01	0.49	0.01
	Bilateral	0.90	0.04	99.76	1.52	1.61	0.02	0.61	0.01	0.39	0.01
	Ipsilateral Side	0.98	0.03	111.35	2.41	1.73	0.03	0.59	0.01	0.41	0.01
	Contralateral Side	1.11	0.03	85.57	6.07	1.65	0.02	0.47	0.03	0.53	0.03
Male #2	Normal	1.18	0.03	102.60	5.81	1.49	0.04	0.60	0.04	0.40	0.04
	Bilateral	1.20	0.02	85.39	2.22	1.76	0.02	0.49	0.01	0.51	0.01
	Ipsilateral Side	1.24	0.04	88.62	4.91	1.74	0.03	0.45	0.01	0.55	0.01
	Contralateral Side	1.21	0.03	108.55	3.01	1.58	0.03	0.60	0.01	0.40	0.01
Male #3	Normal	1.18	0.04	88.84	2.22	1.71	0.04	0.53	0.01	0.47	0.01
	Bilateral	1.03	0.04	100.28	1.52	1.64	0.03	0.58	0.01	0.42	0.01
	Ipsilateral Side	0.96	0.10	113.18	3.03	1.71	0.04	0.58	0.01	0.42	0.01
	Contralateral Side	1.09	0.05	87.18	3.23	1.63	0.03	0.48	0.01	0.52	0.01
Male #4	Normal	1.21	0.05	103.39	1.63	1.47	0.05	0.62	0.01	0.38	0.01
	Bilateral	1.14	0.07	81.82	1.41	1.75	0.03	0.51	0.01	0.49	0.01
	Ipsilateral Side	1.17	0.04	95.45	2.09	1.67	0.05	0.44	0.01	0.56	0.01
	Contralateral Side	1.17	0.09	110.89	2.79	1.59	0.02	0.59	0.01	0.41	0.01
Male #5	Normal	1.27	0.02	87.80	1.60	1.72	0.02	0.52	0.01	0.48	0.01
	Bilateral	1.08	0.04	96.02	1.59	1.55	0.02	0.61	0.01	0.39	0.01
	Ipsilateral Side	1.14	0.04	113.32	2.43	1.62	0.04	0.58	0.00	0.42	0.00
	Contralateral Side	1.28	0.04	89.36	2.43	1.65	0.03	0.48	0.01	0.52	0.01
Male #6	Normal	1.28	0.05	101.48	1.55	1.47	0.02	0.60	0.01	0.40	0.01
	Bilateral	1.22	0.05	71.91	2.21	1.80	0.04	0.43	0.02	0.57	0.02
	Ipsilateral Side	1.32	0.05	91.50	2.72	1.68	0.06	0.43	0.01	0.57	0.01
	Contralateral Side	1.30	0.06	107.33	2.80	1.59	0.04	0.60	0.01	0.40	0.01
Male #7	Normal	1.14	0.03	90.06	1.49	1.69	0.03	0.53	0.01	0.47	0.01
	Bilateral	0.99	0.06	101.09	1.91	1.54	0.04	0.59	0.01	0.41	0.01
	Ipsilateral Side	1.03	0.04	118.56	2.41	1.71	0.04	0.57	0.01	0.43	0.01
	Contralateral Side	1.05	0.03	82.68	3.06	1.58	0.05	0.50	0.01	0.50	0.01
Male #8	Normal	1.29	0.02	103.23	0.98	1.45	0.01	0.62	0.01	0.38	0.01
	Bilateral	1.27	0.07	76.77	3.20	1.75	0.05	0.46	0.01	0.54	0.01
	Ipsilateral Side	1.24	0.03	76.56	4.35	1.54	0.04	0.49	0.01	0.51	0.01
	Contralateral Side	1.25	0.03	98.13	2.29	1.36	0.04	0.60	0.01	0.40	0.01

7.6 EMG

Electromyography for the obstacle trials was also collected from the muscles previously mentioned. Again, there were no remarkable changes outside of the normal deviation that were consistent between the participants. Each participant did have their own adaptations between the conditions to manage with the constraints of the AMMs. An average of all of the participant's

data was noted to "wash out" any of the individual variability; therefore a grand mean was not calculated. One representative sample of the muscle patterns was shown here, while two more samples for each muscle have been included in the appendix.

For the Tibialis anterior muscle, there seemed to be two types of muscle activity patterns. The one shown below exhibited no deviation outside the normal variation whereas the second graph shows a marked decrease in activation during late stance through early swing phases. While most participants followed the activation patterns displayed in the first graph, two of the female participants displayed the patterns shown in Figure 42. These two individuals were also the only two to display a touching down of the toes first rather than the heel in obstacle clearance tasks, which was displayed by analysis of the kinetic data.

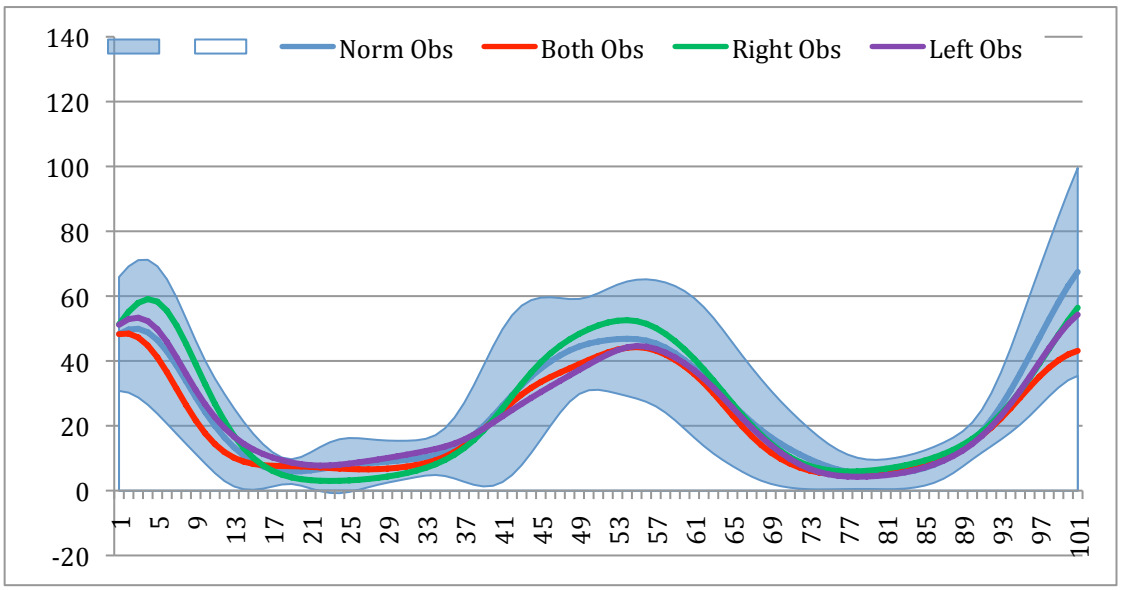

Figure 41: Mean Tibialis Anterior Time series curves for the obstacle walking conditions. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

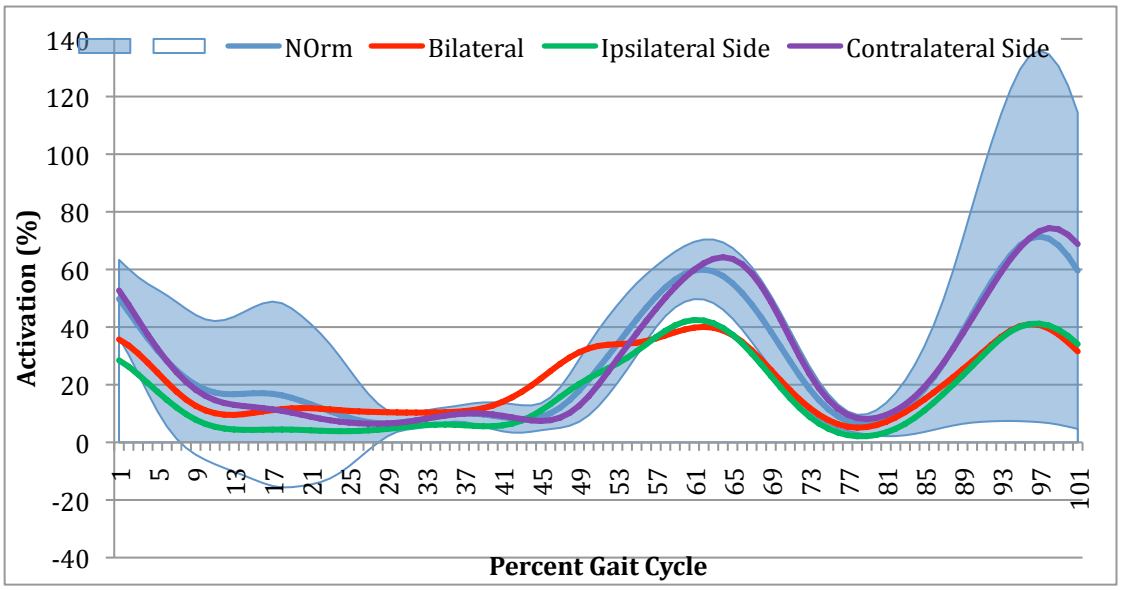

Figure 42: Mean Tibialis Anterior Time series curves for the obstacle walking conditions. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

During the AMM imposed conditions, the gastrocnemius muscle showed very little change in the level of activation through the gait cycle. A small decrease can be seen during the AMM imposed trials but the difference is not significant in that it lies well within the normal variance of the non-AMM trials.

A small deviation from the normal variance is noted during late stance and into early swing of the contralateral foot to the foot wearing the AMM. While the activation level lies outside the normal variance, it is still noticeably low and close to the other AMM conditions.

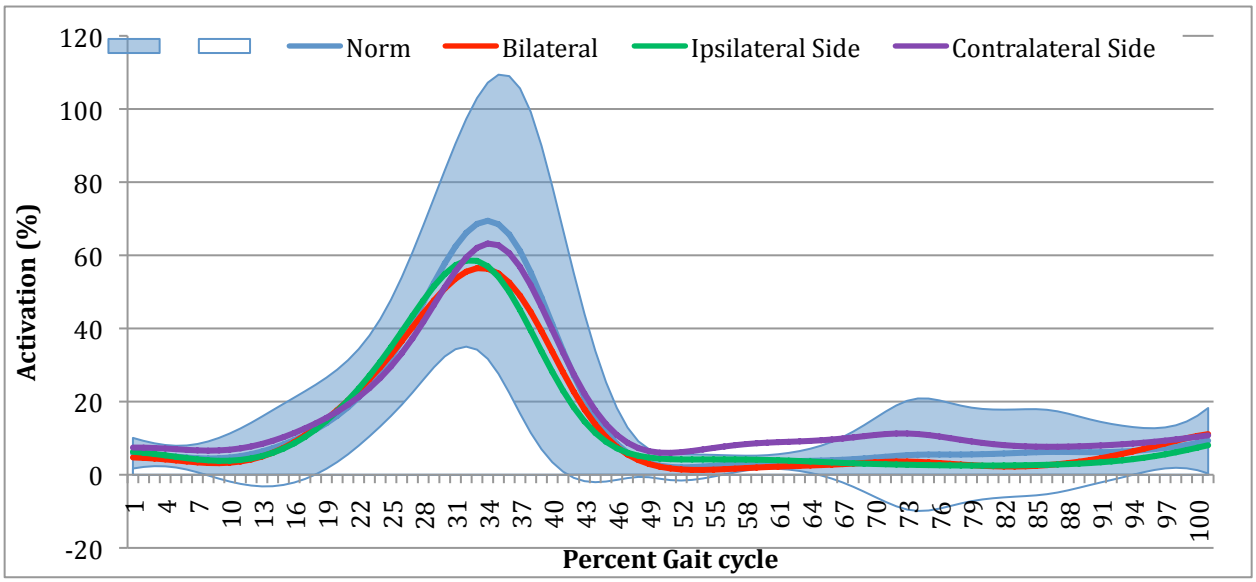

Figure 43: Mean Gastrocnemius Time series curves for the obstacle walking conditions. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

The Rectus Femoris muscle did not deviate from the normal obstacle condition until late in the swing phase. This corresponds with where the obstacle was located. The increase in the normal condition shows an increase in muscle activation to aid in acquiring adequate toe clearance over the obstacle. During the unilateral trials, the contralateral leg showed a very similar activation pattern and did not deviate outside of the normal range. The bilateral AMM condition and the ipsilateral condition displayed an increase in the overall activation level during late swing phase when the participant was clearing the obstacle. The bilateral condition showed a marginally higher activation level then the unilateral condition, however, both lay significantly outside the normal range of variance.

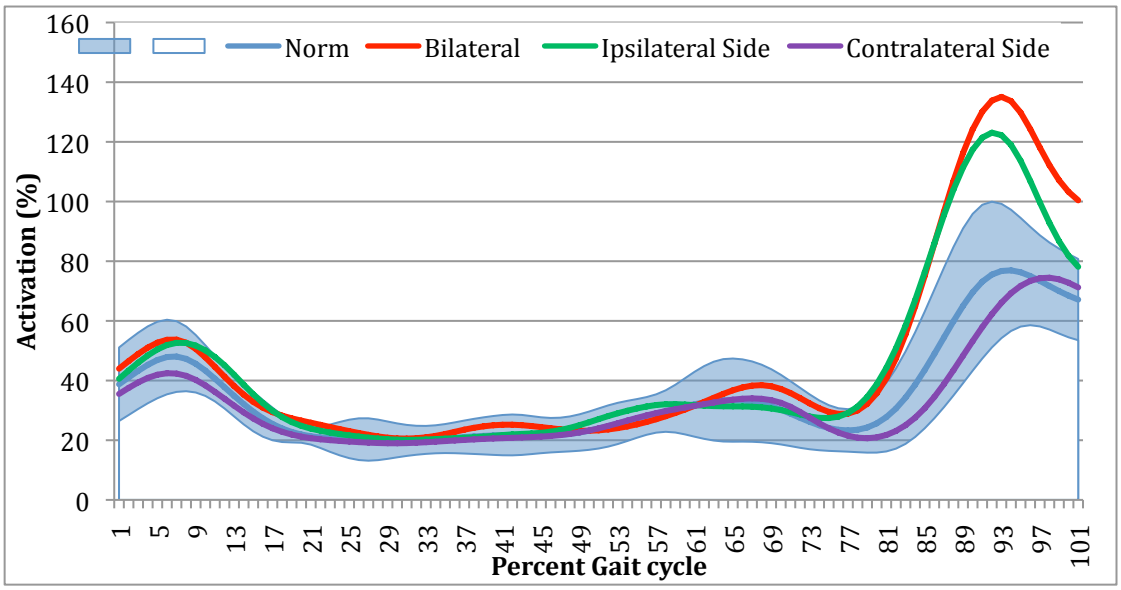

Figure 44: Mean Rectus Femoris Time series curves for the obstacle walking conditions. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

The Biceps femoris muscle showed an increase in activation across all conditions just before Toe off (approx 60%) of the gait cycle. An increase in the contralateral foot to the one wearing the AMM is noted but did not go outside the range of two standard deviations of the normal obstacle trials. This increase is occurred both at toe off and again right before heel strike of the gait cycle. A small decrease in the bilateral AMM and the ipsilateral side of the unilateral AMM conditions occurred during mid to late stance, but again were not significantly different.

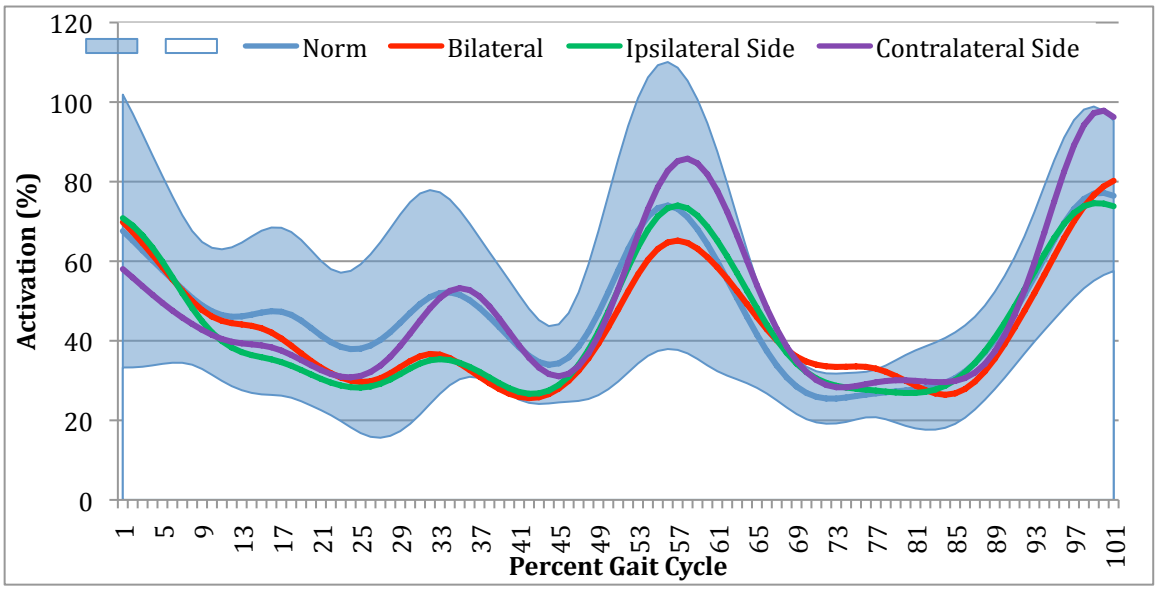

Figure 45: Mean Biceps Femoris Time series curves for the obstacle walking conditions. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

Gluteus Maximus being a hip extensor displays a similar pattern to walking on level ground, with an increase in activation both at the beginning and near the end of the gait cycle. However, an increase in muscle activation is witnessed during obstacle walking just prior to toe off in anticipation to clear the obstacle. All four conditions had a relatively similar level of activation, with the only deviation, for this particular subject, from the norm coming late in swing phase. Most subjects did not show a deviation from the normal condition during the obstacle trials.

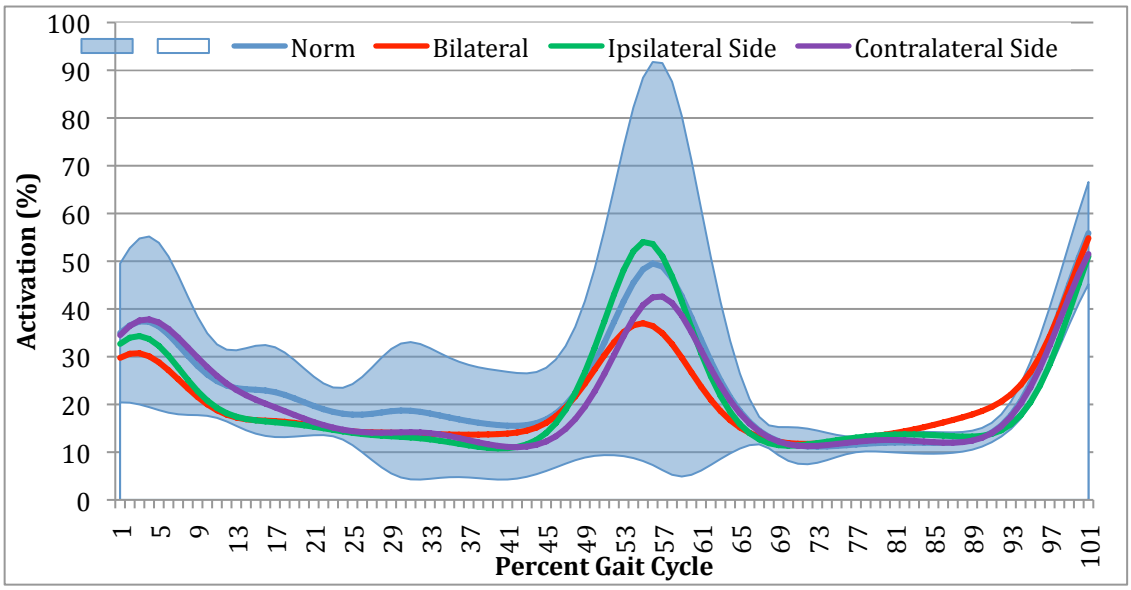

Figure 46: Mean Gluteus Maximus Time series curves for the obstacle walking conditions. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

From most subjects, no deviation from the normal variance was seen in the Gluteus medius muscle. The characteristic increase at the beginning and the end of the gait cycle remained constant from the level walking trials. Two subjects displayed slight deviations during bilateral AMM trials and/or the contralateral side to the AMM during the unilateral trials. The deviations beyond the normal variance was only slight and therefore not of major concern.

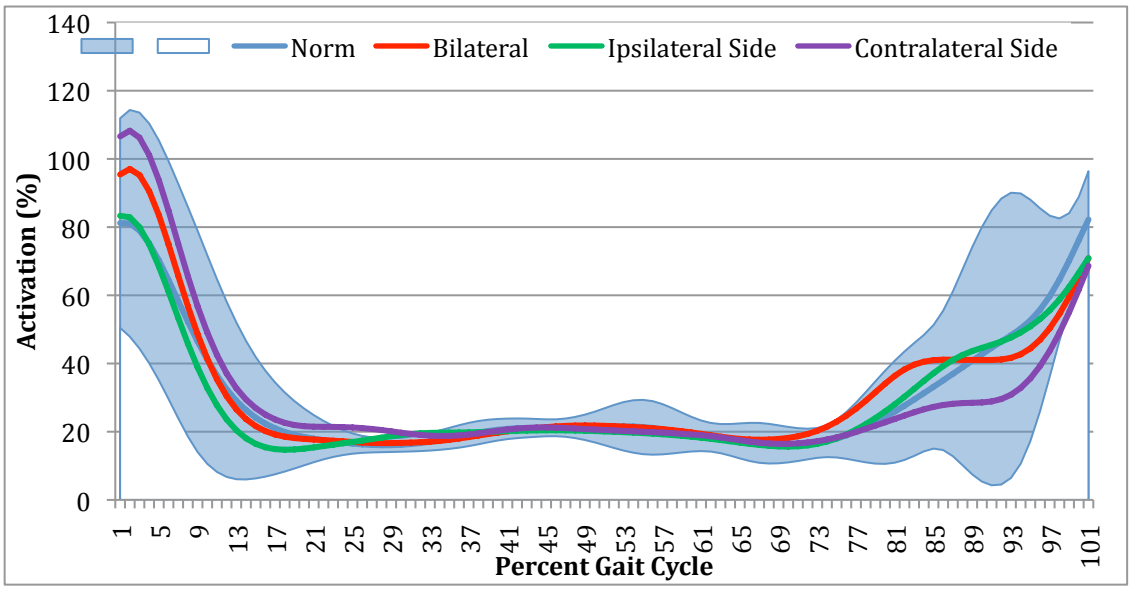

Figure 47: Mean Gluteus Medius Time series curves for the obstacle walking conditions. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

For the majority of participants that Rectus Abdominus muscle followed a similar pattern to that shown in Figure 51. A fairly steady activation pattern is displayed with a large increase in activation toward the end of the swing phase. The increase in activation across all conditions at approx 45% of the gait cycle is in correspondence with the activation pattern during the level ground walking trials. The large increase toward the end of the gait cycle is in accordance with the positioning of the obstacle that was to be avoided. An increase is present from the bilateral condition and less so from the ipsilateral side of the unilateral conditions. Neither of the two conditions elicited a response which would cause the participant to produce an activation level with was outside the normal variance. While this is the case for most participants, a few individuals presented an increase in activation, particularly in the bilateral condition, which did deviate beyond the normal variance of the non-AMM condition.

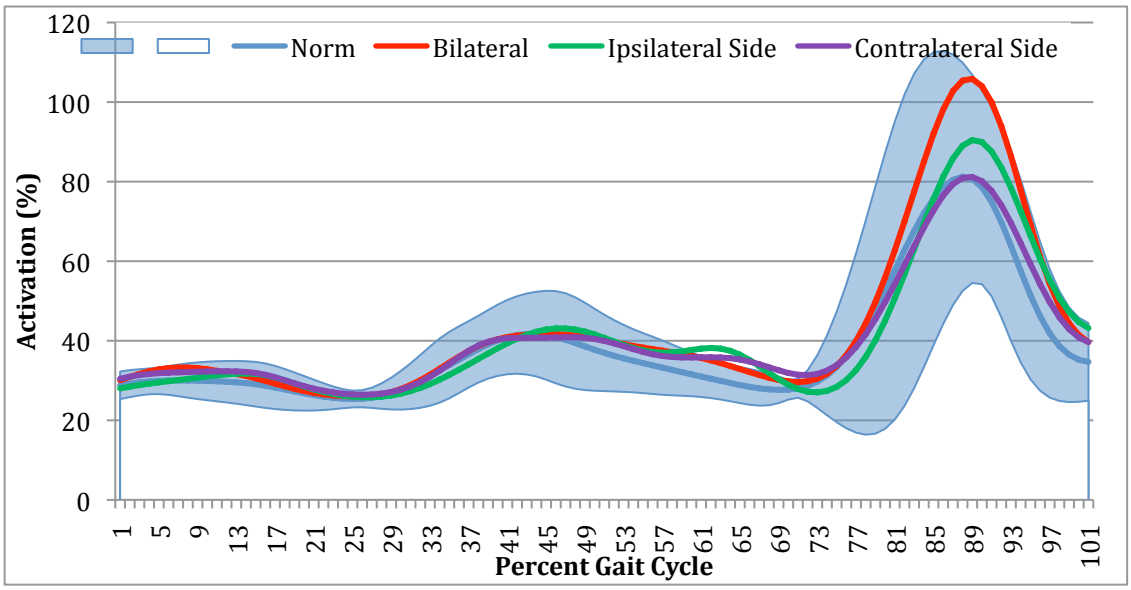

Figure 48: Mean Rectus Abdominus Time series curves for the obstacle walking conditions. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

The normal variance of the erector spinae muscle was quite small compared to most of the other muscles that were collected. For all of the participants except three, the three AMM conditions fell within the narrow band of the normal variance at all times. The increase in activation that was observed at approx 40% of the gait cycle is in accordance with the normal walking ground trials and its relevance were discussed later on. The increase at the very end of the cycle is just prior to heel strike after the obstacle.

Two of the participants that had some variance outside the normal occurred in the bilateral condition as well as the ipsilateral side of the unilateral condition. An increase in activation throughout the entire trial is seen in these participants, whereas the other participant displayed a delay in onset and return to baseline values of the muscle's activation through the middle of the stance phase.

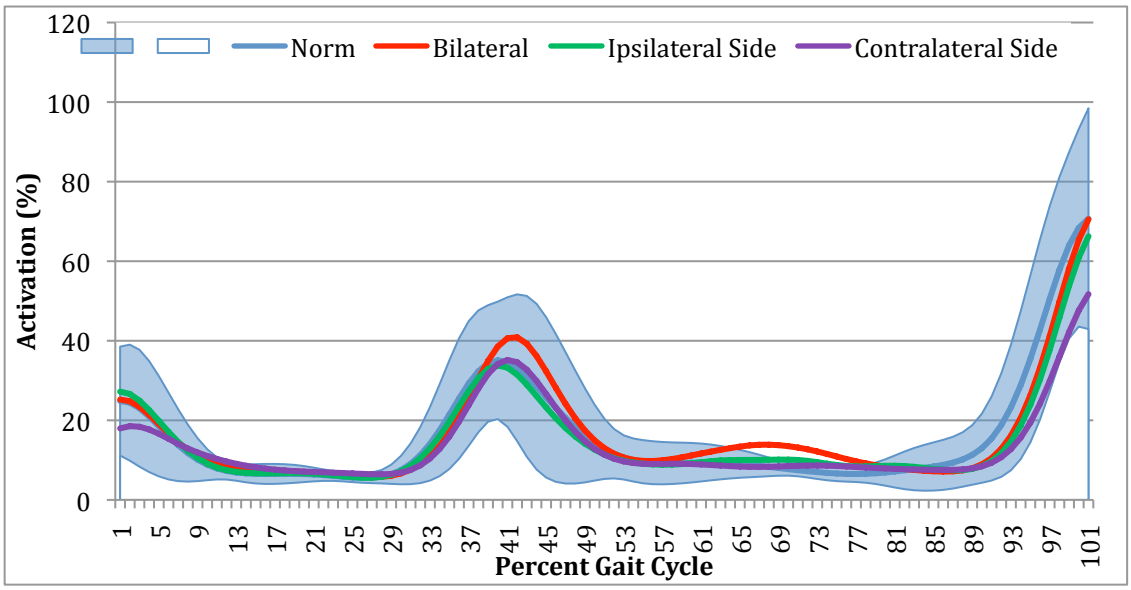

Figure 49: Mean Erector Spinae Time series curves for the obstacle walking conditions. The light shaded blue area represents one standard deviation of the normal (no AMM) condition.

8.0 Discussion

8.1 Addressing Specific aims and hypothesis

Hypothesis 1:

It was hypothesized that the ankle joint will contribute less to the overall forward body propulsion when the ankle is functioning in a limited capacity due to the restriction of the AMM. The results of the ankle joint contribution to propulsion (A1 and A2) showed a large reduction in the amount of work done, both positive and negative, by the ankle throughout the gait cycle. Both the A1 and A2 powers showed a reduction in amplitude and duration in the leg that the AMM was being worn on. This was also evident from the EMG of the gastrocnemius muscle as well, as it showed a decrease in activation during the trials with the AMM. During unilateral AMM walking, there was a slight increase in both peak power as well as the work done by the leg that was not being restricted by the AMM, leading to a very small compensation from the non-AMM ankle. Therefore hypothesis #1 is supported by the results of this investigation.

Hypothesis 2:

It was hypothesized that the hip and knee powers will increase to compensate for a loss of ankle power.

The results of the hip peak powers (H1, H2, and H3) showed an increase in the positive powers as well as a decrease in the negative peak power. The work done showed an increase in generation of power for H1 and H3, while a decrease in the absorption of power through the H2 power burst for the all trials which included the AMM versus walking without the AMM; however, there was only a significant difference in the work performed at the hip and not the

peak powers. The contralateral leg had increases at the hip, and while they were significant, they were of smaller magnitude to that of the ipsilateral leg.

The knee peak powers showed an increase in absorption for the K1 and the K3 power bursts, where as the K2 peak power had an increase in peak power generation and the K4 peak power burst showed a decrease in absorption. However, none of the peak powers displayed significance between the conditions. For the work done at the knee, K1 and K3 both showed a statistically significant increase in power generation during trials in which an AMM was present in comparison to the normal walking trials. However, the K0 burst displayed a significant decrease in the amount of power generated at the knee while the K2 work done had an increase in power generation that was also statistically significant. The K4 power burst did not display a differences between the various conditions. The K1, K2, and K3 power bursts will have an indirect influence on propulsion by allowing other joints to act differently to compensate for the lack of ankle power being generated.

Therefore hypothesis #2 is supported by the results of this investigation.

Hypothesis 3:

It was hypothesized that restraining the ankle joint during walking would lead to an increase in muscle activation of the plantarflexors and the knee extensors.

The results of the gastrocnemius muscle showed a decrease in activation while the AMM was placed on the ipsilateral foot as well as when the AMM was placed bilaterally compared to when the participant was not wearing the AMMs. When the AMM was placed unilaterally, there was an increase in activation shown by the gastrocnemius of the contralateral leg.

The results of the rectus femoris muscle (also a hip flexor) showed a large increase in activation during trials with the AMM versus not wearing an AMM. The increase in activation was more

pronounced on the side that the AMM was on; however, there was still an increase in the activation of the contralateral leg.

Therefore hypothesis #3 was only partially supported by the results of this investigation.

8.2 Walking Velocity

During the AMM trials, a decrease in walking velocity occurred. Since we are manipulating the ankle and causing the ankle power (especially A2) to drop, which it does dramatically, we would expect a decrease in gait speed, if this lack of power was not compensated for. There is indeed a decrease in the gait velocity; since it was not statistically significant it is therefore possible that the remaining joints were able to provide most of the lost propulsion. With a drop in gait velocity, Winter showed that there is a decrease in the hip and knee power bursts throughout the gait cycle. During the slower walking condition the K1, K2, K3 and K4 power bursts all decrease their peak power in relation to the peak powers that were obtained at a natural walking velocity, whereas during the unilateral AMM and bilateral AMM conditions, the peaks of all four knee powers increased, while the gait velocity was slower than that of the normal condition. The contralateral side of the unilateral AMM condition did have a decrease in gait velocity but only had a slight increase in the peak powers at the knee in comparison to the normal condition. The H1 and H3 power burst also normally decrease with a decrease in gait velocity while the H2 power burst has a shorter duration as well as a decrease in its peak amplitude (Winter 91). During the AMM trials, the H1 peak power burst actually increased power generation, while the gait velocity decreases. The H2 burst is slightly shorter in duration then the normal trial and consists of a smaller negative peak which is in accordance with Winter. The H3 burst again, shows an increase in peak power for both the ipsilateral side of the unilateral AMM and bilateral AMM conditions, while the contralateral side of the unilateral condition's peak power is below the normal walking value.

This general trend is in opposition of what Winter found for the normal slowing of gait velocity and therefore the increases in the peak powers cannot be explained by the change in gait velocity

as we would expect to find a decrease in these powers. The decrease in the A1 and A2 peaks are too large to be explained by only a decrease in gait velocity and these can be contributed to the restrictions placed on the ankle joint by the modified AMM. The K4 burst has previously been identified as an eccentric contraction of the knee flexors to slow down the leg just before heel contact, since there a negligible mass being added with the AMM, this contraction does not have to work any harder to slow down the leg as the inertia properties are very similar with and without the AMM. Since the participants are walking at a slower velocity, the swinging leg will not have as large of an inertial force and therefore the knee flexors do not need to activate as much to slow the limb to the desired speed. The H2 power burst has been described as absorption of energy by hip flexors as the thigh is decelerated as it rotates backwards. The power burst stabilizes the trunk and maintains the trunk in an upright posture. During the AMM trials, the gait velocity is decreased in comparison to the normal walking trials and therefore there is less energy required to be absorbed by the hip flexors during the H2 power phase. This is the reason that there is a decrease in the peak activation levels of these muscles at the slower speeds. This slower speed could be part of the reason for the decrease in the peak H2 power burst, although it does not solely explain it. If a decrease in gait velocity was to solely explain the decrease in the peak powers then we would see a much smaller decrease then what was recorded in this study.

8.3 Ankle Actions during Level Walking

The AMMs, while being worn on both feet, have the desired effect of decreasing the amount of ankle power during the stance phase of gait. This lack of plantarflexion that is seen through the power graphs and is displayed in the statistics section. The amount of work done by this joint was significantly limited. While both AMMs were being worn there was a decrease in the

amount of power absorbed during the A1 power burst, this lack of energy absorption would in turn require less energy to be generated to produce the desired propulsion during the push-off phase. While the reduction in power absorption was present, it was not to the extent in which the reduction in power generation that was observed. This directly supports the first hypothesis in that the powers from the ankle joint were able to be restricted through the use of the AMMs. The Gastrocnemius muscle, which is largely responsible for the A2 power burst did show a decrease in peak activation during the ipsilateral side and bilateral AMM trials. Reducing the activation level of the Gastrocnemius muscle during the AMM trials displays a learned behaviour, in that a normal level of muscle activation was fruitless due to the restriction of the AMM. The unilateral condition showed a marked decrease in the activation levels on the ipsilateral side as there was the ability to use an inter-limb compensation strategy. The contralateral limb showed no change in gastrocnemius activation, which does not support the point that the contralateral side during AMM walking produced a compensation strategy with the gastrocnemius for the deficiency of the plantar flexors of the AMM wearing foot, but also did not show any decreases in activity which would be expected during slowed gait. While none of gastrocnemius muscle activation strayed outside of the normal variation, there was an overall trend in the activation pattern over the varying conditions.

The gastrocnemius muscle has the role of controlling forward rotation of the tibia over the talus in a controlled manner. Normally a lack in power absorption here would allow the leg to rotate an excessive amount increasing knee flexion during early and mid-stance. This lack of activation of the bilateral and unilateral effected side conditions during the 10 – 55% of the gait cycle shows that the participant was able to adapt to wearing the AMM and use its rigidity to allow for a controlled A1 power phase to allow a similar to “normal” joint mechanics to occur. The low

level of contraction during the swing phase is presumably to act as a knee flexor to cause adequate knee flexion prior to the leg swinging through to achieve adequate toe clearance and remained similar to the normal condition as the foot is no longer on the ground and the ankle is not being manipulated. The contralateral side to the AMM showed an increase in power absorption at the A1 burst in comparison to the normal trials, which indicates that the contralateral side could be making an attempt to control the forward acceleration of the COM for the loss of control from the side effected by wearing the AMM.

The Tibialis Anterior muscle was noted to show a very similar activation pattern regardless of the condition being applied upon it. During the bilateral conditions, a small increase in activation was observed during late stance until toe off. Also, a decrease in the activation levels occurred during late swing phase right before heel strike. The tibialis anterior muscle is also used to prevent foot slap during the beginning of the gait cycle. While the AMM was placed on the foot, the participants were able to accommodate to the fact that the AMM prevented foot slap and decreased the level of activation during mid stance. This learned movement pattern is evident that enough practice was given to the participants with the AMMs before testing commenced to allow for adaptations to occur.

Tibialis anterior has also been noted to play a small role in pulling the leg forward over the foot during the stance phase. The A1 power burst is noted to be decreased during the trials with an AMM versus the normal trials, however, the activation of the Tibialis Anterior actually remained constant during these trials and over the same period of time. The rigidity of the AMM reduces the effectiveness that this muscle activation has on the shank segment. Participants were possibly trying to attempt to make up forward propulsion through this activation with little to no success.

The sagittal ankle moment remained consistent through all conditions despite the changes seen in the power and angle graphs. As joint power is equal to the moment * angular velocity, this would lead us to believe that the angular velocity has decreased. The ankle moment, which has not changed during the various conditions, suggests that the participants are still attempting to create a plantarflexor moment or using the muscle moment to allow the production of power to continue through the ankle and be effective on the centre of mass.

The ankle angle throughout the gait cycle was altered significantly with a marked decrease in plantarflexion by approximately 20^0 during the trials in which the AMM was placed on the observed foot. The blocking action of the AMMs effectively decreased the range of motion in the plantarflexion direction, but did not affect the amount of dorsiflexion. This allowed for the dorsiflexion of the ankle to occur during the stance phase of the gait cycle. The decrease in ankle range of motion limited the angular velocity that could be achieved and this in turn inhibits an increase in power generation to occur.

8.4 Knee Actions during Level Walking

This lack in propulsion provided by the ankle is, in part, compensated for by the knee. From figure 12, a clear trend is seen through the power bursts. K1 and K3 shows an increase in the power absorbed, where K0 shows a decrease in the power generated, and K2 displays an increase in the power generation during the trials in which an AMM was present on the ipsilateral leg. During the AMM bilateral walking condition, an increase, in the peak power production was not significant but the work performed at the knee was. An increase in magnitude occurred due to the restriction at both ankles as the inability for one leg to compensate through an increase in plantarflexion for the other occurred. While the unilateral conditions did invoke an increase in both power generation at K2 as well as power absorption at K1 and K3 in ipsilateral knee, the

contralateral knee displayed a slight increase during the unilateral trials. This increase is indication that an inter-limb compensation could be occurring. Inter-limb compensation occurs when the opposite limb will generate an excess of power, from normal, to try and maintain a constant level of power generation due to a lack of power being produced else where, in this case the ankle in which the AMM is placed.

The rectus femoris muscle showed an increase in activation in the bilateral and ipsilateral side of the unilateral AMM condition during the phase of peak muscular activation in accordance with the kinematic and kinetic data. The initial peak has previously been documented (Winter, 1991) as an eccentric contraction to help control weight acceptance. While there is no easily understood mechanism for the large increase in activation here, it is proposed that the increase of the rectus femoris muscle could be a way to offset the lack of the effect of the tibialis anterior due to its inability to contribute from the rigidity of the AMM. The biceps femoris muscle is used as a hip extensor during weight acceptance, just after heel contact. At this point, the biceps femoris assists the gluteus maximus in controlling the forward rotation of the thigh and to prevent the forward acceleration of the COM by stabilizing the pelvis. These functions were not altered during the AMM trials, as the biceps femoris muscle activation was not altered due to inertial contributions.

During the AMM unilateral walking condition, the ipsilateral knee showed an increase in both power absorption and power generation, for the first three knee peak power bursts. These are markedly decreased from the bilateral condition as compensation is noted earlier from the contralateral leg. This is confirmed from the EMG data as elevated activity is seen in comparison to the normal condition through the muscle activation graphs. In the rectus femoris graph a second increase seen around toe off (60% of the gait cycle) has two simultaneous functions: first

as hip flexion to pull the swinging limb forward in correspondence with the H3 power burst and secondly as knee extension to slow down the backward swinging leg and foot which corresponds with the latter half of the K3 power burst (Winter, 91). In particular, from looking at the second half of the K3 power burst, it is evident that the power witnessed here is rather similar across all four conditions, showing that an increase in the latter half of the K3 power burst is not necessary for an adequate amount of deceleration to occur.

The K4 power burst, which acts as a braking force to the shank to slow it down at the end of swing phase just before heel contact, was minimally affected by the restrictions placed on the ankle. The mass of the AMM (0.27 kg) was negligible when added to the shank segment during the walking and obstacle trials. Since the overall mass of the AMM/shank segment was similar to that of the shank segment alone, the added mass would not significantly alter the overall inertial properties of the segment. Therefore, allowing the same muscular and joint forces to control the leg during the swing phase of the gait cycle. The decrease in power absorption in the K4 power burst during the conditions compared to normal walking can be explained by the decrease in walking velocity throughout the trials. This decrease in walking velocity, allows for the shank segment to move slightly slower during swing phase, which in turn reduces its inertial magnitude at the end of swing. Since the inertia of the segment has decreased, a smaller joint power is needed to reduce its velocity to achieve a safe heel contact for the beginning of the next step. The EMG data of the biceps femoris muscle supports this as an increase in activation above the normal variance is seen across all conditions and would allow for a greater level of control of the shank /foot segment during placement of the foot during heel contact.

The difference in the amount of work done is directly related to the duration and the peak of the power burst. The peaks, showed no statistical difference across conditions. Where as, the

duration of each of the respective power bursts varied amongst conditions within a single burst, however this was not analyzed.

The difference in the AMM trials and the normal walking condition, could be the culmination of the power bursts to add in forward progression. The total work done, broken down into the generation and absorption of power at the various power bursts could be enlightening to determine if an overall difference is occurring between the conditions.

A small increase in the activation levels of the biceps femoris and the rectus femoris in the bilateral AMM condition is indicative of a co-contraction between the two muscles. This co-contraction is postulated to maintain the upright posture of the trunk during the gait cycle. The increase in rectus femoris could also be inter-limb compensation as the contralateral leg in the unilateral condition shows a minimal digression from the normal walking condition.

The knee moments displayed a trend showing that the bilateral condition and the ipsilateral side had a marked increase and the contralateral side had a small increase over the normal walking condition. The increase on the contralateral side indicates that an inter-limb compensation could be occurring during the unilateral AMM trials. The inter-limb compensation that is occurring, does not seem to be enough to completely compensate for the lack of propulsion from the affected leg. However, this compensation must be of some aid to the forward progression, otherwise it would not occur during the unilateral trials.

The variation in knee angle was quite minimal in the AMM conditions in comparison to the normal walking condition. A slight increase in knee angle of the leg being affected by the AMM is related to the small increase in moment about the knee during these conditions. This increase in the range of motion through the knee joint is thought to be of some aid by a change in the

moment arms of the knee flexors/extensors to allow a greater force to be applied through a more advantageous position.

8.5 Hip Actions during Level Walking

The hip power bursts, showed a statistical difference was present in the work generated at the H1 and H3 power bursts. A decrease in the power absorbed during the H2 power burst was noted but this was not of any significance. The bilateral walking condition did not allow for compensation to occur as there was a decrease in power generation at both ankles. Therefore, each leg was left to generate adequate propulsion to maintain a constant walking velocity. Each power burst had a significantly different amount of work done, but the peaks of these power bursts were not significantly different from each other. This would suggest that the duration of each power burst was in fact different, being longer for the bilateral and ipsilateral side conditions (Figure 16). The combined difference in the peaks and the durations of each condition would be why the total work done during a power burst would be different. An increase during the H1 power burst, would be an attempt to pull the trunk forward over the legs to help with forward propulsion of the centre of mass. This is noted to increase in order from normal to contralateral side, to the ipsilateral side and, finally to the bilateral condition being the largest. Showing that the H1 burst plays a more significant role when the tasks become more difficult and perhaps is not required to compensate during the unilateral conditions as compensation is acquired from elsewhere. This increase in the work done displays a clear trend that as the restriction of the ankle joint becomes more pronounced there is a larger amount of work produced about the proximal joint to make up for the lack of work being performed at the distal joint of the leg

The decrease in the power absorbed during the H2 power burst, has previously been noted to be a stabilization of the trunk through mid-stance and prevent any backwards acceleration (Winter,

1991). While this power burst does play this role during gait, for this particular situation, with a lack of power being generated at the ankle, this lack of power absorption could also be a mechanism to reduce the amount of work lost from the system by decreasing the peak and in turn reducing the work done by this power burst. This lack of absorption could also be due to the slower walking velocity. As the reaction forces and moments would be decreased as the inertial effects of the trunk and upper body would have been lessened, therefore resulting in a decreased amount of power being absorbed.

Finally, the H3 power burst does not display a distinctive difference in the duration of the peak (Figure 16), but does show a large increase in the peak power on the ipsilateral side of the unilateral condition. This pronounced increase is thought to be an attempt to increase the pull-off power at the hip as the ankle is no longer able to generate adequate push-off power and help initiate the swing phase. Since the dorsiflexors of the foot can no longer rotate the tibia over the talus, due to the restriction of the AMM, this forward propulsion is needed to maintain a forward progression during the beginning of the stance phase. While the segmental properties of the leg have fundamentally not been altered this pull-off power would be used to overcome the inertial forces of the leg and accelerate the mass into the swing phase. Since the foot is not in contact with the ground, the ankle has no effect even in normal walking on the propulsion during the swing phase of the gait cycle, which is why the mechanics during this phase seem to be relatively unchanged during the imposed conditions.

The increase in the power generated and the decrease in power absorbed at the hip joint, along with the increase in power generated and absorbed at the knee shows a clear ability that the knee and the hip joint are able to compensate for the lack of power production at the ankle. The gluteus maximus muscle has an increase in recruitment during mid stance or at the latter half of

the first activation burst that is seen. This muscle activation is known to help with forward propulsion and the increase in the bilateral AMM condition indicates an attempt to accomplish this. This level of activation seems to be a last resort effort to maintain the desired level of forward propulsion as there is a limited increase in the activation level of this muscle during the unilateral condition in either the altered leg or the non- altered leg.

In accordance with the kinematic and kinetic data, the leg that was not wearing the AMM increased the activation of the gluteus medius muscle to perhaps provide an assist in swinging the contralateral limb. This function has been previously documented, in which the anterior fibres act to anteriorly rotate the pelvis since the thigh is in weight bearing and to assist the contralateral limb in the beginning of the swing phase (Winter 91).

The second increase of the rectus femoris muscle, as seen around toe off (60% of the gait cycle), has two simultaneous functions of different magnitudes. The lesser function being knee extension, to slow down the backward swinging leg and foot, which corresponds with the latter half of the K3 power burst, while the more predominate function being hip flexion, to pull the swinging limb forward in correspondence with the H3 power burst (Winter, 91). This accounts for the increase seen in the H3 power burst during the trials with an AMM. This increase as previously mentioned, is to aid in the pull-off power that is needed to initiate the leg into swing phase. As the inertial changes to the shank-foot segment are minimal due to the addition of the AMM, it is thought that the secondary function of knee extension is not as important. From looking at the second half of the K3 power burst, it is evident that the power witnessed here is rather similar across all four conditions, showing that an increase in the latter half of the K3 power burst is not necessary in order for an adequate amount of deceleration to occur.

The peak hip moments did not significantly change across the four conditions. The peak maximum moments for the AMM trials increased < 5%. While none of these changes were significant and the values are quite low, a combination of knee and hip moments could play a vital role in the summation of the moments to counteract the loss of the plantarflexor moment. The range of motion of the hip, stayed very consistent during the four conditions, not varying more than 2^0 between any two conditions. This is a good indication that the trajectory of the hip is able to stay consistent despite the limitations that were placed upon the limb more distally. As the gait velocity decreases, a decrease in the hip angle primarily during the beginning of the stance phase is observed (Winter, 1991). This difference in hip angle across different walking velocities was reported as 4.1^0, and had a correlation factor of 0.99 from slow to natural walking (Winter 91). While the hip range of motion did not deviate to this extent between the conditions, the decrease in gait velocity could be one explanation to the change in hip angle as well as the overall decrease in the range of motion, regardless of the extent.

8.6 Temporal Measures

As the results of the kinetic/kinematic data was not robust as previously anticipated, it was thought that the individual variation of each participant was being lost as data was collapsed across subjects. As walking velocity is the product of cadence by step length, the data of each participant was focussed on individually. Each participant showed one of two trends during the bilateral condition in comparison to the normal walking condition. Six of the participants displayed an increase in cadence and a decrease in stride length during the bilateral condition, while having the walking velocity stay at a relatively constant value. This trade off, of cadence and stride length has previously been reported in detail (Winter, 91). These participants also showed an increase in the stance ratio and decrease in swing ratio during the bilateral AMM

condition. Interestingly, all of these subjects did not display the previously reported “normal” stance/swing split of 60% stance/40% swing. Most participants in this subgroup performed a gait pattern that consisted of a 40 - 45% stance phase to a 60 – 55% swing phase during the normal walking condition. While this subgroup preferred to walk at a faster cadence and smaller stride length, the 60%/40% split should still apply. During the bilateral condition, this subgroup had the 60%/40% split. This initial change from the “normal” walking stance/ swing ratios, could be a possible mechanism in dealing with the instrumentation that was placed on them, once the added difficulty of the AMM was applied a return to the “normal” ratio did occur.

Seven participants made up a second subgroup that showed an opposing trend, of a decrease in cadence and an increase in stride length from the bilateral condition in comparison to the normal walking condition. These participants initial had a stance swing ratio that was very close the the 60% /40% split that would be expected. During the bilateral walking condition, this subgroup tended to decrease the stance ratio to approx 45 – 50% and increase the swing ratio to 50-55%. While this subgroup increased their swing ratio, it was most likely a by-product of increasing the stride length as the foot must travel a farther distance. However, while the stride would take a longer amount of time to be completed, the amount of time spent in stance vs. swing would be thought to remain constant.

One participant (Female #1) showed a decrease in cadence while maintaining stride length, however, the walking velocity slightly increased between the two conditions. To further divide the group and analysis each subgroups kinetic/kinematic data would be a logical next step for future research, so that the means of particular variables, work done, power, moments, etc, would not be lost in the differences between two distinctive groups.

8.7 Other EMG

The rectus abdominus muscle showed a very consistent pattern across the conditions. Since the rectus abdominus muscle has a marginal role in walking, the low steady amplitude of this muscle is to be expected. The activity of this muscle may be a minor co-contraction to the erector spinae muscles, which are used to balance the spinal column. During the normal walking condition, a very slight increase in amplitude can be determined at both 20% and 70% of the gait cycle. By looking at the normal variance this becomes much more evident.

Perhaps a slight increase in stabilization of the trunk is indeed needed during the conditions that are affected by the wearing of an AMM. The dynamic stability of the lower limbs through the foot may no longer be adequate to keep the upper body in a stable position due to the lack of exproprioception due to the rigid AMM being under foot. The rigidity of the AMM does not allow for subtle movement changes of the ankle to correct for the centre of mass and perhaps centre of pressure and compensation of the rectus abdominus and the erector spinae is needed in minimal amounts to maintain the upper body's stable.

The erector spinae muscles show a large peak right after heel strike and another during the end of stance. Each peak serves the same role, of controlling the forward acceleration of the trunk as each limb accepts weight. The first peak is associated with the weight acceptance of the right limb, while the middle peak is associated with weight acceptance of the left limb.

During the unilateral condition, the ipsilateral side to the AMM shows an increase in activation during early to mid stance phase. After approx. 40% of the gait cycle, a decrease in the total muscle activation occurs relative to the normal walking condition. For the leg in which the AMM was worn, a decrease in activation compared to the normal walking condition is seen throughout the entire gait cycle. This would lead to the conclusion that a minimal amount of compensation is

occurring via the stability of the erector spinae muscles. However, for an inter-limb compensation strategy to occur a stable connection is needed between the two halves of the lower body. The pelvis and lower trunk facilitate this, supplying the linkage that the system needs to be able to transfer power and energy from one leg to the other and also to propel the body forward.

During the bilateral AMM condition, the braking force normally being applied by the gastrocnemius during the early stance phase is inhibited to do so from the restrictions placed on it from the AMM. Controlling the forward acceleration of the trunk is therefore placed on a combination of the gastrocnemius working with the rigidity of the AMM and the erector spinae muscles. This is displayed through an increase in the erector spinae muscles in activation and duration in the bilateral as well as the non-effected side of the unilateral conditions. The effected side does not show an increase in activation and is not remarkable in comparison to the normal condition.

The central peak is associated with weight acceptance of the left foot. The peak activation for the AMM conditions were all less than the normal condition, however, these were not of statistical significance. There seems to be a decrease in effort for the erector spinae to control the forward acceleration of the trunk as the left leg is accepting weight. This small decrease may be due to the small changes in gait velocity that was observed or perhaps to maintain some of the forward propulsion by not inhibiting the forward rotation of the trunk. As the erector spinae muscle act in an eccentric fashion to inhibit forward rotation, a lack of inertia would correspond with the lack of muscle activation as it is no longer necessary to overcome these forces.

8.8 Ankle Actions during Obstacle Walking

During the obstacle trials, participants were required to step over a 30 cm obstacle which was placed in the middle of the swing phase for the right foot. Therefore, participant's right limb would be the lead limb over the obstacle regardless of the combination of the AMM placement. While the AMMs were worn bilaterally, the desired effect again occurred by decreasing the amount of ankle power generated as well as the peak power produced throughout the gait cycle. The contralateral side to the AMM showed a slight decrease in peak power, which could be attributed to the decrease in walking velocity that was discussed earlier. These changes were expected since a similar pattern occurred during the level ground walking conditions. The absorption of power during the first 40% of the gait cycle is not easily distinguishable in the EMG data. No significant differences were seen in either the gastrocnemius or the tibialis anterior. The amount of work done from the ankle was also limited during the A1 power burst, a significant decrease in the work done by the ipsilateral side of the unilateral condition and the bilateral condition were seen, however, an increase in power absorption from the contralateral side was observed. This increase is thought to be an effort to slow down the forward progression of the COM just before they reach the obstacle and need to step over it. During the contralateral condition, the lead limb did not have the AMM placed on the ankle. Therefore, an adequate "push off" force would able to be provided enough height to safely clear the obstacle as well as provide the desired propulsion.

The propulsion from the A2 power burst also had a similar trend to the level ground walking conditions. A decrease in the peak power burst during the bilateral and the ipsilateral side of the unilateral conditions seen and a slight increase was again observed in the contralateral side. This decrease in propulsion from the ankle during the "push-off" phase, again showed that the AMM

was able to restrict the ankle power during those conditions and supports the first hypothesis. The timing of the ankle power profile during the obstacle conditions is shifted earlier during the gait cycle in comparison to the level walking conditions. This is due to the prolonged swing phase that occurs while clearing the obstacle. The muscle activation pattern of the gastrocnemius muscle were similar across conditions and to the level walking condition with the exception of the major activation burst being earlier during the gait cycle. A very minute increase in activation is seen just prior to heel contact at the end of the gait cycle. This burst was to attempt to have the toes be placed on the ground first rather than the heel, as this would allow the participant to have a second contact point with the ground sooner than if they wanted for the heel to touch down. This plantarflexion of the ankle is also noted in the ankle angles of the normal and contralateral conditions. The work done during the A2 power burst was significantly diminished during the obstacle trials. Since the decrease in the ipsilateral side was larger than the bilateral condition, this would lend weight to the theory that the opposite limb was able to compensating for the lack in propulsion from the ipsilateral limb and therefore the ipsilateral limb did not attempt to generate as much power from the ankle joint. The gastrocnemius muscle during the normal obstacle trials would create an increase in the elevation of the COM through an earlier plantarflexion, since this is inhibited by the rigidity of the AMM, this increase in elevation is needed to come from elsewhere in order to safely clear the obstacle.

The tibialis anterior muscle showed one of two activation patterns depending on the participant. The first muscle pattern displayed a decrease in activation in the unilateral AMM condition; the contralateral foot's activation pattern mirrored those of the normal walking condition, with the exception of an increase in the amplitude and duration of the activation burst during toe-off. This increase in activation is an attempt to increase the toe clearance over the obstacle through the use

of dorsiflexion. Since the AMM is inhibiting the gastrocnemius muscle from supplying the vertical translation of the pelvis through plantarflexion. This elevation through the ankle is known as “Vaulting”, which is a pushing up onto the toes of the foot to increase the vertical translation of the hips. The first muscle pattern seen in Figure 41 shows no deviation outside of the normal variance, but the second muscle pattern (Figure 42) for the tibialis anterior muscle was shown by two female subjects. Both subject displayed a decrease in activation during the bilateral and ipsilateral conditions through the normal activation bursts. This decrease in muscle recruitment is thought to be a learned reliance on the structural stiffness of the AMM to prevent the toes with coming in contact with the obstacle. The second decrease in muscle activation is to avoid inhibiting an attempt to plantarflex the foot in the attempt to touch down with the toes first after the obstacle. Both participants were observed to prefer a toes first landing when the ankle was not restricted by the AMM. This was confirmed through the kinetic data, primarily the ankle angle during late swing. During the unilateral conditions, an increase in the gastrocnemius muscle of the contralateral foot (not wearing the AMM) is able provide adequate vertical displacement to achieve a safe distance of the toes from the top of the obstacle. Once the obstacle was cleared, the gastrocnemius muscle returned to a lower “relaxed” level of activation at 80% of the gait cycle.

The moment of force that was produced at the ankle was very similar across all conditions, which was of interest, since the range of motion of the ankle joint as well as the power absorbed and generated had significant differences. Since the ankle joint power, which decreased during the AMM conditions, is comprised of the ankle moment and the ankle angular velocity, a decrease would be expected in at least one of the two variables. Since the moment about the

ankle did not change, this would suggest that the angular velocity decreased substantially during the AMM conditions.

The angular velocity can be seen from the slope of the line representing ankle angle in Figure 32. The slopes of the lines representing the no AMM condition as well as the contralateral side to the AMM are markedly increased in comparison to the bilateral and ipsilateral conditions. This is due to the decreased range of motion that the joint can functionally use and therefore it takes less time for the end of the range of motion to be reached. There were very slight differences between the peak moments during each condition, but these were statistically insignificant and lend value to the theory of angular velocity being the limiting factor in the power production.

The plantarflexion of the ankle was highly limited during the AMM trials. The range of motion of the ankle during the middle of the gait cycle, or where toe off would occur for the obstacle conditions, was therefore restricted to a range that is approximately 10^0 less then when it is not restricted. While this seems to be a minimal amount, these 10^0 of motion are during the critical phase of plantarflexion to provide the participant with adequate "push-off" force to not only provide propulsion but also elevation of the leg in the obstacle trials.

8.9 Knee Actions during Obstacle Walking

The biggest difference between the knee power bursts of level walking and obstacle walking is that an extra K5 power burst will occur in the obstacle trials. This K5 burst increases the hip flexion during obstacle avoidance instead of increasing the H3 burst to increase the hip angle to elevate the distal end of the femur and effectively elevate the toes. While walking with an AMM on level ground is able to perturb the system and force changes and compensations to occur in both the knee and the hip, an added challenge of stepping over an obstacle could lend more

insight into the system as now the knee is further constrained in that it is required to generate the K5 power burst and perhaps will not be as well suited to assist in the generation of power to provide forward propulsion.

All six power bursts showed significant statistical differences in the peak powers across conditions and for the work done at the knee joint except for the K4 and K5 bursts. While differences did occur at the K0 power burst, this is not of clinical significance, just like the level walking trials, a small change in the power generated would cause a larger difference between the conditions as the power generated here is of such small magnitude. Again the K0 peak power was not determined due to the variable nature of this value.

The increase in power absorption as well as the peak of the K1 power burst for the AMM conditions hints to the fact that during the early stance phase the knee plays a very dominate role in weight acceptance in comparison to normal walking obstacle avoidance. The K2 power burst displayed an increase in both peak power and power generated. This power burst has traditionally been associated with a straightening out of the leg after weight acceptance, during the AMM conditions, it is proposed that the increase in power generation could contribute to the overall forward propulsion to help assist for the lack of power generated at the ankle.

While these increases were statistical significant, the muscle activation profiles for both the rectus femoris and the biceps femoris muscles showed no deviation from the normal variance across any of the conditions during the stance phase. A small increase in the rectus femoris muscle is noted during early stance which corresponds to the end of the K1 power burst and the beginning of K2 to aid in the absorption of work, but more importantly to create a generation of power about the knee.

The K3 power burst slows down the leg through knee flexion and prevents the foot from coming in contact with the participants gluts. The increase in the absorption of power here is not intuitive as the inertial properties of the shank/foot/AMM segment are relatively similar and would not require an increase in power due to the placement of the AMM. During this time however a decrease in activation is observed from the biceps femoris muscle and no change occurs to the rectus femoris muscle. This decrease of the biceps femoris would allow for the rectus femoris to have a larger influence on the joint and create a larger power burst, even though it is an absorption of power, to perhaps help initiate the forward swing phase of the foot/shank segment.

The K5 power burst, previously mentioned, is only observed in obstacle walking. This power burst is utilized to increase the hip flexion and will generate a rotational energy to induce a movement-dependant flexor moment about the hip (McFadyen & Winter, 1991; Niang & McFadyen, 2004). The power generated at this power burst was not significantly different across conditions, however, the peak power was statistically significant. Due to the lack of power generation observed at the ankle, it would be intuitive to believe that this power burst would increase while an AMM was present rather than decrease the amount of work it is generating. As the foot is no longer on the ground, the main objective of the K5 power burst is to achieve adequate knee elevation through flexing the hip. The previous K3 absorption of power was increased while an AMM was present, in an attempt to initiate forward swing of the leg. The increase in momentum of the leg from the increase of the K3 burst would therefore require less power from the K5 burst in order to achieve an adequate power to ultimately achieve a satisfactory level of knee elevation to safely clear the obstacle with the toes.

Once a satisfactory level of toe clearance or max toe height is achieve the K4 burst starts to slow down the leg to prevent hyperextension and place the foot for the next heel strike. No differences were seen in the absorption of power, which is to be expected as the inertial properties of the shank/foot segment are relatively similar across condition and would therefore require a similar power to bring the foot to a safe contact velocity in preparation for heel strike. The peak powers did show a difference as the AMM conditions were reduced in comparison to the normal condition. Since the walking velocity during the obstacle conditions were of a significant difference, this could be the reason to have a decrease in power absorption, since the shank/foot segment was moving at a slower rate relative to the overall movement.

The normal obstacle condition shows an increase in the rectus femoris activation level at approx 80% of the gait cycle, which is in correspondence with an increase in hip flexion but would be in opposition to the desired knee flexion. Therefore causing an increase in the level of co-contraction about the knee and thus creating a need for an increase in the biceps femoris muscle. The increase of the rectus femoris muscle corresponds with the K5 power burst to elevate the leg to allow it to clear the obstacle.

At this point a significant increase in activation occurred in both the bilateral condition as well as the ipsilateral side of the unilateral condition. This increase in activation is thought to be used as a hip flexor rather than a knee extensor to increase the height of the distal end of the femur to counter the inability to create a vertical translation of the pelvis through a plantarflexion of the toes.

The biceps femoris muscle showed a similar pattern during the obstacle conditions with the addition of an activation peak at approx 45 – 50 % of the gait cycle. This peak was used by participants to increase the flexion at the knee before swing phase in anticipation to clearing the

obstacle. This was done so that when the leg was swung over the obstacle, the toes would be in an elevated position. Within the obstacle condition, there is no deviation of the AMM conditions outside of the normal variance. Since the AMM is of negligible weight in comparison to the shank/foot segment, this level of activation to cause flexion in the knee, would be sufficient to achieve the desired results.

While there were no statistical differences between the mean peak moments produced about the knee during the four conditions, a point of interest did arise showing that the AMM conditions had an increase over the normal walking condition. The inter-limb compensation that is occurring does not seem to be enough to completely compensate for the lack of propulsion from the affected leg, which was previously seen in the level walking trials and hold true for the obstacle condition as well. However, this compensation must be of some aid to the forward progression, otherwise it would not occur during the unilateral trials.

As previously mentioned, the variation in knee angle was quite small, with an increase in the bilateral and ipsilateral side and a decrease in the contralateral side. This increase in knee angle of the lead leg over the obstacle and also being the leg affected by the AMM is due to the increase in muscular activation of the rectus femoris through the mid swing phase of the trials. The near normal trajectory of the knee angle remains consistent and within a very narrow range of variance regardless of condition, which infers that the mechanism used to clear the obstacle was successful as the outcome had the desired effect. This was also determined from the consistency of the max toe height which was achieved. A slight increase in max toe height was observed during the trials in which the leading leg had an AMM. Since the max toe height actually increased during the AMM conditions, it is evident that the overall movement pattern is not compromised by the lack of plantarflexion or power generation at the ankle joint, which

could be hypothesized is a decrease in max toe height had occurred in only the conditions were the lead limb was wearing the AMM.

8.10 Hip Actions during Obstacle Walking

The hip power bursts show the same differences across conditions during the stance phase as was seen during the level ground walking conditions. The H1 power burst serves the same function during the obstacle trials, that of helping pull the trunk forward over the stance leg during early stance; this increase is needed to overcome the restriction placed on the ankle joint and to help in the forward propulsion of the body. During the unilateral AMM trials, there is an increase in the ipsilateral leg but there is also an increase in the contralateral hip flexors, which again indicate the uses of a compensation mechanism to help the participant with forward propulsion. The bilateral trial showed an increase in the amount of power generated by the joint. While all of these changes seem to be quite large, the magnitude of the work done by the hip joint was quite small (0.09 – 0.13 J/kg), therefore making any small change a significant amount in terms of percentage but not having any clinical significance.

The H2 power burst, just like the level walking trials, shows the opposite trend, where the power absorbed decreased for all AMM conditions. Neither the peak powers nor the amount of power absorbed were of statistical significance. This decrease is thought to be due to the slower walking velocities of participants during the obstacle conditions.

The generation of power during the H3 power burst did significantly change across conditions. The increase in power generation that occurs during this power burst are expected as this H3 power burst is the “pull-off” force to initiate the swing phase of the gait cycle. A slight increase is noted to occur during the conditions where an AMM is involved, but the overall initial

properties of the leg have not changed and therefore the required amount of work to swing the leg through the swing phase of the gait cycle should remain the same. However, this increase in power generation could be an alternate location in which the loss of ankle power could be made up. Since the timings of both the H3 and the A2 power bursts happen at the same point during the gait cycle, it is logical that this would be an optimal location for this lack of propulsion to compensated for.

From the EMG patterns, the gluteus maximus muscle has an increase in recruitment during mid stance or at the latter half of the first activation burst that is seen. This muscle activation is known to help with forward propulsion; however a decrease in the bilateral AMM condition and the ipsilateral side of the unilateral condition would suggest that the increase seen through the level walking conditions were not transferred over to the obstacle walking conditions. The increases seen during the level walking conditions would actually hinder the participants here as there is the need to "relax" the gluteus maximus muscle to allow an adequate amount of hip flexion to occur in order to safely clear the obstacle.

The central peak seen in Figure 46 is characteristic of an increase in muscle recruitment, since the gluteus maximus muscle is a hip extensor and assists in controlling forward thigh rotation, it is directly linked to the amount of knee flexion that would occur. This increase in activation thus, attempts to increase the knee flexion to safely complete the obstacle trials. While minor differences did exist between the obstacle conditions, none were of statistical significance for these movements. The gluteus maximus serves the same purposes during level ground walking as well as obstacle walking, controlling hip flexion during weight acceptance as well as to decelerate the forward swinging thigh and reverse its rotation through the swing phase.

This “pull-off” power is aided by the second activation burst seen during the rectus femoris EMG profile which has a second function of hip flexion to pull through the swinging limb. This accounts for the increase seen in the H3 power burst during the conditions with an AMM. This increase is to aid in the “pull-off” power that is needed to make up for the lack of “push-off” from the gastrocnemius muscle. This “pull-off” power witnessed as a large increase in activation of the rectus femoris, a decrease in activation of the biceps femoris and gluteus maximus is to maximize the flexor moment of the hip to attempt to make up for the loss of the plantarflexor moment which is being hindered due to the AMM. This is evident when observing the peak power produced about the hip during the H3 power burst. Both instances when the AMM is placed on the lead limb (bilateral and ipsilateral conditions) an increase in the peak power is seen (Figure 37). During the condition when the AMM is on the contralateral foot, or the trailing limb over the obstacle, the leading limb is able to generate the plantarflexor moment to achieve the desired “push-off” force and get the needed toe clearance through translation of the pelvis. Gluteus Medius did not show a significant amount of variation across conditions. This large contraction during the beginning of the stance phase has previously been noted in the literature (Winter, 91) to be a co-contraction against the hip flexors as well as control the drop of the pelvis during weight acceptance. The increase in the bilateral condition as well as the contralateral side of the unilateral condition indicate that gluteus medius from the opposite side to that of the AMM is attempting to rotate the pelvis to assist in swinging the limb as the A2 push-off power has been diminished. This “pull-off” power is seen during both level ground walking as well as obstacle avoidance tasks. There seemed to be no alteration of the muscle activation pattern in timing or amplitude to show that it aided in obstacle avoidance.

The peak hip moments did not significantly change across the four conditions. However, a combination of knee and hip moments could play a vital role in the summation of the moments to counteract the loss of the plantarflexor power. While it would be expected that the moments about the hip would increase to clear the obstacle, especially during the AMM conditions, this did not occur as the muscular contributions differed to acquire the same result through an increase in the moment about the knee.

The range of motion of the hip, stayed very consistent during the four conditions, not varying more than 3^0 between any of the conditions. This is a good indication that the trajectory of the hip is able to stay consistent despite the limitations that were placed upon the limb more distally. However, a trend was present that showed that a decrease in the range of motion occurred during the conditions which included the AMMs versus the condition that omitted them. Again, the plasticity of the mechanics of the lower body is shown here as the desired result was still obtained even though a different means was needed to achieve this.

8.11 Temporal Measures

From the temporal measures, two subgroups again were able to be distinguished amongst the participants. The first group were able to maintain walking speed, with the majority accomplishing this through increasing stride length and decreasing cadence. While this seem counterintuitive, the temporal measures were calculated through the one full stride that included the obstacle at mid-swing. The increase in stride length is thought to be a mechanism for reaching the foot over the obstacle and is not indicative of a consistent stride length for the steps leading up to over following the obstacle.

The second group showed a decrease in walking velocity over the obstacle and this subgroup was divided 4 participants having a decrease in cadence and an increase in stride length while the other four had an increase in cadence with a decrease in stride length. While the subject had a increase in one variable and a decrease in the other, it is noted that since walking velocity was not maintained that the decrease in first variable was larger then the increase in the second variable.

The large variability across subjects displays a lack in consistency between the methods utilized to complete the task. It would seem to be that this population does not have one particular response in overcoming the conditions placed before them and that the spatial temporal data is not as helpful or telling as it was in the level ground walking conditions.

However, since the temporal variables were only taken from the stride in which the obstacle was involved a bias among the data could be a factor in that participants may be reaching to achieve a foot placement, spending more time in swing and/or less time in stance would change both sets of ratios and, changing the walking velocity to either clear the obstacle more quickly or to have more time to anticipate for the obstacle task.

8.12 Other EMG

The rectus abdominus muscle showed a consistent activation pattern across the 4 conditions during the obstacle trials. These activation patterns were similar to those presented in the level walking conditions, with the exception of a large increase in muscle recruitment through mid swing phase. While the rectus abdominus has been previously noted to play a minor role in walking, this increase in activation suggests that an increase in stabilization of the trunk is needed when participants raise their leg to clear the obstacle. The increase in activation of the

rectus abdominus would be to prevent the pelvis from tilting sideways or rotating about the anterior/posterior axis. The gluteus medius is also involved in this role, so to not see an increase in activation was unexpected. The increase in stabilization of the trunk through the increase in co-contraction of the rectus abdominus and the erector spinae muscles is used to aid in the increase of hip flexion to raise the leg in preparation to step over the obstacle.

The erector spinae muscle displayed a large amplitude power burst at approx. 40% of the gait cycle. This increase in activation was similar during the level walking trials and would have the same role, being weight acceptance of the contralateral limb, through the gait cycle. The bilateral condition showed an increase in activation during this period, but it did not deviate outside of the normal range of variance determined by the normal obstacle condition. This increase in recruitment could be a subtle increase in stabilization of the trunk to allow for a steady foot placement as the limited control of the foot, which is restricted by the AMM, would hinder this finer level of control.

The increase of activation at the end (90 – 100%) of the gait cycle was much larger during the obstacle trials then during the level walking conditions. This activation burst would be in preparation for weight acceptance of the leading limb. An increase in activation here would indicate that an increase in control is needed as the foot needs to be placed on the floor again. While there is a difference between the normal and the obstacle trials, the difference among the four obstacle conditions did not differ outside of the range of the normal variance. The trend that was observed was a delay in onset of activation of the erector spinae muscle but the amplitude of the activation was similar except for the contralateral side where activation of the erectors had decreased slightly.

8.13 General Adaptations due to minimization of Ankle Motion

From the kinetic and kinematic data, a large reduction in ankle power was achieved through the use of an AMM. This lack of power generation via the A2 power burst was not noticeably taken up in such magnitude at either the knee or the hip. The plasticity of the lower limb has been previous documented (Winter, 1991) with the largest trade-off between joints occurring between the knee and hip joint moments. The trade-off between the knee and the hip is most evident through the support phase of the gait cycle. Both knee and hip extension are required to maintain the centre of mass from collapsing. A small decrease in the amount of hip extension can be counteracted by an equal increase in the amount of knee extension as compensation. The variability of the joint moment can be observed across different individuals, however, as long as the desired support moment is achieved, the separate ankle, knee and hip moments can vary within reason to provide the support moment.

This trade-off is also seen with respect to the joint powers as the joint power is comprised of the joint moment and the angular velocity. As previously mentioned, the support moment was looked at through the use of the joint moments, in the same way that the propulsion and thrust was observed through the use of the individual joint powers. The trade-off through the various joint powers is similar to the support moment in that it is predominately comprised between the knee and the hip powers. However, this trade-off is seen to a smaller extent from the ankle and that a loss in ankle power resulted in smaller increases at both the knee and hip at multiple power bursts.

Performing a summation of all of the power generations and absorptions at each joint across the conditions would not provide any insight as an arthimatic summation would not divulge the

implications that would arise from a functional stand point. The combination of interactions from the power generations and absorptions at each joint allowed the limb to be successful in providing forward propulsion to the centre of mass. This interaction between the different power bursts makes it difficult to pinpoint the compensations that are occurring as the response is spread over various joints and during various power bursts.

While, Vanderpool et al. 2008, had previously determined a similar finding, through the use of energetics, that these were taken up partially by both the hip and the knee without a distinct point of compensation, this current study focused on the joint powers and the work performed.

While Vanderpool focused on the changes that occurred through the use of air casts, to minimize ankle motion, with a rocker bottom attached, this also changed the weight of the segment and the rocker bottom provided some transfer of energies to aid in the recovery of the lost in power generation. With the rocker bottom placed on the sole of the air cast, it is questionable that the movement has changed too much from normal walking to be an adequate comparison. The use of the AMM did not substantially change the overall weight of the lower shank/foot segment and did not introduce a new rocking motion into the task to allow for a fair comparison while limiting ankle motion throughout the gait cycle.

The obstacle conditions placed a addition set of demands on the lower limbs as not only did the loss of ankle power need to be overcome, but an obstacle had to be stepped over during mid-swing of the gait cycle. During the obstacle task, an extra power burst, K5, is needed to aid in elevating the knee to an adequate level and flex the adjacent joints to avoid the toes coming in contact with the obstacle. With this extra power burst that is required, the ability of the knee to

still aid the loss of ankle power is questionable. Therefore, requiring the hip to play a larger role in compensation throughout these trials.

While, a lack in ankle movement will effect numerous variables, such as weight support and braking ability, the focus of this investigation was placed on forward progression through joint powers and the work done at these joints. The means of most variables are a common way to express the trend of a group and the mechanisms that are being utilized to achieve a certain task. However, participants will not always perform the task in the same way and can have quite large variability in their measures when collapsed over all the participants. Therefore, the segregation of participants into subgroup based on movement patterns, since this investigation focused on variables at the joint level, is paramount in determining consistent changes across subjects. While, this was witnessed at the joint level, a greater variability can be seen at the level of muscle activation, especially when the muscle spans across two joints. This level of plasticity is supremely important to allow for different movements to be performed as the situations deems necessary.

Applications in a clinical Environment

This investigation showed that the plasticity of the lower limbs during gait is useful in overcoming a situation in which ankle plantarflexion is restricted. While this investigation was performed on healthy, able-bodied participants and was done from a phenominalogical stand point, the documentation of the plasticity is useful for those that have ineffective ankle contributions that other joints could be able to compensate for. While most patients who do not have the ability to generate power through the A2 power burst would be walking at a markedly

decreased velocity to the ones reported here, it is a starting point to understanding how to best utilize the remaining joints to produce power by the most effective means to try and overcome the loss in forward propulsion at the ankle.

8.12 Limitations:

This investigation investigated the effects of limiting plantarflexion about the ankle during level walking as well as obstacle walking tasks. No real bilateral data was collected due to the limitations of lab geometry as it was not possible to position the cameras in such a way to capture markers from the left side of the body. While data was collected for the bilateral trials, meaning when the AMMs were worn on both feet, it was collected for the right side of the body and inferred to the left side. Unilateral trials were also collected on the right side of the body, so while the AMM was on the right leg data was collected for the ipsilateral limb and while the AMM was on the left leg, then data was collected on the right leg, which was then the contralateral limb.

Since data was only collected on the right side of the body, it was not as easy to capture the frontal plane to determine pelvic tilt even though both greater trochanters and iliac crests were captured with motion capture markers.

Also three of the female participants had an ACL replacement surgery in the past. These surgeries had not been in the last 2 years and when their data was collected and compared to the other female participants, who had not had an ACL replacement; there was no difference in any of the kinematic or kinetic data, or any distinguishable differences in the EMG profiles from any of the eight muscles collected. While no differences were seen between these participants and the

rest of the female participants, subtle differences could possibly exist which would distort the mean of each of the variables that were determined.

8.13 Suggestions for future research

Further investigations could directly address the limitations of this study, mainly by collecting data from both sides of the body, stiffening the AMM further to try and fully lock the ankle and not being easily able to capture pelvis tilt in the frontal plane.

By collecting data from both sides of the body, this would decrease the number of trials by 20 (25% of total trials) and thus reduce the collection time needed for each subject. Due to laboratory and equipment constrains this was not possible, so only the right side was captured and this was then inferred to the left side of the body. The ability to know what both sides of the body were doing versus inferring from one side to the other would add fidelity to the study as a whole. Also adding trunk markers would give an indication of trunk angle which was unable to be determined, as it was an afterthought to the analysis.

Unfortunately, the temporal data was not looked at until after the kinetic/kinematic analysis had been complete. This temporal data clearly suggests that different subgroup did exist in the larger subject pool. A segregation of these subgroups for a full analysis of the kinetic/kinematic data might show a clearer picture of the mechanisms that are occurring during a restriction of the ankle joint. Since this was not completed, a difference in one group and an opposing difference in another group could cause neither difference to be noted as they would directly counteract one another.

A similar study that only allowed for a 1-2^0 ROM for the ankle instead of 20^0 minimum would lead to more insight as the ankle joint would be severely restricted and the trends that were seen

in this study might then have some statistical significance due to the decrease in ankle range of motion leading to a more concrete conclusion of events and strategies that were occurring. The ability to capture the pelvis in its entirety in the frontal plane would be helpful in determining if pelvic tilt was occurring during some of the conditions. From this, a better understanding of hip hiking could be investigated and its contribution to the overall strategy and compensation mechanisms that were taking place.

9.0 Conclusions

In identifying differences in kinematics, kinetics and EMG data of level ground walking and obstacle avoidance tasks, this investigation yield two main conclusions based on the specific aims and hypotheses:

- Propulsion that is contributed from the ankle joint through the A2 power phase is significantly diminished during both ground walking and obstacle avoidance trials, when participants are wearing an AMM when compared to not wearing an AMM.
- Compensation that occurred due to the AMM placement was taken up predominately by the knee rather than the hip, even though an increase in the amount of work done increased in both joints.

This demonstrates that the characteristic gait pattern is plastic enough to maintain a consistent walking velocity despite the restrictions, being decrease ankle plantarflexion, that were imposed upon it.

The current investigation was able to quantify the angles, moments and powers about the ankle, knee and hip to complete both walking on level ground as well as walking over an obstacle. The conditions were comprised of walking without AMMs, walking with AMMs placed bilaterally, and walking with an AMM placed unilaterally. All participants were able to complete all tasks with minimal difficulty.

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Appendix A: Information and Consent Form

Information and Consent Form
Department of Kinesiology
University of Waterloo

Study Title: Contribution of the ankle joint to support, propulsion, and positioning during gait.

Conducted by:

Student Investigator: Eoghan Landy Email: e3landy@ahsmail.uwaterloo.ca

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The goal of this study is to better understand the muscular contributions to while the ankle joint is being restricted in its contribution. The data that is collected in this experiment will be used to help determine the muscular demands and the compensation strategies of ankle restriction using Ankle Foot Orthotics, while using normal ground walking as a baseline. This study is being conducted by Eoghan Landy, under the supervision of Dr. Stephen Prentice.

Though you may not benefit personally from participation in this study, your participation will help researchers gain a better understanding of the muscular demands and compensation strategies required to walk when minimal power output can be generated at the ankles and how this relates to level ground walking.

If you have had surgery on your lower limb in the past year, or have a musculoskeletal or neurological condition that you believe may influence your ability to participate in the study, you should not participate.

The first portion of the study will *involve* your height and weight being recorded.

Pairs of small electrodes will be placed over a number of muscles of your legs in order to measure the activity of the muscles controlling your leg. To make this process easier you will need to wear shorts, a t-shirt and comfortable running shoes during the experiment. These electrodes only measure electrical activity and there is no chance of you being shocked by these electrodes, as the electrodes are optically isolated from any electrical sources. These electrodes will be placed over muscles on the calf (Gastrocnemius), shin (Tibialis Anterior), Quadriceps Group on the front of the thigh (Rectus Femoris), Hamstring group on the back of the thigh (Biceps Femoris and Semitendinosis) on both the right and left legs. Two electrodes will be used for each muscle; therefore a total of 16 electrodes will be used. In order to achieve a good quality signal, the areas where these electrodes are placed will have to be shaved, and cleaned with rubbing alcohol.

Once the muscle electrodes have been placed, Twenty-six infra-red light emitting diodes (IRED markers) will then be attached to your right and left lower limbs and pelvis by placing the markers on rigid plastic shells which will be attached to your leg using tensor bandages and medical-grade skin tape. Four markers will be placed on each plastic shell, and these will be placed over your thigh, lower leg and feet of each leg and a plastic plate will have 4 markers which will be attached to your pelvis by an elastic belt. These markers will be used to track the position of your legs and pelvis. The markers will allow special cameras and a computer system to track your movements.

Once the markers are in place the locations of the joints will be digitized by a research assistant. In order to do this you will be asked to stand in the view of the cameras, and the research assistant will determine the location of the joints of the lower limb, and some bony prominences (skeletal anatomical landmarks).
You will then begin the experimental tasks. You will be asked to walk across a room during normal level ground walking as well as normal ground walking while wearing the AFOs, normal obstacle avoidance and, obstacle avoidance while wearing the AFOs. The order of these conditions will be presented randomly. You will be asked to walk at a self-selected pace which is comfortable to you. There will be an opportunity to rest between each walking trial. The walking trials will all performed on a level surface. The AFOS are attached to the foot with a single strap around the shin. The plastic mold will run down the back of the leg and under the foot. This will then be placed in your shoe and a strap will be placed around the toe of your shoe and strapped to the top of the AFO to limit the amount of plantarflexion and dorsiflexion that you will be able to create. Ten trials will be completed at each condition, giving a total of 40 walking trials. Also a calibration/standing trial will be completed for each height condition resulting in a total of 44 trials. Videotaping may be done for this study.

The total time commitment for this study will be 2 hours.

Study Risks & Safeguards:

The risks associated with this study are minimal. There is a minimal chance that you may experience slight skin irritation due to the adhesive that is used to attach the IRED markers; however it should dissipate within 24-48 hours. **There is a chance that the electrodes used to measure electrical activity may cause minor skin irritation, this is usually minor and lasts no more than 48 hours.**

There is a minimal risk of falling while wearing the AFOs; however this is similar to the risks of normal everyday walking. Also, a research assistant will also be present to assist you with any difficulties.

If you wish to stop participating at any time, simply state to the research assistant “I no longer wish to participate.” Data collection will stop immediately, and the research assistant will help remove all of the equipment from you.

Confidentiality and Security of information:

All information collected will be kept confidential. All data collected in the study will be kept in the Gait and Posture Lab, within **Burt Matthews Hall**, at the University of Waterloo, **and is only accessible to lab personnel which are involved in the study**. This data will be kept indefinitely to allow further analysis to be completed in the future. At no time during any of this future analysis will your identity be made known as your data will be associated with a participant number.

EXCLUSION CRITERIA OF PARTICIPANT

Due to the nature of the study, if you have had any lower leg (ankle, knee, hip) problems, which have included either surgery, physiotherapy, rehabilitation, or other treatment in the past 6 months, you are not eligible to participate. In signing the consent below you are stating that you have not had any lower leg problems in the past 6 months. In signing the consent below you are stating that you have none of the aforementioned conditions.

CONSENT OF PARTICIPANT

I have read the information presented on the attached sheet about the procedures and risks involved in this study and have received satisfactory answers to my questions related to the study. If you have any further questions you can contact the student investigator, Eoghan Landy, at **e3landy@uwaterloo.ca.** This project has been reviewed and received ethics clearance through the Office of Research Ethics. In the event that I have any questions or concerns about my participation in this study I understand that I may contact Dr. Susan Sykes, the Director Office of Research Ethics, at (519) 888-4567, ext 36005. I understand that I may withdraw from the study at any time by advising the researchers of this decision and that such withdrawal may be done without penalty. With full knowledge of all the information on the previous pages I agree, of my own free will, to be a participant in this study. I have received a signed, original copy of this consent form, for my own records.

_____	_____
Participant's name (print)	Participant's signature
_____	_____
Dated at University of Waterloo	Witness's signature

Appendix B:

Tibialis Anterior Level Walking Conditions

Tibialis Anterior Obstacle Conditions

Gastrocnemius Muscle Level walking Conditions

Gastrocnemius Obstacle Conditions

Rectus Femoris Level Walking Conditions

Rectus Femoris Obstacle Conditions

Biceps Femoris Level Walking Conditions

Biceps Femoris Obstacle Conditions

Gluteus Maximus Level Walking Conditions

Gluteus Maximus Obstacle Conditions

Gluteus Medius Level Walking Conditions

Gluteus Medius Obstacle Conditions

Rectus Abdominus Level Walking Conditions

Rectus Abdominus Obstacle Conditions

Erector Spinae Level Walking Conditions

Erector Spinae Obstacle Conditions