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# Biomechanical Comparison of Lower Limb Unloading Between Common Modalities of Ankle Foot Orthoses

## Abstract

Tibial stress fractures and other lower extremity injuries can be treated using an ankle foot orthosis (AFO). AFOs are popular because they allow the patient to ambulate somewhat naturally while reducing weight bearing on the injured limb. Despite their popularity, it is currently unclear how well AFOs reduce lower extremity weight bearing.

The first objective of this investigation was to examine the ability of three commonly used ankle foot orthoses to reduce weight bearing within the lower limb by comparing the ground reaction force measured from a force platform to the force measured from an insole pressure measurement device inside the AFO. Results indicated that the corset-style AFO was more effective in reducing the load compared to the camwalker and patellar-tendon bearing AFOs.

The second objective was to examine and quantify the kinematic and kinetic changes made in the natural gait pattern of the participants when wearing an AFO. Wearing an AFO alters the geometry and inertial properties of the limb which causes the wearer to alter their natural gait pattern which could lead to additional problems such as low back pain. Results indicated that the camwalker AFO changed gait the most, the patellar-tendon bearing AFO altered gait to some extent, and the corset-style AFO changed gait minimally.

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by

Andrea B. Wanamaker

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Title: Biomechanical comparison of lower limb unloading between common modalities of ankle foot orthoses

Advisor: Bradley S. Davidson, PhD

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## **Abstract**

Tibial stress fractures and other lower extremity injuries can be treated using an ankle foot orthosis (AFO). AFOs are popular because they allow the patient to ambulate somewhat naturally while reducing weight bearing on the injured limb. Despite their popularity, it is currently unclear how well AFOs reduce lower extremity weight bearing.

The first objective of this investigation was to examine the ability of three commonly used ankle foot orthoses to reduce weight bearing within the lower limb by comparing the ground reaction force measured from a force platform to the force measured from an insole pressure measurement device inside the AFO. Results indicated that the corset-style AFO was more effective in reducing the load compared to the camwalker and patellar-tendon bearing AFOs.

The second objective was to examine and quantify the kinematic and kinetic changes made in the natural gait pattern of the participants when wearing an AFO. Wearing an AFO alters the geometry and inertial properties of the limb which causes the wearer to alter their natural gait pattern which could lead to additional problems such as low back pain. Results indicated that the camwalker AFO changed gait the most, the patellar-tendon bearing AFO altered gait to some extent, and the corset-style AFO changed gait minimally.

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# 1. Introduction

Ankle foot orthoses (AFOs) have become a common treatment method for lower-extremity injuries such as stress fractures of the tibia. AFOs have become popular because they allow the patient to ambulate somewhat naturally and reduce weight bearing in the injured limb, thereby aiding in the healing process. While these devices are widely used, it is currently unclear the amount of weight reduction that occurs within the injured limb while wearing an AFO.

The overall goal of this project was to gain insight on how effective three commonly used AFOs are in reducing the weight bearing on the lower limb. In addition to examining the weight reduction capabilities of three common AFOs, this investigation also evaluated how wearing the AFOs altered gait patterns. Gait is considered a symmetrical process and deviations in a gait pattern indicate an underlying deviation. Asymmetric gait patterns could cause low back and hip pain. Because a patient may wear an AFO for long periods of time, evaluating gait asymmetries caused by the AFOs is important.

This thesis document is organized as follows: Chapter 2 presents background information and previous studies looking at the use of AFOs in regards to healing lower limb injuries. Chapter 3 details the investigation done to evaluate the effectiveness of

three common ankle foot orthoses. Chapter 4 presents background information on the gait cycle and previous studies done detailing the effect of gait asymmetries. Chapter 5 details the investigation done to examine the effect three common AFOs have on gait. Chapter 6 provides a summary of the results.

## **2. Tibial Stress Fractures and Treatment Using Ankle Foot Orthoses**

### **2.1 Introduction**

Tibial stress fractures, and other lower extremity injuries such as fractures of the bones within the foot and ankle, and ankle sprains, are common injuries that are typically treated by unloading the lower leg using crutches or a bracing mechanism (Sarmiento and Latta 2006, Sarmiento 1970). Complete disuse is often achieved through crutches. However, crutches are cumbersome and limit mobility. In addition, they can cause discomfort in the upper arms and shoulders and frequently result in patient dissatisfaction and poor compliancy (Stewart and Murray 1997).

Some studies have shown that some weight bearing at the fracture site is beneficial to healing and promotes more osteogenesis compared to complete unweighting and immobilization (Sarmiento and Latta 2006, Sarmiento et al. 1989, Sarmiento et al. 1974, Dehne 1961). Because of this, ankle foot orthoses (AFOs) have become a common treatment for lower limb injuries such as stress fracture. AFOs allow greater patient mobility than crutches, having greater patient compliancy (Sarmiento 1967, McCollough et al. 1978, Sarmiento 1970, Saltzman et al. 1996), and also do not completely reduce the weight on the injured limb, thereby aiding in the healing process (Sarmiento and Latta 2006, Carlson et al. 1991).

While studies have shown that AFOs are an acceptable form a treatment, it is still unclear which AFOs are more effective than others. This review examines the main causes behind stress fractures of the lower extremity, and presents current literature on the effectiveness of various AFOs to reduce weight bearing through the lower limb.

## **2.2 Stress Fractures**

Stress fractures occur due to repetitive loading in the limb. The repetitive loading leads to weakening in the bone and eventually to micro-fractures. If the micro-fractures are not given time to heal then they'll expand to a stress injury and then into a stress fracture (Patel et al. 2011, Fayed et al. 2005). Bones have an endurance limit to how much force they can take before weakening and then buckling under the stress (Kaeding et al. 2005). Stress fractures are one of the top five injuries experienced by runners with 33-55% of the lower extremity fractures located in the tibia (Milner et al. 2006). In regards to stress fractures of the entire body, 23.6% are located in the tibia, making the tibia the bone with the highest instance of stress fractures (Brukner et al. 1996). Part of this is due to the fact that the tibia is a main weight bearing bone in the body.

The tibia is most effective at absorbing and bearing a load when the force applies pure compression through the shaft (Kaeding et al. 2005). However, most runners strike the ground with some amount of heel inversion or eversion, which directs the ground reaction force at an angle to the tibia resulting in a moment arm and additional stress on the tibia (Milner et al. 2006, Kaeding et al. 2005, Ardent 2000). Even with symmetric loading mechanics during running, excessive repetitive loading can lead to stress fractures (Ardent 2000, Patel et al. 2011).

Biology and genetics play a role in the likelihood of a stress fractures occurring. Stress fractures are more prevalent in women (Pester and Smith 1992, Ardent et al. 2003, Patel et al. 2011, Matheson et al. 1987). There are multiple reasons for this including internal factors such as bone density, body size, body composition, flexibility, skeletal alignment, nutrition, and hormone production (Ardent 2000, Milner et al. 2006, Wentz 2011). Women have a much broader range of hormonal changes throughout life than men because of regular menstrual cycles, pregnancy, and menopause. Lack of menses as well as birth control can significantly reduce bone density which increases the likelihood of a fracture (Ardent 2000).

Female athletes in sports such as figure skating and gymnastics, as well as long distance runners and track athletes, are at high risk due to the Female Athlete Triad which states that eating disorders, amenorrhea, and osteoporosis are commonly found in those sports as well as others (Ardent 2000, Milner et al. 2006, Patel et al. 2011). Women in these sports strive for a lean body, and as a result, may over-work their body and under eat, which contributes to weakened bone structure (Ardent 2000).

Another group with particularly high prevalence of stress fractures are active military personnel. This is mostly due to external factors such as the ground surface, external loading, footwear, and physical training parameters, such as carrying a heavy military backpack (Bennell et al. 1999, Patel et al. 2011). During basic training, the intensity of activity coupled with heavily loaded backpacks and uneven ground surfaces, results in increased magnitude and volume of tibial loading (Ardent 2000, Milner et al. 2005, Pester and Smith. 1992). The military also suffers due to the “too much, too soon” conundrum. Recruits are often not prepared for the level of exercise required during basic

training and thus subject their bodies to high levels of stress without proper preparation (Ardent 2000, Patel et al. 2011). Since females in general are more prone to stress fractures, female military recruits are at an even higher risk (Ardent 2000, Lappe et al. 2005).

Biomechanical factors also increase risk of stress fractures in common people. Structurally, a smaller than average tibial cross-sectional area may result in increased fracture risk because the small area indicated a smaller moment of inertia and less resistance to the bending moment (Milgrom et al. 1989). One investigation indicated that excessive tibial varum may cause increased instance of stress fractures (Crossley et al. 1999), but another investigation did not find such a link (Bennell et al. 2004). Therefore, while bone structure does play a role in stress fracture risk, it appears to depend on the person and other confounding factors as well.

Foot strike patterns during running can also alter the risk of stress fractures (Lieberman et al. 2010). Midfoot and forefoot strikers make ground contact anterior to the heel, altering the location of the center of force and reducing the risk of a tibial fracture as the load is not fully applied to the tibial shaft. However, forefoot strikers, which land on their toes and their heels sometimes never touch the ground during the stance phase of running, may increase their chance of a tarsal or metatarsal fracture (Milner et al. 2006). Rear-foot strikers hit with their heel first, which creates the largest vertical impact peak and highest instantaneous loading rates, potentially increasing stress on the tibia (Milner et al. 2006).

In summary, there are a multitude of possible reasons why stress fractures occur in some runners and athletes but not in others. These reasons include internal factors, such as biological makeup, and external factors, such as running style.

### **2.3 Healing Fractures with an Ankle Foot Orthosis**

The majority of stress fractures do not require surgery or casting to heal. Instead, rest and complete unweighting of the injured limb for a sufficient period of time is often enough to allow the break to come to union, or heal (Sarmiento et al. 1989). As mentioned previously, an AFO is a potential device to achieve partial reduction in weight bearing. Sarmiento (1970) noted the success of using a functional below the knee brace in healing not only tibia stress fractures, but tibial fractures including open fractures. In the case of fractures, they waited anywhere between one to four weeks before applying the brace so that the fracture could be properly set and begin healing (Sarmiento 1970, Sarmiento and Latta 2006). McCollough et al. (1978) specifically cited the success of AFOs with children.

AFOs allow for motion in the knee and partial weight bearing, which Sarmiento has concluded can help osteogenesis (Sarmiento and Latta 2006, Sarmiento 1970). While it has been shown that prefabricated braces are acceptable in healing any type of tibial fracture (Sarmiento et al. 1989, Sarmiento and Latta 2006, Sarmiento et al. 1984, Dehne 1961), this may not be the most ideal situation in regards to comfortableness.

Several AFOs are currently used to treat tibial stress fractures with large differences in cost and design (Sarmiento and Latta 2006, Sarmiento and Latta 1995, Carlson et al. 1991, Aita et al. 1998, Tanaka et al. 2000). Three common designs include

the camwalker boot, the corset-style brace, and the Patellar-Tendon Bearing (PTB) cast (Fig. 2.1).



**Figure 2.1** – Three common designs.. a) Camwalker AFO, b) Corset-Style AFO, c) Patellar Tendon Bearing AFO.

The camwalker boot is a prefabricated AFO commonly used to treat many lower-extremity conditions including stress fractures and ankle sprains. It is also sometimes used post ankle and foot surgery as a method for protecting the surgery site. It is the current standard of care when treating tibial stress fractures with an AFO. The camwalker is a bulky device with plastic supports that locks the ankle into place. Velcro straps are used to tighten the boot around limb. It adds height, mass, and width to the injured limb. The sole of the AFO is a rocker type designed to help the patient ambulate with a locked ankle joint. Some designs are equipped with an air bladder which may be pumped to tighten the brace around the ankle joint. Attractive features of the camwalker are that it protects the site of injury and provides an affordable option for most patients since it is prefabricated and comes in many sizes to fit anyone.

Two emerging options for an alternative to the camwalker AFO are the PTB and the corset-style AFO. Both of these AFOs are custom made, making them more expensive than the prefabricated camwalker, and were originally designed for use as a prosthesis but have since been adapted for use as an AFO (Sarmiento and Latta 2006,

Carlson et al. 1991, Tanaka et al. 2000, Saltzman et al. 1996). The PTB and corset-style AFOs apply circumferential pressure on the calf, which is hypothesized to reduce stress on internal tissue and bone, thereby reducing the weight bearing through the shank (Sarmiento et al. 1974, 1984, Latta et al. 1998, Carlson et al. 1991).

The corset-style AFO is a lace-up design that provides circumferential pressure in a similar manner of a typical corset. The AFO has a foot plate which is connected to the leather portion that covers the main muscle belly of the calf. Laces are used to tighten the corset, which increases the circumferential pressure on the soft tissue and tibia and fibula (Carlson et al. 1991, Saltzman et al. 1996).

The PTB is similar to the camwalker in that they both involve tightly strapping a medium against the shank. While the camwalker is mainly comprised of soft padding with plastic supports along the side of the brace more intended for site protection, the PTB is almost completely made of hard plastic that has been molded to the user's shank. The PTB provides bearing on the patellar tendon area, allowing the load during walking to be transferred directly to the knee and effectively surpassing the shank (Tanaka et al. 2000, Sarmiento 1967).

Many studies have found the PTB successful in healing tibial fractures by having a high rate of union and low morbidity (Mollan and Bradley 1979, Suman 1977). However, this design, as noted by Carlson et al. (1991), takes extraordinary customization to effectively bear weight on the patellar tendon. In addition, only a few studies have reported the unloading effect of the PTB though the lower limb (Tanaka 2000, Aita et al. 1998, Svend-Hansen et al. 1979, Mimura et al. 1986, Lehmann et al. 1971). Of these studies, Tanaka et al. (2000) reported weight reduction through the limb

of 30%, which they found disappointing, but they could increase the weight reduction to as high as 90% by increasing the air gap between the shoe and the heel, which helps suspend the limb in the AFO. However, Aita et al. (1998), only found 11% reduction in weight bearing at heel strike. Lehmann et al. (1971), noted that the load bearing of the PTB cast depended greatly on its style and design which may explain the large variability in results from various studies.

Despite the use of the corset-style and PTB AFOs, along with the already popular camwalker, it remains unclear which of these is the most effective at reducing load transfer through the injured limb. The camwalker has been the focus of several biomechanical studies (Sarmiento 1970, Sarmiento et al. 1984, Sarmiento et al. 1989), however, the amount of weight reduction through the lower limb it provides has not been fully investigated. Glod et al. 1996 found that two AFOs similar to the camwalker reduced the pressure against the forefoot by an average of approximately 54%, though they did not report the reduction against any other portion of the foot or against the entire plantar surface. As previously mentioned, the weight reduction occurring in the PTB AFO varies between studies from effectively total weight bearing to as much as 90% reduction in weight bearing (Tanaka et al. 2000, Aita et al. 1998). The corset-style AFO has yet to be investigated in this context for use outside of its original prosthetic use (Carlson et al. 1991), though patients have noted moderate pain relief and comfort in using a corset-style AFO which is promising (Saltzman et al. 1996).

## **2.4 Conclusion**

In conclusion, stress fractures of the tibia are incredibly common and require intervention to heal. Typical intervention includes wearing an ankle foot orthosis. However, as discussed in this review, there are many types of AFOs with various strengths and weaknesses, making it unclear as to which AFO is most effective. Many studies have shown that prefabricated bracing is successful in healing fractures (Sarmiento et al. 1989, McCollough et al. 1978), but opinion appears to be shifting toward using custom designed braces like the PTB and corset-style AFO (Tanaka et al. 2000, Carlson et al. 1991, Saltzman et al. 1996). These custom AFOs may prove to be more successful in treating stress fractures.

# **3. Weight Reduction Capability of Three Common Ankle Foot Orthoses Used to Treat Tibial Stress Fractures**

## **3.1 Introduction**

Lower extremity injuries such as stress fractures require protection around the injured site and typical treatment sometimes includes complete reduction in weight bearing of the injured limb (Sarmiento and Latta 2006, Sarmiento 1970). Complete weight reduction is often achieved through crutches, which can be cumbersome and annoying to use. In addition, crutches can cause discomfort in the upper arms and shoulders and frequently result in patient dissatisfaction and poor compliancy (Stewart and Murray 1997). Some weight bearing on the injured limb may promote higher amounts of osteogenesis and healing than complete weight reduction and immobilization (Sarmiento and Latta 2006, Sarmiento et al. 1989, Sarmiento et al. 1974, Dehne 1961). Because of this, ankle foot orthoses (AFOs) have become a common treatment for lower limb stress fractures because they allow for greater patient mobility than crutches, have greater patient compliancy (Sarmiento 1967, McCollough et al. 1978, Sarmiento 1970, Saltzman et al. 1996), and do not eliminate weight bearing on the injured limb, thereby aiding in the healing process (Sarmiento and Latta 2006, Carlson et al. 1991).

Three primary AFOs, with large differences in cost and design, are currently used to treat tibial stress fractures (Sarmiento and Latta 2006, Carlson et al. 1991, Aita et al. 1998, Tanaka et al. 2000): the camwalker AFO (Fig. 3.1a), the corset-style AFO (Fig. 3.1b), and the Patellar-Tendon Bearing (PTB) AFO (Fig. 3.1c).



**Figure 3.1** – Three bracing modalities that will be compared. a) Camwalker AFO, b) Corset-Style AFO, c) Patellar Tendon Bearing AFO.

The camwalker AFO is a prefabricated AFO commonly used to treat lower-extremity stress fractures, ankle sprain, and also following ankle and foot surgery. It is the current standard of care when treating tibial stress fractures with an AFO. The camwalker AFO is comprised of soft material with two plastic supports up its sides and lock the ankle in place. Velcro straps are used to tighten the AFO around the limb. The sole of the AFO is a rocker type designed to help the patient ambulate with a locked ankle joint. Some designs are equipped with an air bladder which may be pumped to tighten the AFO around the ankle joint. Attractive features of the camwalker are that it protects the site of injury and provides an affordable option for most patients since it is prefabricated. The main disadvantage of the camwalker AFO is that it adds height, mass, and width to the injured limb.

The camwalker AFO has been the focus of several biomechanical studies (Sarmiento 1970, Sarmiento et al. 1984, Sarmiento et al. 1989); however, the amount of weight reduction it provides has not been fully investigated. Glod et al. (1996) found that

two AFOs similar to the camwalker reduced the pressure against the forefoot by approximately 54%, though they did not report reduction in other portions of the foot.

The corset-style AFO is a custom made AFO and was originally designed for use as a prosthesis but has since been adapted for use as an AFO (Carlson et al. 1991, Saltzman et al. 1996). The corset-style AFO is a lace-up design that provides circumferential pressure around the shank, which is hypothesized to reduce stress on internal tissue and bone, thereby reducing the weight bearing through the shank (Sarmiento et a. 1974, 1984, Latta et al. 1998, Carlson et al. 1991). The corset-style AFO transfers the force from the ground to the superior interface of the AFO located on the shank, thereby surpassing the injured site. The laces are used to tighten the corset, which increased the circumferential pressure on the soft tissue and tibia and fibula (Carlson et al. 1991).

The corset-style AFO has yet to be investigated for its effectiveness in weight reduction (Carlson et al. 1991). Patients using the device have noted moderate pain relief and comfort in using a corset-style AFO which is promising (Saltzman et al. 1996). Due to its design, the corset-style AFO is more form fitting than the camwalker AFO. The main disadvantage of the corset-style AFO is it is custom made and can be costly.

The PTB AFO is almost completely made of hard plastic that has been custom molded to the user's shank. Like the corset-style AFO, the PTB AFO has been adapted from its original prosthesis form for use as an AFO. The PTB attempts to provide weight bearing completely on the patellar tendon area, allowing the force during walking to be transferred from the ground to the knee (Tanaka et al. 2000, Sarmiento 1967).

The PTB AFO evolved from the PTB cast, which is of the same design as the AFO except it's made of plaster casting (Sarmiento et al. 1967, Dehne 1961). The PTB cast has shown to be easily applied and successful in treating fractures of the lower limbs (Sarmiento et al. 1967, Suman et al. 1977, Mollan and Bradley 1979). Mimura (1986) reported a weight reduction of 30% in use of the PTB cast. Svend-Hansen (1979) showed by using a load cell under the heel inside the cast that the length of the PTB cast was irrelevant to weight reduction, as in a below the knee cast, the PTB cast, and an above the knee cast were all the same. Aita et al. (1998) and Tanaka et al. (2000) used an insole pressure measurement device from F-SCAN to determine weight reduction in the PTB cast. Aita et al. (1998) found a reduction of 11% during heel strike, and Tanaka et al. (2000) found an average of 30% reduction across stance phase. While these results are for the PTB cast, they can be extended to the AFO as the designs are similar. Sarmiento (1970) adjusted his original plaster design to a plastic AFO and noted that it was just as successful as the PTB cast (Sarmiento et al. 1989, Sarmiento and Latta 2006). The PTB AFO can be costly and it has been noted that it takes extraordinary customization to effectively bear weight on the patellar-tendon (Carlson et al. 1991). Lastly, Lehmann et al. (1971), noted that the load bearing of the PTB depended greatly on its style and design.

Despite the prevalence of camwalker use and increased use of the corset-style and PTB AFOs, it remains unclear which of these is the most effective at weight reduction through the shank. Therefore, the objective of this investigation was to compare the ability of each of the three AFOs to reduce the weight bearing in the lower limb. Amount of weight reduction is quantified using an unloading index based on a comparison of

insole pressure and ground reaction force. The weight reduction analysis included examining the weight reduction throughout the entire stance phase, as well as during the three main periods of stance (loading response, midstance, and terminal stance) as we believed that the weight reduction would not be constant throughout stance phase. We hypothesized that the camwalker AFO would have a lower weight reduction effect when compared to the corset-style and PTB AFO. We also hypothesized that the corset-style AFO would be more effective in weight reduction than the PTB AFO.

### **3.2 Methods**

Fourteen volunteers with no known gait abnormalities participated in a single data collection session at the Interdisciplinary Movement Science Laboratory (IMSL) located on the University of Colorado Denver Anschutz Medical Campus. All participants provided a written, informed consent in accordance with the Colorado Multiple Institutional Review Board (COMIRB) prior to the testing session.

Participants were instrumented with insole pressure measurement devices and performed gait trials across a level surface with embedded force platforms. They performed the gait trials on a 15m walkway while wearing tennis shoes (normal condition), and while wearing an AFO on one limb (braced condition). The relative percentage of weight reduction was calculated as the difference between the insole pressure device, which measures the pressure between the foot and the sole of the brace or shoe, and the external force platform. Difference in weight reduction measured in the braced limb while braced and the braced limb while in a tennis shoe was taken to be the

relative weight reduction. Results were then compared across the three bracing modalities.

For the braced condition, six participants used the camwalker, five used the corset-style AFO, and three used the PTB AFO (Table 3.1). Because the corset-style and PTB AFOs are custom made, the participants in those arms of the investigation used their own brace that they had been previously prescribed for the treatment of tibial stress fractures. At the time of testing they were completely healed, in no pain, and walking normally. Participants who used the camwalker were healthy volunteers and were provided the camwalker at the time of testing.

**Table 3.1:** Participant demographics

AFO Used	Total Participants	Female	Male	Age±SD (yrs)	Height±SD (cm)	Weight±SD (lbs)
Camwalker	6	5	1	22.7±2.1	173.0±11.5	166.2±37.7
Corset-Style	5	4	1	38.4±8.9	168.2±9.4	165.0±57.4
PTB	4	3	1	38.3±7.1	166.6±8.3	130.7±25.0
Total	15	12	3	33.1±9.0	169.3±3.3	154.0±20.1

The participants performed walking trials over a level surface at an enforced speed of 1.5 m/s. They performed the walking trials under two conditions: normal walking in tennis shoes on both feet, and braced walking with the brace on one limb and tennis shoes on the contralateral limb. Embedded in the walking surface were two force platforms measuring at 2000 Hz (Bertec, Columbus, OH).

Participants were instrumented with F-SCAN's system of insole pressure measurement devices (Tekscan, Boston, MA) recording at 50 Hz. The insole devices measured the pressure against the sensor which quantified the force against the foot (Tanaka et al. 2000). Because they are thin and flexible, the devices can be cut to fit any size of shoe (Razak et al. 2012). Two sets of insoles were cut for each participant and

used throughout the entire testing session either inside the tennis shoe or brace. A new set of insoles were used when the participant switched between normal and braced walking. Each insole was calibrated to the participant's weight according to F-SCAN system's directions.

After being instrumented with the insole pressure measurement devices, participants were given time to become familiar with the testing equipment and environment. In addition, they were given practice sessions until they could walk consistently at the enforced speed during normal and braced walking. Once familiarized, participants were instructed to walk at the enforced speed while data was collected on the force platforms and insole devices simultaneously. They were not informed they were walking over force platforms to prevent targeting. Data were collected until three successful, clean, foot strikes were recorded against a force platform for each foot during each bracing condition within 5% of the target speed.

In post-processing, data from the force platforms and insole devices were filtered using a 4<sup>th</sup> order, zero lag, Butterworth low-pass filter at 12 Hz. The stance phase of each trial was isolated. The reduction in weight bearing was assessed by measuring the resultant force calculated by the insole and comparing it to the ground reaction force calculated by the force platform using the following Percent Weight Reduction (PWR) formula:

$$\text{PWR} = \left(1 - \frac{F_{IPM}}{F_{FP}}\right) * 100\%$$

where  $F_{IPM}$  is the force measured from the insole and  $F_{FP}$  is the force measured from the force platform. A PWR of zero indicates no weight reduction occurred through the shank.

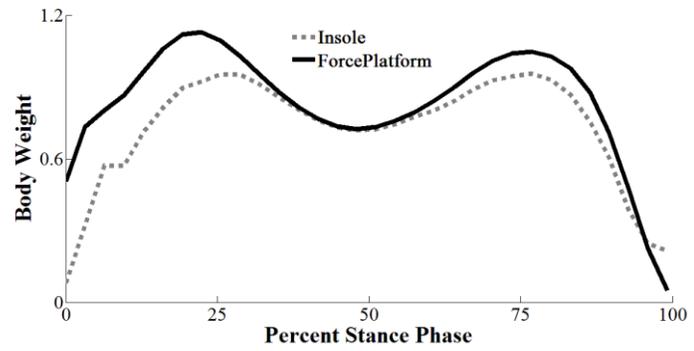
PWR was calculated for the braced limb during braced walking and for the same limb during normal walking. A relative PWR was obtained by subtracting PWR in the braced limb during the normal condition from PWR in the braced condition. Mean PWR was found for the entire stance phase for each trials, as well as the mean PWR for each of the three main periods of stance phase: loading response (2-27% of stance phase), midstance (28-62%), and terminal stance (63-95%). In order to estimate where each AFO reduced the weight in the limb with a higher resolution than examining the mean results from each period, a 95% confidence interval was obtained from the ensemble averages for each group over the entire stance phase.

The mean PWR values were compiled from the three trials for each subject and averaged together, producing six results for the camwalker group, five for the corset-style group, and three for the PTB group. The mean PWR results were then compared between groups using an unpaired, two-tailed, Student's *t*-tests, with an alpha level of 0.05. Effect sizes were also calculated.

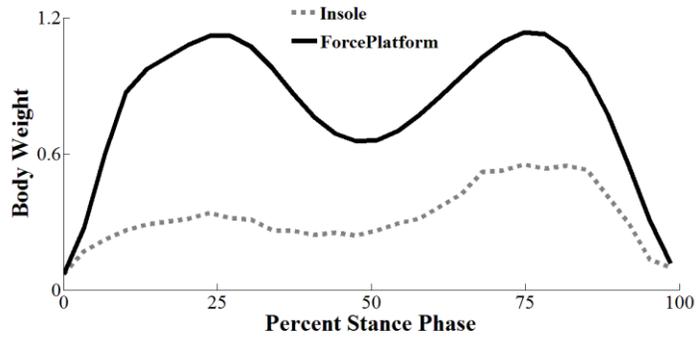
### **3.3 Results**

Three representative plots depicting the comparison of the ground reaction force ( $F_{FP}$ ) and the force from the insole measurement devices ( $F_{IPM}$ ) from a single subject within each group are shown in Figure 3.2. Small differences exist in the camwalker brace (Fig. 3.2a) and in the PTB (Fig. 3.2b). Large differences between the two forces are present in the corset-style AFO (Fig. 3.2c). Ensemble group average of the PWR results with 95% confidence bounds are shown in Figure 3.3.

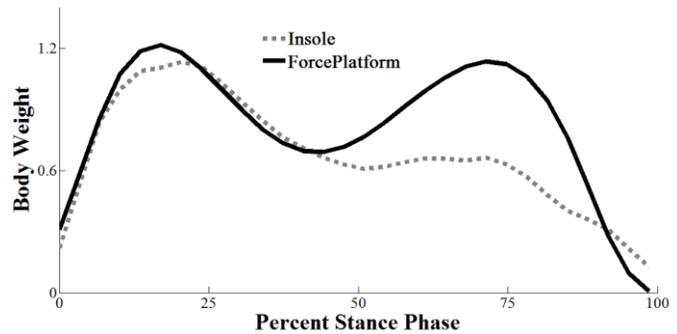
(a) Camwalker participant



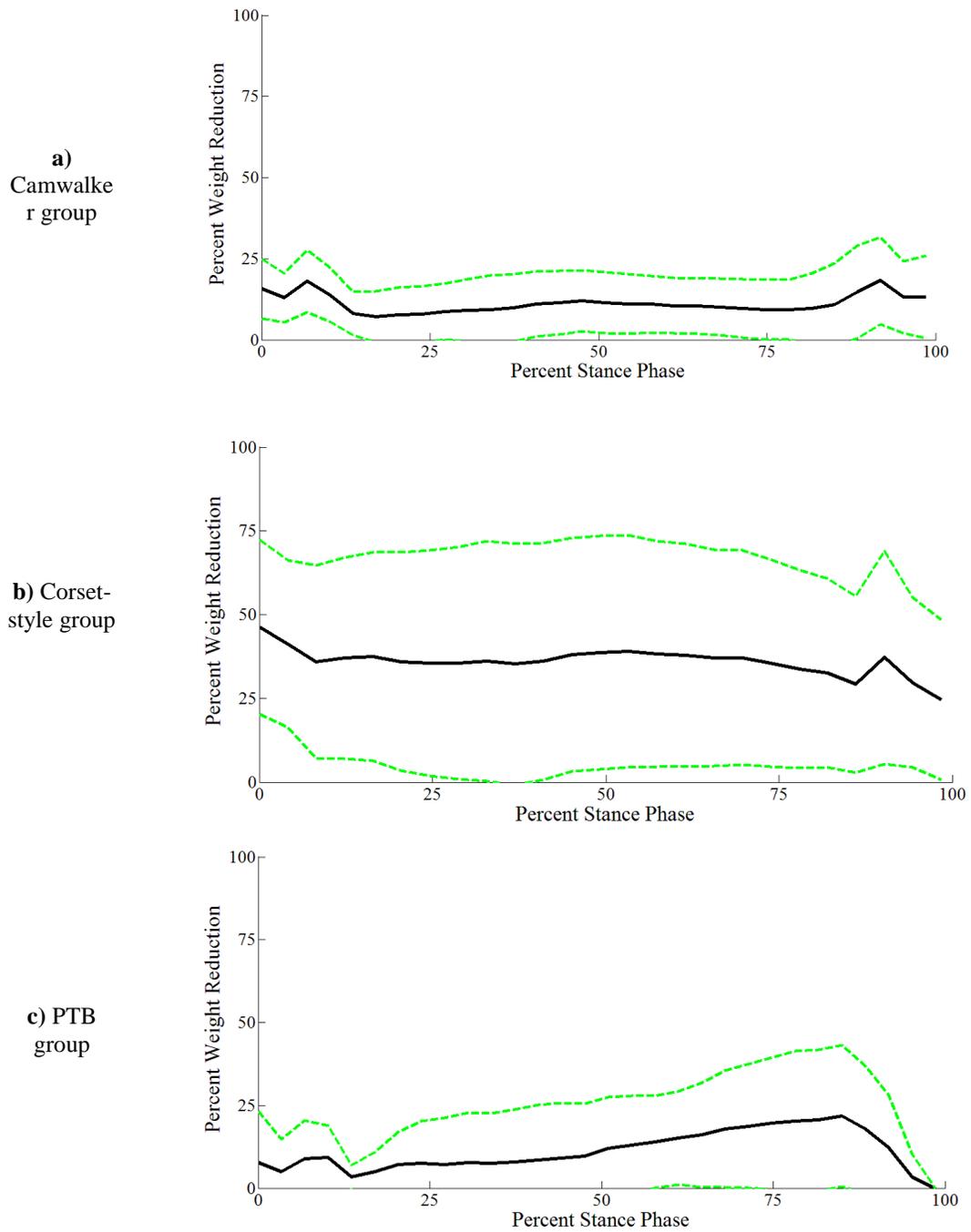
(b) Corset-Style participant



(c) PTB participant



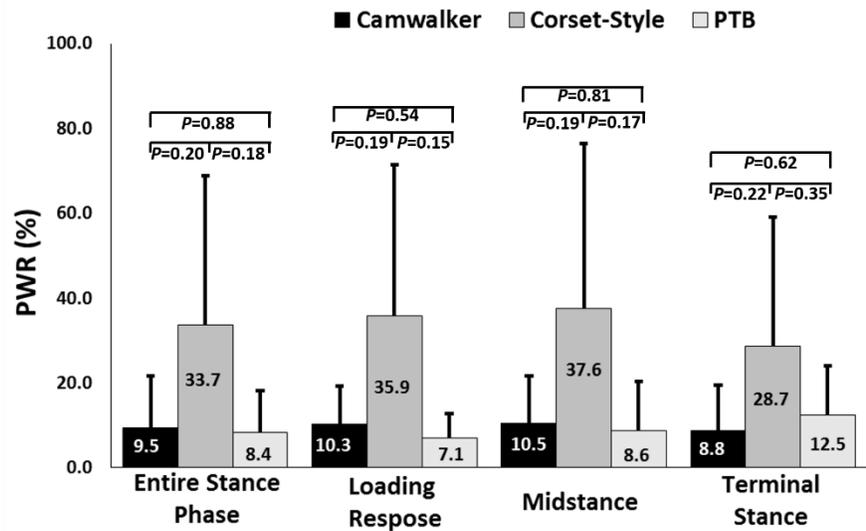
**Figure 3.2:** Comparison of  $F_{FP}$  and  $F_{IPM}$  from one representative subject for each bracing modality for the braced limb over stance phase. Force has been normalized to body weight.



**Figure 3.3:** Ensemble group averages of PWR for entire stance phase with 95% confidence bounds **a)** camwalker group, **b)** corset-style group, and **c)** PTB group.

No significant differences were found between any of the bracing modalities (Fig. 3.4). Effect sizes are listed in Table 3.2.

**Figure 3.4:** Ensemble group means (SD) during each stance phase period with *P* values.



**Table 3.2:** Effect Sizes across groups

	Camwalker and Corset-Style	Camwalker and PTB	Corset-Style and PTB
Stance Phase	0.42	0.05	0.44
Loading	0.44	0.21	0.49
Midstance	0.43	0.08	0.45
Terminal	0.40	0.16	0.33

### 3.4 Discussion

In this investigation we quantified the amount of weight reduction that occurs in the camwalker, corset-style, and PTB AFOs by using a Percent Weight Reduction calculation. There were no statistical differences in mean weight reduction between the bracing modes. While the mean value results for each period of the stance phase were not significant, the more detailed results seen in the 95% confidence interval tests do indicate weight reduction occurring.

The 95% confidence interval plots (Fig. 3.3) show where the ensemble averages from the bracing groups are significantly different from zero. In the camwalker group, there was significant weight reduction occurring during 0-17%, 27-30%, 43-80%, and 90-

100% of the stance phase. These results are surprising as we believed there would be no reduction in weight in the camwalker AFO. However, the confidence interval indicates that the camwalker provided weight reduction during loading response, some of midstance and terminal stance, and during toe-off. While the confidence interval shows a difference from zero, or no weight reduction occurring, during these portions, it should be noted that the amount of weight reduction, approximately 10-15%, is disappointing.

The corset-style AFO almost provided weight reduction during the entire stance phase. The confidence interval is greater than zero for 94% of the gait cycle. The portion in which the confidence bounds are below zero are in 37-43% of the gait cycle. This occurs during midstance. During midstance the foot is flat to the ground and perhaps during this period of the stance phase the corset-style brace is not as effective as the limb could be falling down in the AFO as during this portion of stance all of the weight bearing is straight through the tibia. The corset-style AFO may not be properly suspending the shank during this portion of stance.

The PTB AFO's confidence interval is only above zero for a total of 16% of the stance phase. The first portion where it is above zero is from 60-73% and the second portion is from 84-87%. These results are surprising as we expected the PTB AFO to be more effective in reducing the weight bearing through the shank than the camwalker AFO.

The mean level of weight reduction through the shank for the PTB AFO, approximately 9.5%, is similar to that presented in previous work by Aita et al. (1998) who reported an average of 11% in weight reduction during heel strike. The results from the camwalker group cannot be properly compared to those presented in the work of Glod

et al (1996) because they measured only the forefoot pressure and we measured the entire plantar surface. To our knowledge, this is the first examination of the weight reduction capabilities of a corset-style AFO.

The camwalker and PTB AFOs had similar values of weight reduction and were lower than those of the corset-style AFO. The camwalker AFO had no visible means to create weight reduction in the shank. It does not use circumferential pressure like the corset-style brace, and it has no visible means like the PTB does in transferring the force from the ground to elsewhere on the shank. Thus, its low weight reduction values make sense.

The low weight reduction found in the PTB was unexpected, as its method of load transfer to the patellar-tendon had been previously documented as successful (Tanaka et al. 2000). The higher levels of weight reduction in the corset-style AFO indicate that it is probably successful in transferring the force from the ground to the soft tissue in the calf by circumferential pressure.

The main limitation to this study is that weight reduction in the lower extremities cannot be directly measured. The method used in this investigation is only a surrogate measurement. Second, the insole devices used only measure the force normal to the sensor and thus we cannot measure the shear components, which causes the sensor to report a lower force value than the true force occurring. However, we feel that by calculating a relative PWR we have compensated for this error and produced a conservative measurement. Lastly, we had a low sample size for each group and the participants could not be tested in every bracing modality. Because the corset-style and PTB AFOs are custom made, we had a limited population from which volunteers could

be drawn and thus we could not fully control for population demographics. There are large variances in height, weight, and age across the three bracing groups which may affect our results. In addition, the participants were healthy volunteers and their use of an AFO may not reflect the actual use of the device when worn during fracture management.

### **3.5 Conclusion**

In conclusion, the corset-style AFO reduces the weight bearing through the lower limb by an average of 33.7% throughout stance phase which is comparatively better than the 9.5% and 10.2% weight reduction in the camwalker and PTB AFOs, respectively. This pilot investigation was developed to examine the effectiveness of the three braces and to determine if a clinical trial was needed. From our results, it is evident that a clinical trial should be performed so that a large population may be tested and other types of AFOs may be examined in addition to the three presented.

## **4. Observational and Quantitative Analysis of Gait Asymmetry**

The most common method of human locomotion is to use a bipedal gait cycle. While gait seems coordinated and effortless, it is a very complicated processes involving the collaboration of two multi-jointed and segmented lower limbs (Perry and Burnfield, 2010). Normal gait is assumed to be a repeatable and generally symmetrical process, as this is an effective and efficient way to move (Carollo and Matthews, 2009).

Gait consists of two main periods, the stance and swing periods. During stance the limb accepts the weight of the body and acts as support and the contralateral limb swings forward. Then the tasks switch. The stance limb becomes the swing limb and the swing limb is now the supporting limb. Gait is also interesting because it contains a double support portion, where both limbs are on the ground at the same time as they switch tasks (Perry and Burnfield, 2010, Carollo and Matthews, 2009).

The stance period makes up approximately 60% of the gait cycle. Contained in this period are five main phases that the limb goes through. First is the initial contact with the ground, or heel strike. As the heel strikes the ground the body shifts its weight towards that limb and the limb acts as a shock absorber. Then the limb begins to accept

the weight of the body during the loading response. At approximately 12% of gait cycle, the foot is now flat on the ground and the contralateral limb has started its swing phase. This is the first portion of the gait cycle with single limb support. As the contralateral limb swings through, the support limb shifts forward, lifting the heel as terminal stance begins at 30% of the gait cycle. Then the body continues to move forward, the contralateral limb makes heel contact, and the support limb now can lift its toes off the ground and begin the swing period.

At 60% of the gait cycle, the limb loses contact with the ground and begins swing phase. The contralateral limb is now the single support limb. As the limb swings through, its main goal is to clear the ground and advance the body forward. At 75% of the gait cycle, the initial swing period transitions into the mid-swing. This is seen by the swinging foot passing by the stance foot as it advances. Then for the last 13% of the gait cycle, the swinging limb straightens, the tibia becomes vertical with respect to the ground, and the limb prepares for heel strike (Perry and Burnfield, 2010, Perry 1992).

Throughout the gait cycle many critical events occur. These events are considered critical as without them, or without them being completed fully, gait becomes difficult. In all there are approximately thirteen critical events needed during gait to make it run smoothly. These events include joints reaching certain flexion and extension angles, muscles activating at certain times, positioning the lower limb so that the heel strikes first during the start of stance, and controlling not just the lower limbs but the torso as well. For example, one of the critical events is dorsiflexion of the ankle and flexion of the knee in order for the foot to clear the ground during mid-swing. If the ankle does not dorsiflex enough, the foot will point downward and the toes will hit the ground, possibly causing

the person to trip. If the knee does not flex enough and doesn't lift the limb high enough leading up to mid-swing, then the foot, even if it is properly functioning, will not clear the ground (Perry and Burnfield, 2010). Other critical events include the controlled movement of the hip muscles throughout the gait cycle. The hips are very important during the gait cycle as not only do they assist in propelling the body forward, but they control and help balance the torso with the lower limbs.

When the critical events do not occur, gait becomes difficult and even impossible to perform. However, clinicians can use observational analyses to diagnosis pathologies in gait (Carollo and Matthews, 2009). The clinicians use the absence or modification of any of the thirteen critical events to narrow in on what the problem is. For example, excessive knee flexion during the stance period, which can cause the person to crouch as they walk and cause excessive loading of the quadriceps, can be treated by lengthening the hamstrings (Delp et al. 1998). Typical knee flexion during the loading response of stance peaks around 15-20 degrees, however, someone with excessive knee flexion could be peaking around 40-50 degrees (David 2000). The excessive knee flexion not only alters the normal gait pattern of the knee, but forces the other joints to compensate as well, would could lead to other issues in those joints, such as tendon strain from atypical use (Perry and Burnsfield, 2010).

While observational gait analysis is a powerful clinical tool, it cannot give the details that a kinematic and kinetic analysis can. Biomechanical measurement of kinematics, kinetics, and muscle activity now plays a large role when diagnosing the cause of gait abnormalities (Yang et al. 1984). Asymmetries can be detected through kinematic and kinetic gait analyses as well as spatial-temporal variables such as stride

and step length. These gait measurement are commonly used to classify normal and abnormal gait (Andriacchi et al. 1977, Allet et al. 2011). Because these variables are objective and quantifiable, they can reflect difference between limbs that are not obvious to an observer (Herzog et al. 1989). Since gait is considered a symmetric process, the analysis of gait symmetry can lead to the diagnosis of gait pathologies.

Herzog et al. (1989) introduced a quantitative metric of gait asymmetry called the Absolute Symmetry Index (ASI). In doing so, they noted that if gait symmetry is “the perfect agreement of the external kinetics and kinematics of the left and right legs...[then] this definition immediately implies that human gait is not symmetrical” (Herzog et al. 1989). The ASI provides a direct comparison between the left and right sides using any quantifiable measurement, such as joint angles, joint torques, and muscle activation times. The amount of asymmetry is assessed by the numerical distance away from zero, which indicates perfect symmetry. For example, if the left knee is flexing during the loading response at 15 degrees, and the right knee is doing the same during its loading response, then the limbs are in symmetry with respect to that kinematic variable. However, if the right knee is instead flexing at 25 degrees, then the limbs are not symmetrical and there is an issue somewhere in the lower limbs causing the asymmetry. While the ASI provides a simple way of analyzing gait, it is sometimes unclear what quantitative measures are best to use in the ASI when comparing normal gait to pathological gait (Carollo and Matthews, 2009).

Several investigations examined the effect of leg length discrepancies on gait asymmetry (Perttunen et al. 2004, Delacerda and Wikoff 1982, Liu et al. 1998). Perttunen et al. (2004), applied the ASI to compare ground reaction force results between the long

and short limb. They found that the duration of the stance period during level natural walking speed was longer on the long limb. During a faster walking speed, the long limb's duration of stance period increased, creating a definitely asymmetry with the shorter limb. They also found that the average plantar pressure was greater on the long limb than the short limb (Perttunen et al. 2004). Delacerda and Wikoff (1982), examined a 30 year old female with a leg length difference of 1.25 inches. They investigated the effect of the leg length discrepancy on the duration of the phases of gait and if using a heel lift could correct the deviations. They found that an asymmetry existed in the duration of the gait phase between limbs and that is could be reduced to no asymmetry when a lift was used (Delacerda and Wikoff 1982). Liu et al. (1998), used the ASI to determine the acceptable range of leg length inequality in which gait asymmetry was small. They found a difference of 2.23 cm between limbs to be acceptable, though they noted that large standard deviations in the ASI undermine the accuracy of their results (Liu et al. 1998).

Low back pain and gait asymmetry often coexist, indicating an etiological link between the two. Childs et al. (2003), found that individuals with lower back pain exhibited weight-bearing asymmetry. In addition, two investigations clearly link lower extremity asymmetry in amputees with the prevalence of low back pain (Kulkarni et al. 2005, Baum et al. 2008). Although many factors may cause asymmetries in amputees, such as knee pain, an ill-fitted prosthetic, a degenerative muscular disorder, or leg length difference, it appears that regardless of the cause, the asymmetry can lead to problems elsewhere in the body (Baum et al. 2008).

In conclusion, gait asymmetries can cause wear and tear on the human body. Observational analysis can be used to identify gait asymmetries but object measures should be used to quantify the level of asymmetry present as seen when using the ASI. It is clear that gait abnormalities and asymmetries can be harmful to the body; however, no studies have been conducted to examine the effect that treatment methods, specifically ankle foot orthoses, on lower extremity injuries, like stress fractures, may have on the rest of the body. If AFOs or other devices affect the natural gait cycle of the user, then they may also create back pain and tendon strain similar to individuals with amputations and pathological gait issues.

# **5. Biomechanical Analysis of Three Ankle Foot Orthoses on Gait Symmetry**

## **5.1 Introduction**

Lower extremity injuries, such as tibial stress fractures, are common injuries that are typically treated by using an ankle foot orthosis (AFO) (Sarmiento et al. 1989, Sarmiento and Latta 2006). AFOs are used because they allow the patient to ambulate somewhat normally while providing protection around the injured limb. However, wearing an AFO alters the geometry and inertial properties of the lower limb and could cause an abnormal gait pattern.

Normal gait is assumed to be repeatable and a generally symmetric process and deviations from symmetry can be used to examine an underlying pathology (Carollo and Matthews 2009). It is possible for long-term gait asymmetries to result in hip or low back pain as seen with long-term unilateral prosthetic use (Kulkarni et al. 2005, Baum et al. 2008). Asymmetries can be detected through kinematic and kinetic gait analyses as well as spatial-temporal variables such as stride and step length. These gait measurements are commonly used to classify normal and abnormal gait (Andriacchi et al. 1977, Allet et al. 2011).

Three primary AFOs, with large differences in cost and design, are currently used to treat tibial stress fractures (Sarmiento and Latta 2006, Carlson et al. 1991, Tanaka et al.

2000): the camwalker AFO (Fig. 5.1a), the corset-style AFO (Fig. 5.1b), and the Patellar-Tendon Bearing (PTB) AFO (Fig. 5.1c).



**Figure 5.1** – Three bracing modalities that will be compared. a) Camwalker AFO, b) Corset-Style AFO, c) Patellar Tendon Bearing AFO.

The camwalker AFO is a prefabricated AFO commonly used to treat lower-extremity stress fractures, ankle sprain, and also following ankle and foot surgery. It is the current standard of care when treating tibial stress fractures with an AFO. The camwalker AFO is comprised of soft material with two plastic supports up its sides and lock the ankle in place. Velcro straps are used to tighten the AFO around the limb. The sole of the AFO is a rocker type designed to help the patient ambulate with a locked ankle joint. Attractive features of the camwalker are that it protects the site of injury and provides an affordable option for most patients since it is prefabricated. The main disadvantage of the camwalker AFO is that it adds height, mass, and width to the injured limb which could cause the patient to ambulate awkwardly.

The corset-style AFO is a custom made AFO and was originally designed for use as a prosthesis but has since been adapted for use as an AFO (Carlson et al. 1991, Saltzman et al. 1996). The corset-style AFO is a lace-up design that provides circumferential pressure around the shank, which is hypothesized to reduce stress on internal tissue and bone, thereby reducing the weight bearing through the shank

(Sarmiento et al. 1974, 1981, 1984, Latta et al. 1998, Carlson et al. 1991). While more expensive than the camwalker AFO, the corset-style AFO has been used because patients note that it reduces pain in the injured limb and is non-obtrusive and easier to wear (Saltzman et al. 1996). Its low profile may create less deviation in the patient's normal gait pattern.

The PTB AFO is almost completely made of hard plastic that has been custom molded to the shank of the patient. Like the corset-style AFO, the PTB AFO has been adapted from its original prosthesis form for use as an AFO (Carlson et al. 1991). Because the PTB is molded to the shank, it has a low profile and may create less asymmetrical gait. However, some designs have a locked ankle joint that may create gait deviations from normal walking.

All AFOs alter the geometry of the lower limb in some manner; however, it is currently unclear how that may affect gait kinematic and kinetics. Observational gait analysis of the AFOs indicate the existence of gait asymmetry, which may result in similarly painful conditions seen in unilateral amputees when used long term (Kulkarni et al. 2005, Baum et al. 2008).

The objective of this investigation was to quantify the lower body kinematic and kinetic gait asymmetry during unilateral AFO use. We used two methods to examine the asymmetry. The first was comparing peak kinematic and kinetic variables across limbs during braced and normal walking conditions. Second, we calculated an Absolute Symmetry Index (ASI) examining the level of asymmetry between bracing conditions. We hypothesized that asymmetry would be present in each of the three AFOs, and that

the camwalker, due to its larger size and mass, would have the highest number of asymmetrical kinematic and kinetic variables.

## **5.2 Methods**

Fourteen volunteers with no known gait abnormalities participated in a single data collection session at the Interdisciplinary Movement Science Laboratory (IMSL) located on the University of Colorado Denver Anschutz Medical Campus. All participants provided a written informed consent in accordance with the Colorado Multiple Institutional Review Board (COMIRB) prior to testing.

The participants performed walking trials over a level surface with embedded force platforms while instrumented with reflective markers for motion capture. They performed the gait trials on a 15m walkway while wearing tennis shoes (normal condition), and while wearing an AFO on one limb (braced condition). Three-dimensional joint kinematics and kinetics were then calculated for each subject and compared.

For the braced condition, six participants used the camwalker, five used the corset-style AFO, and three used the PTB AFO (Table 5.1). Because the corset-style and PTB AFOs are custom made, the participants in those arms of the investigation used their own brace that they had been previously prescribed for the treatment of tibial stress fractures. At the time of testing they were completely healed, in no pain, and walking normally. Participants who used the camwalker AFO were healthy volunteers and were provide the camwalker at the time of testing.

**Table 5.1:** Participant demographics

AFO Used	Total Participants	Female	Male	Age±SD (yrs)	Height±SD (cm)	Weight±SD (lbs)
Camwalker	6	5	1	22.7±2.1	173.0±11.5	166.2±37.7
Corset-Style	5	4	1	38.4±8.9	168.2±9.4	165.0±57.4
PTB	3	3	0	41.7±2.3	162.7±3.1	119.0±10.8
Total	14	12	2	34.2±10.2	168.0±5.2	150.1±26.9

The participants performed walking trials over a level surface at an enforced speed of 1.5 m/s. They performed the walking trials under two conditions: normal walking in tennis shoes on both feet, and braced walking with the brace on one limb and tennis shoe on the contralateral limb. Embedded in the walking surface were two force platforms measuring at 2000 Hz (Bertec, Columbus, OH). Participants were instrumented with 36 reflective markers on the lower limbs (Fig. 5.2) and motion capture data was collected at 100 Hz (Vicon, Centennial, CO).



**Figure 5.2:** Camwalker participant instrumented with 36 reflective markers for kinematic and kinetic analysis.

After being instrumented with the markers, participants were given approximately five to ten minutes to become familiar with the testing environment. In addition, they were given practice sessions until they could walk consistently at 1.5 m/s during normal and braced walking. Once familiarized, participants were instructed to walk at the

enforced speed while data was collected on the force platforms and marker data simultaneously. They were not informed they were walking over force platforms to prevent targeting. Data were collected until three successful, lean, foot strikes were recorded against a force platform for each foot during each condition within 5% of the target speed.

In post-processing, data from the force platforms were filtered using a 4<sup>th</sup> order, zero lag, Butterworth low-pass filter at 12 Hz. Marker data was filtered using a 4<sup>th</sup> order, zero lag, Butterworth low-pass filter at 7 Hz. The stride corresponding to the clean foot strike was isolated and three-dimensional kinematics and kinetics variables were calculated (Table 5.2, 5.3). The focus of the kinematic and kinetic analysis was to examine the sagittal plane movement of the hip and knee as we believed this would be where the most deviations from normal gait would occur.

**Table 5.2:** Kinematic Dependent Variables. Variables calculated from each stride for each bracing condition and limb.

<b>Stance Phase</b>	<b>Swing Phase</b>
Peak Hip Extension	Peak Hip Flexion
Peak Hip Adduction	Peak Hip Abduction
Peak Mediolateral Pelvic Tilt	Peak Mediolateral Pelvic Tilt
Peak Knee Flexion	Peak Knee Flexion

**Table 5.3:** Kinetic Dependent Variables. Variables calculated from each stride for each bracing condition and limb

Peak Ankle Moment
Peak Knee Extensor Moment
Peak Knee Flexor Moment
Peak Hip Extensor Moment
Peak Hip Flexor Moment
Peak Hip Abduction Moment

This was done for each limb during each condition: braced and unbraced limb during braced walking, and braced and unbraced limb during normal walking. Peak values of all the variables were taken from each of the three trials to create an average for each subject.

Paired Student's *t*-tests at an alpha level of 0.05 were used to compare across limbs within each bracing modality, as in the braced limb during the braced condition was compared to that same limb during normal walking for the subjects within the camwalker group. This was then done for the corset-style and PTB groups. An Absolute Symmetry Index (ASI) was found between conditions for each subject (Equation 5.1), as in symmetry was calculated for the normal condition and compared to the symmetry result for the braced condition quantitatively. An ASI of 0% indicates perfect symmetry (Herzog 1989).

$$ASI = \left| \frac{X_B - X_U}{0.5(X_B + X_U)} \right| * 100\% \quad \begin{array}{l} X_B = \text{value from braced limb} \\ X_U = \text{value from unbraced limb} \end{array} \quad (5.1)$$

### 5.3 Results

Kinematic results for the camwalker, corset-style, and PTB groups are in Tables 5.4-5.6. The tables show the peak kinematic variable comparison between limbs across bracing condition, as well as the ASI results for each condition. Kinetic results for the camwalker, corset-style, and PTB groups are in Tables 5.7-5.9. The kinetic tables are formatted identically to the kinematic tables.

**Table 5.4:** Kinematic Variables Camwalker Group

Mean (SD) [degrees] values for each dependent variable and results of independent *t*-tests comparing the limb across conditions. ASI values shown underneath *P* values indicate level of asymmetry in the braced condition or normal condition.

Dependent Variable	Braced Limb			Unbraced Limb			ASI	
	Braced Condition	Normal Condition	<i>P</i> value	Braced Condition	Normal Condition	<i>P</i> value	Braced Condition	Normal Condition
<u>Knee Flexion</u>								
Flexion (Stance)	15.7(5.3)	15.6(4.5)	<i>P</i> =0.48	10.6(2.5)	16.9(5.8)	<b><i>P</i>=0.01</b>	35.6(21.4)	11.6(10.6)
Flexion (Swing)	58.5(8.2)	56.2(4.5)	<i>P</i> =0.28	57.3(5.1)	59.5(7.2)	<i>P</i> =0.16	7.3(6.9)	6.8(2.0)
<u>Hip Ext/Flex</u>								
Flexion	43.2(9.9)	40.4(7.2)	<b><i>P</i>=0.05</b>	41.6(8.0)	41.8(8.0)	<i>P</i> =0.47	7.9(4.9)	4.4(4.3)
Extension	6.8(8.0)	6.2(6.8)	<i>P</i> =0.33	8.3(6.3)	7.1(6.5)	<i>P</i> =0.13	33.1(42.9)	14.5(20.1)
<u>Hip Ad/abduction</u>								
Adduction	6.7(1.9)	5.4(1.9)	<b><i>P</i>=0.02</b>	6.8(1.3)	6.4(1.3)	<i>P</i> =0.19	8.9(5.6)	16.2(14.6)
Abduction	-1.7(1.5)	1.2(1.2)	<b><i>P</i>&lt;0.001</b>	-0.1(1.2)	0.1(1.2)	<i>P</i> =0.30	474.7(438.7)	181.7(200.1)
<u>Pelvic Tilt</u>								
Tilt (Stance)	5.9(0.8)	4.8(1.9)	<i>P</i> =.09	5.6(1.0)	5.3(1.3)	<i>P</i> =0.32	7.9(7.8)	22.6(35.8)
Tilt (Swing)	4.4(2.2)	6.0(1.2)	<i>P</i> =0.11	5.7(1.5)	5.7(1.1)	<i>P</i> =0.48	34.3(43.0)	16.0(14.1)

**Table 5.5:** Kinematic Variables Corset-Style Group

Mean (SD) [degrees] values for each dependent variable and results of independent *t*-tests comparing the limb across conditions. ASI values shown underneath *P* values indicate level of asymmetry in the braced condition or normal condition.

Dependent Variable	Braced Limb			Unbraced Limb			ASI	
	Braced Condition	Normal Condition	<i>P</i> value	Braced Condition	Normal Condition	<i>P</i> value	Braced Condition	Normal Condition
<u>Knee Flexion</u>								
Flexion (Stance)	20.5(17.5)	17.5(9.1)	<i>P</i> =0.28	14.7(13.4)	13.4(4.5)	<i>P</i> =0.18	31.3(26.7)	25.5(16.3)
Flexion (Swing)	68.6(5.0)	63.7(6.2)	<i>P</i> =0.09	59.6(2.0)	59.4(2.8)	<i>P</i> =0.41	13.9(5.9)	6.7(7.5)
<u>Hip Ext/Flex</u>								
Flexion	36.3(7.6)	33.4(7.8)	<i>P</i> =0.10	33.6(5.8)	32.5(5.4)	<i>P</i> =0.35	11.0(9.9)	6.1(4.9)
Extension	13.1(5.4)	14.8(4.9)	<i>P</i> =0.20	13.3(3.6)	14.9(2.6)	<i>P</i> =0.22	14.2(4.9)	12.6(5.7)
<u>Hip Ad/abduction</u>								
Adduction	5.4(1.4)	5.9(1.4)	<i>P</i> =0.14	6.2(1.5)	5.9(1.7)	<i>P</i> =0.15	22.0(11.2)	18.5(11.3)
Abduction	0.9(2.4)	0.8(1.6)	<i>P</i> =0.47	2.2(1.9)	0.9(1.3)	<i>P</i> =0.07	246.0(223.4)	143.4(99.0)
<u>Pelvic Tilt</u>								
Tilt (Stance)	2.2(3.9)	1.9(3.3)	<i>P</i> =0.24	6.5(2.5)	5.3(2.1)	<i>P</i> =0.16	153.8(159.0)	143.4(151.8)
Tilt (Swing)	6.2(3.1)	5.6(1.9)	<i>P</i> =0.31	2.1(4.0)	2.0(3.6)	<i>P</i> =0.45	170.0(153.6)	145.1(154.4)

**Table 5.6:** Kinematic Variables PTB Group

Mean (SD) [degrees] values for each dependent variable and results of independent *t*-tests comparing the limb across conditions. ASI values shown underneath *P* values indicate level of asymmetry in the braced condition or normal condition.

Dependent Variable	Braced Limb			Unbraced Limb			ASI	
	Braced Condition	Normal Condition	<i>P</i> value	Braced Condition	Normal Condition	<i>P</i> value	Braced Condition	Normal Condition
<u>Knee Flexion</u>								
Flexion (Stance)	14.5(4.9)	12.1(6.6)	<b><i>P</i>=0.04</b>	17.6(5.0)	14.2(8.7)	<i>P</i> =0.16	20.4(24.9)	13.5(6.2)
Flexion (Swing)	60.7(4.0)	59.6(5.5)	<i>P</i> =0.26	61.7(3.3)	61.1(5.6)	<i>P</i> =0.38	5.4(4.0)	3.5(1.4)
<u>Hip Ext/Flex</u>								
Flexion	33.0(6.1)	31.4(7.3)	<i>P</i> =0.30	33.6(4.9)	31.0(7.7)	<i>P</i> =0.15	5.9(6.1)	2.5(0.6)
Extension	13.3(4.5)	14.0(5.9)	<i>P</i> =0.31	12.7(5.3)	15.1(7.7)	<i>P</i> =0.13	25.4(15.0)	16.6(9.9)
<u>Hip Ad/abduction</u>								
Adduction	5.3(3.7)	5.0(2.3)	<i>P</i> =0.39	4.6(2.0)	4.8(1.1)	<i>P</i> =0.39	45.5(42.0)	15.4(12.4)
Abduction	1.3(1.2)	1.7(1.5)	<i>P</i> =0.21	1.6(1.1)	2.3(1.1)	<b><i>P</i>=0.02</b>	120.9(94.1)	55.0(50.6)
<u>Pelvic Tilt</u>								
Tilt (Stance)	1.7(1.8)	4.0(3.0)	<i>P</i> =0.10	4.4(0.5)	3.2(3.7)	<i>P</i> =0.28	106.7(87.5)	151.7(145.7)
Tilt (Swing)	4.7(1.4)	2.8(3.2)	<i>P</i> =0.15	2.3(1.9)	3.8(3.3)	<i>P</i> =0.18	95.8(86.1)	155.0(146.9)

**Table 5.7:** Kinetic Variables Camwalker Group

Mean (SD) [N-m/kg] values for each dependent variable and results of independent *t*-tests comparing the limb across conditions. ASI values shown underneath *P* values indicate level of asymmetry in the braced condition or normal condition.

Dependent Variable	Braced Limb			Unbraced Limb			ASI	
	Braced Condition	Normal Condition	<i>P</i> value	Braced Condition	Normal Condition	<i>P</i> value	Braced Condition	Normal Condition
<u>Ankle Moments</u>								
Dorsiflexion	N/A	1.3(0.2)	N/A	1.6(0.1)	1.3(0.3)	<i>P</i> =0.11	44.0(57.8)	13.9(18.1)
<u>Knee Moments</u>								
Extension	0.7(0.4)	0.5(0.4)	<b><i>P</i>=0.01</b>	0.9(0.9)	0.7(0.3)	<i>P</i> =0.29	23.1(20.4)	52.6(30.3)
Flexion	1.0(0.7)	0.8(0.4)	<i>P</i> =0.25	1.1(0.9)	1.1(0.8)	<i>P</i> =0.36	45.6(36.0)	34.8(22.4)
<u>Hip Moments</u>								
Extension	1.2(0.5)	0.9(0.3)	<i>P</i> =0.12	1.7(1.7)	1.1(1.2)	<i>P</i> =0.06	37.2(36.5)	104.7(79.2)
Flexion	1.1(0.3)	1.4(0.4)	<i>P</i> =0.06	1.8(1.7)	2.3(1.5)	<i>P</i> =0.11	59.9(62.9)	46.0(45.2)
Abduction	1.1(0.4)	0.8(0.2)	<i>P</i> =0.14	1.1(0.5)	1.3(0.8)	<i>P</i> =0.18	58.2(10.6)	45.9(30.2)

**Table 5.8:** Kinetic Variables Corset-Style Group

Mean (SD) [N-m/kg] values for each dependent variable and results of independent *t*-tests comparing the limb across conditions. ASI values shown underneath *P* values indicate level of asymmetry in the braced condition or normal condition.

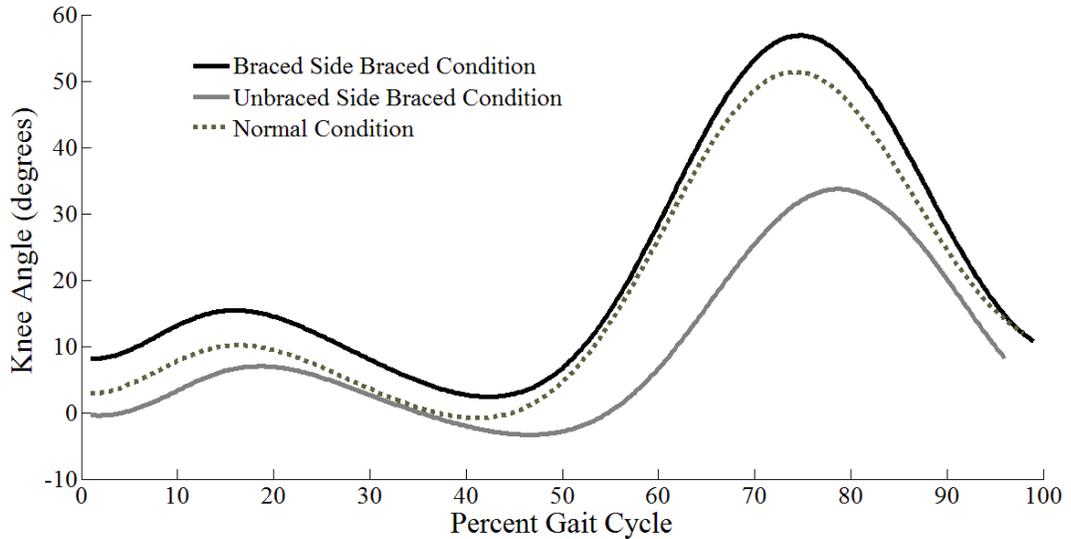
Dependent Variable	Braced Limb			Unbraced Limb			ASI	
	Braced Condition	Normal Condition	<i>P</i> value	Braced Condition	Normal Condition	<i>P</i> value	Braced Condition	Normal Condition
<u>Ankle Moments</u>								
Dorsiflexion	N/A	1.5(0.1)	N/A	1.6(0.1)	1.5(0.0)	<i>P</i> =0.13	15.4(14.6)	3.9(2.3)
<u>Knee Moments</u>								
Extension	0.5(0.2)	0.4(0.2)	<i>P</i> =0.21	0.5(0.2)	0.4(0.2)	<b><i>P</i>=0.05</b>	37.2(35.3)	56.4(35.2)
Flexion	0.4(0.2)	0.5(0.2)	<i>P</i> =0.18	0.4(0.2)	0.5(0.2)	<i>P</i> =0.07	42.7(14.8)	27.3(29.5)
<u>Hip Moments</u>								
Extension	1.1(0.6)	1.1(0.4)	<i>P</i> =0.46	1.1(0.3)	1.1(0.4)	<i>P</i> =0.40	36.0(21.8)	24.7(14.4)
Flexion	1.1(0.5)	1.0(0.4)	<i>P</i> =0.27	1.2(0.3)	1.0(0.4)	<i>P</i> =0.17	49.4(24.7)	42.8(32.6)
Abduction	0.7(0.3)	0.6(0.3)	<i>P</i> =0.09	0.8(0.2)	0.8(0.3)	<i>P</i> =0.18	20.7(18.9)	29.4(24.6)

**Table 5.9:** Kinetic Variables PTB Group

Mean (SD) [N-m/kg] values for each dependent variable and results of independent *t*-tests comparing the limb across conditions. ASI values shown underneath *P* values indicate level of asymmetry in the braced condition or normal condition.

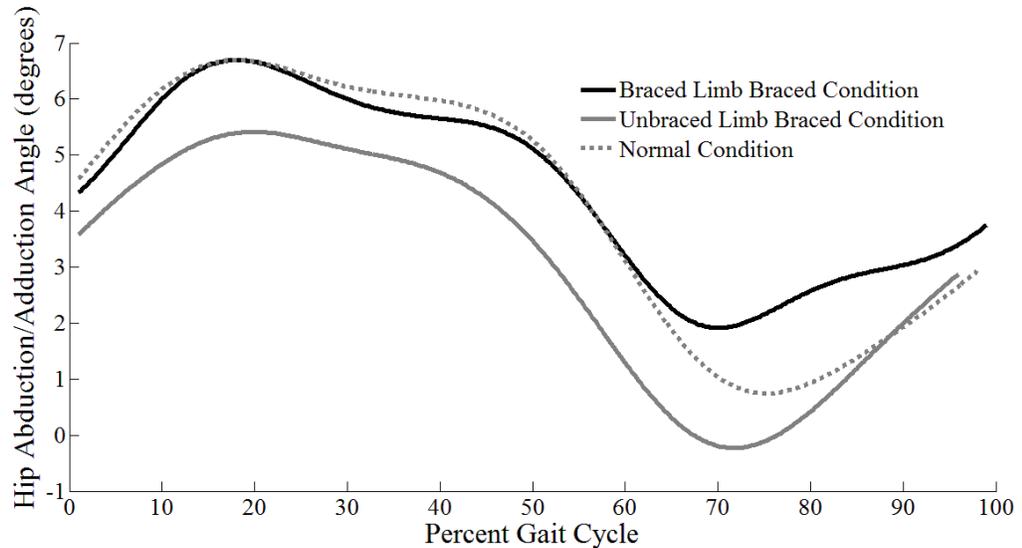
Dependent Variable	Braced Limb			Unbraced Limb			ASI	
	Braced Condition	Normal Condition	<i>P</i> value	Braced Condition	Normal Condition	<i>P</i> value	Braced Condition	Normal Condition
<u>Ankle Moments</u>								
Dorsiflexion	N/A	1.5(0.1)	N/A	1.6(0.1)	1.6(0.1)	<i>P</i> =0.20	3.7(3.0)	2.9(0.9)
<u>Knee Moments</u>								
Extension	1.4(1.0)	1.5(1.3)	<i>P</i> =0.30	0.5(0.1)	0.3(0.2)	<i>P</i> =0.11	77.3(61.1)	104.8(49.3)
Flexion	1.0(0.7)	2.0(1.7)	<i>P</i> =0.14	0.7(0.1)	0.7(0.0)	<i>P</i> =0.41	57.0(36.2)	72.0(66.3)
<u>Hip Moments</u>								
Extension	1.5(1.8)	2.8(2.3)	<i>P</i> =0.16	1.2(0.5)	1.0(0.4)	<b><i>P</i>=0.04</b>	75.2(65.3)	73.2(53.8)
Flexion	1.8(3.7)	3.2(2.7)	<i>P</i> =0.23	1.4(0.6)	1.4(0.5)	<i>P</i> =0.46	136.8(189.8)	76.4(55.3)
Abduction	1.6(1.3)	1.3(1.0)	<i>P</i> =0.17	0.5(0.1)	0.4(0.1)	<i>P</i> =0.20	103.2(48.0)	83.1(54.4)

In the camwalker group, four variables were significantly different. The difference between the knee flexion during stance phase in the unbraced limb during normal walking compared to the unbraced limb during braced walking can be found in Figure 5.3. The knee flexion was  $6.3\pm 3.3^\circ$  lower during the braced condition ( $P=0.01$ ).



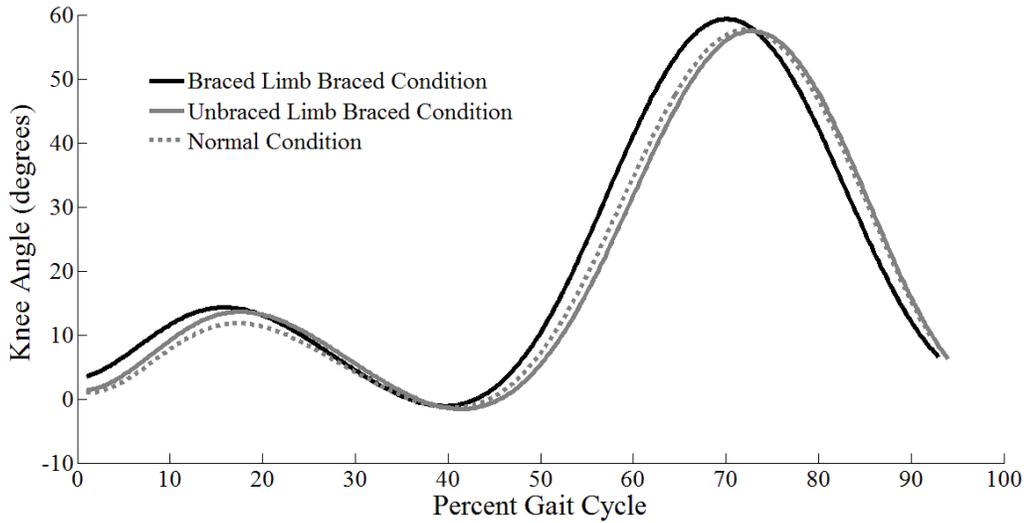
**Figure 5.3:** Ensemble average of knee angle over the gait cycle for the camwalker group showing the differences between braced and unbraced limb during braced condition with normal condition.

Hip adduction in the braced limb was  $1.3\pm 0.1^\circ$  higher in the braced limb during the braced condition compared to the normal condition ( $P=0.02$ ), and hip abduction in the braced limb was  $2.9\pm 0.3^\circ$  lower during the braced condition compared to the normal condition ( $P<0.001$ ) (Fig. 5.4).

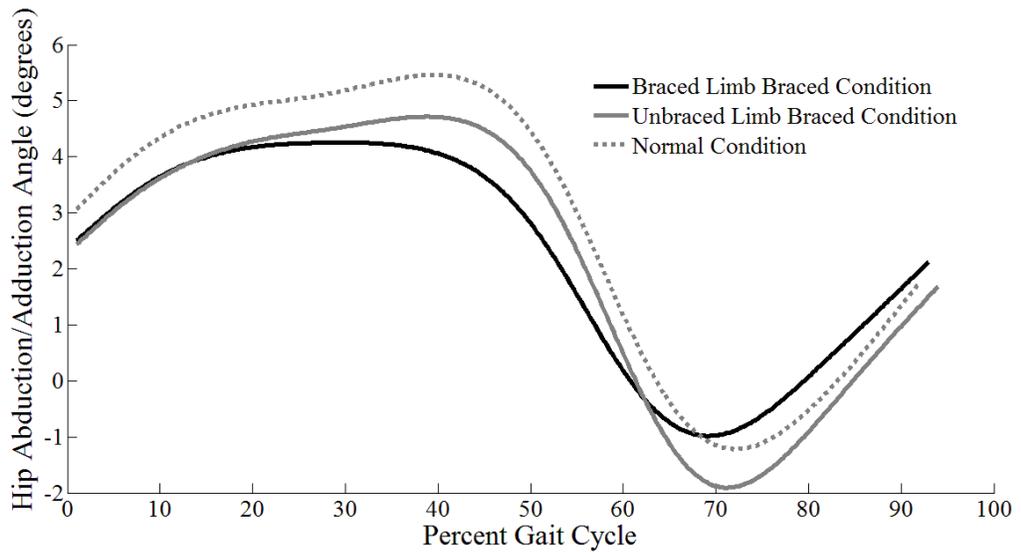


**Figure 5.4:** Hip adduction/abduction over the gait cycle for camwalker group showing the difference between braced and unbraced limb during braced condition with normal condition.

In the PTB group, two variables showed differences. Peak knee flexion during stance was  $2.4 \pm 1.6^\circ$  higher in the braced limb during the braced condition compared to the normal condition ( $P=0.04$ ) (Fig. 5.5). Peak hip abduction was  $0.7 \pm 0.1^\circ$  lower in the unbraced limb during the braced condition compared to normal condition ( $P=0.02$ ) (Fig. 5.6). There were no differences found in the corset-style group kinematics.

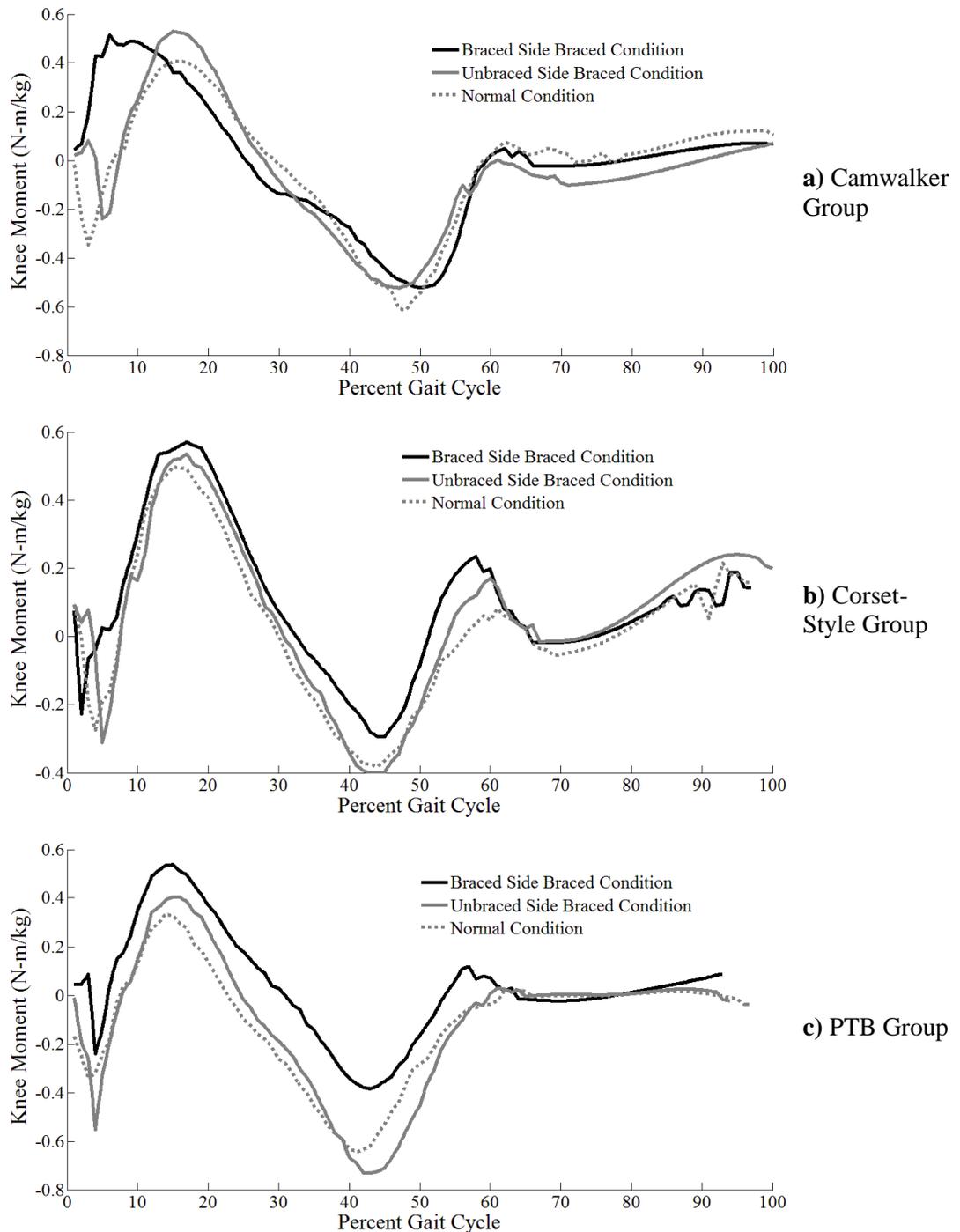


**Figure 5.5:** Knee flexion angle over the gait cycle for PTB group showing the difference between braced and unbraced limbs during braced condition with normal condition.

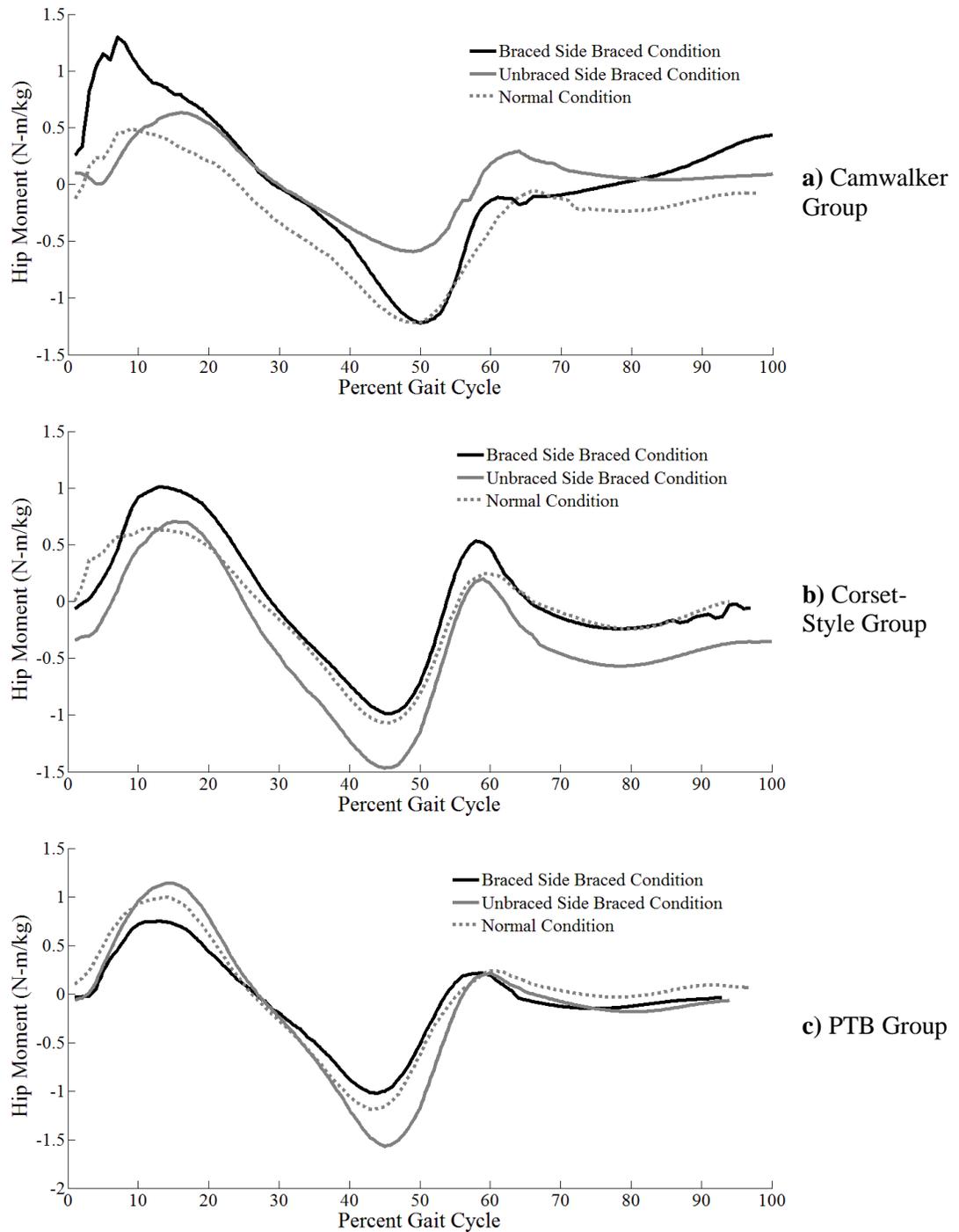


**Figure 5.6:** Hip adduction/abduction over the gait cycle for PTB group showing the difference between braced and unbraced limbs during braced condition with normal condition.

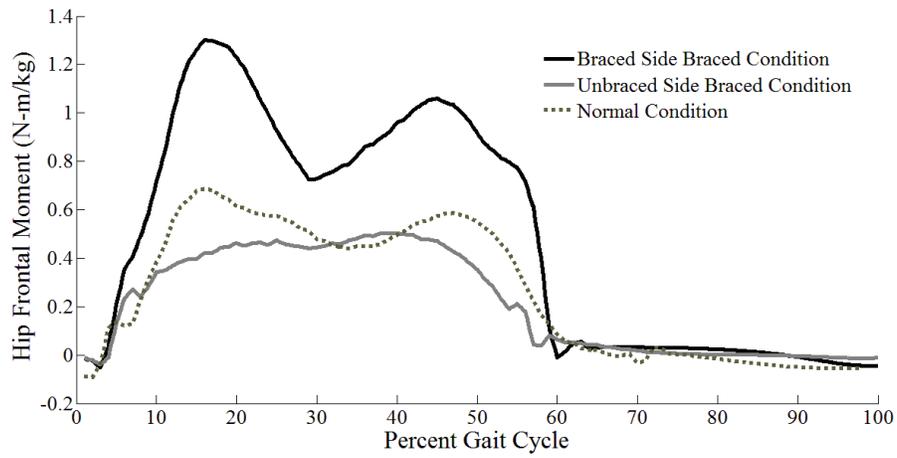
Kinetics varied in the braced limb across conditions for all three bracing groups. In the camwalker group, the knee extension moment was  $0.2 \pm 0.1$  N-m/kg higher in the braced limb during the braced condition compared to the unbraced condition ( $P=0.01$ ). In the corset-style group, the knee extension moment was  $0.1 \pm 0.1$  N-m/kg higher in the unbraced limb during the braced condition compared to the normal condition ( $P=0.05$ ). In the PTB group, the hip extension moment was  $0.2 \pm 0.1$  N-m/kg higher in the unbraced limb during the braced condition compared to the normal condition ( $P=0.04$ ). Ensemble averages of knee moments (Fig. 5.7), sagittal hip moments (Fig. 5.8), and frontal hip moments (Fig. 5.9) show differences between the both limbs in the braced condition to the normal condition.



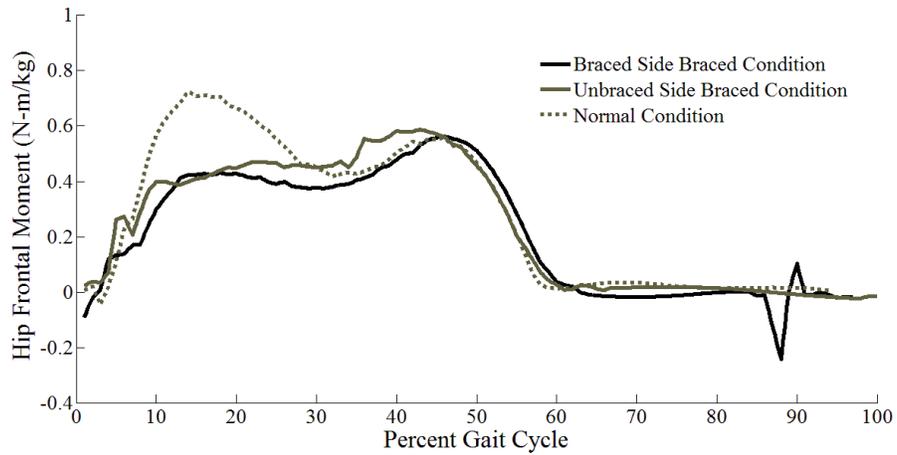
**Figure 5.7:** Knee moment for **a)** camwalker, **b)** corset-style, and **c)** PTB groups over entire gait cycle showing differences between braced and unbraced limb during braced condition with normal condition.



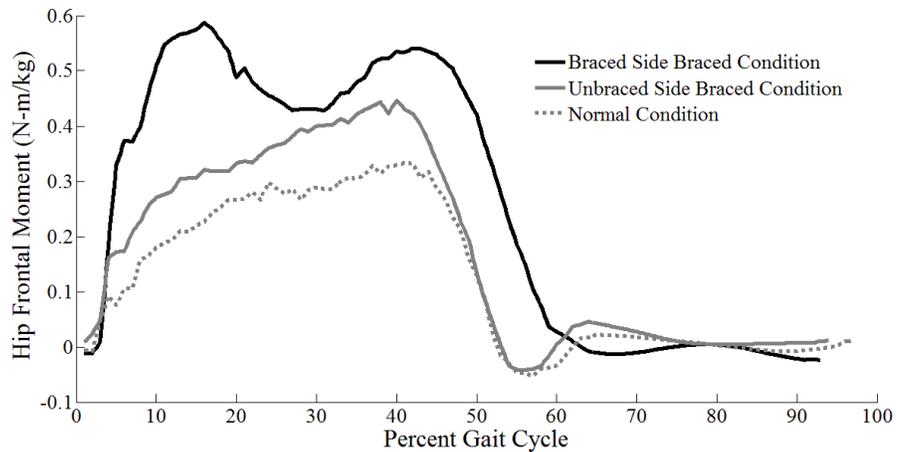
**Figure 5.8:** Hip sagittal moment for **a)** camwalker, **b)** corset-style, and **c)** PTB groups over entire gait cycle showing differences between braced and unbraced limb during braced condition with normal condition.



**a) Camwalker Group**



**b) Corset-Style Group**



**c) PTB Group**

**Figure 5.9:** Hip frontal moment for **a)** camwalker, **b)** corset-style, and **c)** PTB groups over entire gait cycle showing differences between braced and unbraced limb during braced condition with normal condition.

## 5.4 Discussion

The objective of this pilot investigation was to evaluate the change in the participants' gait pattern while wearing an AFO. Since the AFOs each altered the height, width, and mass of the lower limb, the participants needed to alter their walking style from their natural gait pattern during the braced condition.

The camwalker group had the most changes in their gait pattern from normal walking. This is most likely because the camwalker AFO creates the largest changes to the lower limb geometry. The increase in hip extension during stance on the braced limb, along with the higher pelvic tilt and hip adduction, may be due to the added height. The very low hip abduction in the braced limb during the swing phase of gait is unexpected. We believed the main strategy to overcome the camwalker AFO would be to swing it outward to get floor clearance, however, the abduction results do not show this. We believe the results show the participants adapted a lifting strategy. This is supported by decreased knee flexion in the unbraced limb during braced walking during stance. The limb is more extended to give extra height to the hips to help the braced limb clear the ground.

In the PTB group, higher knee flexion was present in the braced limb during the braced condition. This was probably due to the design of the AFO. It makes contact with the knee and interferes with its motion. The PTB AFO may not allow the knee to extend naturally, thereby increasing the knee's flexion throughout stance. In addition, the PTB group exhibited lower hip abduction angle in the unbraced limb during the braced condition. However, the difference in peak angle is less than one degree, and thus most likely there is no difference.

The corset-style group showed no differences in kinematic variables. This may be because the corset-style AFO is the most form fitting AFO and does not completely lock the ankle into position. Therefore, the differences caused by the corset-style AFO were minimal.

The kinetic results show increased ankle moment on the unbraced side during the braced condition in the camwalker group, though we couldn't test significance as the ankle moment on the braced limb is artificially create by the AFO. The increased ankle moment follows from the lack of an ankle on the braced limb, as the unbraced limb would need to compensate for no ankle on the contralateral side. The increased knee extensor moment on the braced limb is probably due to compensation from not having free moving ankle joint.

The corset-style group only had one significantly different variable which was an increased knee extensor moment on the unbraced side during braced walking. This could be created by the extra effort the unbraced limb must take to compensate for the minor loss of control of the braced limb. Even though the corset-style AFO minimally affected the kinematic data due to its low profile, adding the corset-style AFO to the limb is still awkward and it does limit the motion of the ankle joint. If the stance phase on the braced limb is not fully controlled, the contralateral limb can regain control during its stance phase, as seen by the increased knee extensor moment.

Lastly, in the PTB group, there was a higher hip flexor moment on the unbraced side during braced walking. The hip flexor moment is a body controlling mechanism. The hips help control and balance not just the lower limbs but the torso as well. The increase

hip flexor moment could be indicating that the PTB patients had more imbalance that needed to be corrected through the hips.

The results of this pilot study indicate that each of the AFOs do alter the participants' natural gait pattern. The level to which they affect it are different, with the camwalker AFO creating the most deviations from the natural gait cycle of the participants. It should be noted however that our sample size was small and that the corset-style and PTB AFO users had previously worn those braces and thus may have trained their bodies to correct their gait deviations. The camwalker participants had only a few minutes to adjust to the AFO, which is not enough time to learn the most efficient way to wear the AFO. Thus, some of the deviations seen in the camwalker may lessen the longer the wearer uses the camwalker.

## **5.5 Conclusion**

Gait asymmetry can be a serious issue, creating other problems such as lower back and hip pain. The results of this pilot investigation indicate that wearing an AFO alters the geometry and inertial properties of the lower limb and thus can create gait abnormalities. A clinical trial involving more participants and additional AFOs is necessary to fully examine the effect an AFO has on a patient's natural gait cycle.

## **6. Conclusions**

There were two primary objectives for this investigation. The first was to examine the ability of three commonly used ankle foot orthoses to reduce weight bearing on the lower limb. Secondly, this project attempted to quantify the changes made in the natural gait pattern of the participants when wearing one of the AFOs.

### **6.1 Weight Reduction Capability of Three Common Ankle Foot Orthoses**

The percent weight reduction occurring in the braced limb was found for each participant by examining the difference in force output from an external force platform and an insole pressure measurement device. The results showed that the corset-style brace was the most effective AFO in reducing the weight bearing through the lower shank by approximately 30%. The camwalker and PTB AFOs only reduced the weight bearing by approximately 10% and 8%, respectively. While none of our results were significant, they still could be clinically relevant as they clearly demonstrate a lack of weight reduction in two of the AFOs as indicated by the results from the 95% confidence interval. The confidence interval showed that the corset-style AFO is significantly different from no reduction in weight 94% of the gait cycle. The camwalker AFO was significantly different from no reduction in weight 67% of the gait cycle, and the PTB AFO 16% of the gait cycle.

Future development of this project should address the inaccuracies inherent in the insole pressure measurement devices. In addition, a much larger sample size would make the results more impactful and our results indicate that a clinical trial should be done on AFOs. Lastly, a future clinical trial should include other types of AFOs, as there is a wide variety in design and cost so that the best, most effective device may be found.

## **6.2 Biomechanical Analysis of Three Common Ankle Foot Orthoses**

Using the motion capture data, a kinematic and kinetic analysis was performed to compare the dynamics of braced and normal walking. Kinematic variables examined included peak hip flexion, hip extension, and knee flexion during stance and swing phase, as well as peak hip adduction, abduction and mediolateral pelvic tilt. Kinetic variables examined included peak ankle dorsiflexion, knee, and hip extension and flexion moments, as well as peak abduction moment. An Absolute Symmetry Index was calculated for each variable to help quantify the changes to the participants' gait cycles.

All three of the AFOs examined caused the participants to alter their natural gait pattern. The results showed that the camwalker AFO affected more kinematic and kinetic variables than the corset-style and PTB AFOs. It appears that the camwalker participants adopted a lifting strategy to overcome the bulky addition of the brace rather than a swinging strategy. The strategies of the corset-style and PTB participants was more subtle, but it also seems that they used a lifting strategy as well.

Future development of this project should more closely examine the connection of gait asymmetries and lower back or hip pain. It is not clear yet if long-term AFO use

could lead to lower back pain, but studies linking low back pain with long term prosthetic use indicate that AFO use may cause the same comorbidities.

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