EVALUATION OF GAIT KINEMATICS AND KINETICS USING A POWERED ANKLE-FOOT ORTHOSIS FOR GAIT ASSISTANCE IN PEOPLE WITH MULTIPLE SCLEROSIS

BY

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THESIS

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ABSTRACT

Gait impairment is one of the many motor symptoms of multiple sclerosis (MS) but has the largest perceived impact on overall quality of life for people with the disease. Currently, there is no cure for MS and no treatments or therapies can reverse the progression of symptoms. Typically, gait speed becomes slower with disease progression requiring the prescription of assistive devices to maintain or improve movement ability and promote independence. For people with MS who suffer from drop foot (i.e., the diminished or lack of ability to dorsiflex the foot), prescription of a passive ankle-foot orthosis is common. The purpose of these devices is to hold the foot in neutral position and prevent the foot from dropping during forward swing. Although these devices can be beneficial, ankle-foot orthoses generally prevent the ability to plantarflex the ankle, and successful implementation of these devices has been limited. Recently, powered orthotics and exoskeletons have been developed for people with neurological impairments or injuries that impact gait function. These devices can provide active motion control and assistance in both plantarflexion and dorsiflexion of the ankle. The purpose of this thesis was to investigate the utility of a powered ankle-foot orthosis for gait assistance in people with MS. Specifically, the analyses performed in this thesis focused on the kinematic and kinetic changes in gait associated with a powered ankle-foot orthosis compared to a passive ankle-foot orthosis and walking without an assistive device. Results indicate that the vertical ground reaction force during the propulsive phase of gait (i.e., terminal stance) was increased with the powered ankle-foot orthosis, counteracting the diminished behavior typically observed in people with MS. However, peak plantarflexor torque during this same phase of gait was diminished compared to walking without an assistive device, which may have been due to a slower gait speed. The only difference between a powered ankle-foot orthosis or prescribed passive ankle-foot orthosis was an increased mid-stance minima in the vertical ground reaction force, which could also be an indicator of slower gait speed. The findings of this study provide an initial baseline for future studies of powered orthoses in people with MS. Further investigation and tuning of the powered orthosis used in this study is needed.

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TABLE OF CONTENTS

TABLE OF FIGURES

TABLE OF TABLES

1 Literature Review

1.1 Walking Gait Analysis

Through years of development, humans acquire a specific method of walking, or gait. The purpose of gait is to move the body forward through a repetitive pattern of lower limb movements (Perry 1992). The basic tasks of a person's gait are to maintain support of the upper body during stance, prevent falling, and allow for safe clearance of the swing limb as it comes forward (Winter 1989). The central nervous system coordinates muscle activations to control the forces (kinetics) generated by the lower limbs to achieve the desired motion (kinematics) (Winter 1984). A gait cycle is defined as starting with the heel strike of one leg and ending with the next heel strike of the same leg. The walking gait cycle for an adult human can be divided into two phases, stance and swing. Stance is approximately the first 60% of the gait cycle when the foot is on the ground and swing is when the foot is off the ground for the last ~40%. These phases are typically divided into sub-phases and events that have functional significance (Perry 1992).

Initial Contact (~0-2% of the gait cycle): Heel strike of the lead (ipsilateral) limb occurs while the trailing (contralateral) limb is still in contact with the ground.

Loading Response (~2-10% of the gait cycle): The contralateral limb propels the weight of the body towards the ipsilateral limb. During this sub-phase, the ground reaction forces (GRFs) in the vertical and anterior-posterior directions steadily increase under the ipsilateral limb, and the ankle generates a dorsiflexor torque as the foot comes in contact with the ground (Winter 1984).

Mid-Stance (~10-30% of the gait cycle): The ipsilateral limb fully supports the weight of body while the contralateral limb begins to swing forward for the next step. The kinetics associated with this phase of gait are a slight decrease in the vertical GRFs while the anterior-posterior GRFs approach zero under the ipsilateral limb. Additionally, the ankle joint steadily progresses into dorsiflexion with an increasing plantarflexor torque being generated (Winter 1984).

Terminal Stance (~30-50% of the gait cycle): After mid-stance, the weight of the body moves in front of the ipsilateral limb as it becomes necessary to propel the body towards the contralateral limb. This is accomplished through an increase in vertical and anterior-posterior GRFs and plantarflexor torque at the ankle of the stance (ipsilateral) limb. At this point, the GRFs generated by the body are propulsive to drive the body forward (Winter 1984).

Pre-Swing (~50%-60% of the gait cycle): The contralateral limb comes in contact with the ground as the ipsilateral limb propels the weight of the body towards the contralateral limb. During this sub-phase, the body is in double support again with both legs on the ground. At the end of pre-swing, the stance leg comes of the ground and the contralateral limb accepts the weight of the body.

Initial Swing (~60%-73% of the gait cycle): This phase is the beginning of swing for the ipsilateral limb. No ground reaction forces are present for the ipsilateral limb because it is not in contact with the ground. The ankle joint of the ipsilateral limb generates a dorsiflexor torque to prevent the toes from contacting the ground (Winter 1984).

Mid-Swing (~73%-87% of the gait cycle): The ipsilateral limb moves from being in line with the contralateral limb to slightly ahead of it at the end of this sub-phase. A dorsiflexor torque on the ankle of the ipsilateral limb continues to be generated to ensure toe clearance (Winter 1984).

Terminal Swing (~87%-100% of the gait cycle): The ipsilateral limb moves into position for contact with the ground. A dorsiflexor torque is still being generated to prevent the foot from slapping down during heel strike (Winter 1984). The sub-phase ends with heel strike of the ipsilateral limb.

Using the tools of biomechanics to measure the kinetics and kinematics of the lower limbs, it is possible to address the potential changes and underlying neural control of movement. This information provides insight into possibilities for interventions and rehabilitation for gait disorders to improve overall gait function.

1.2 Multiple Sclerosis

Multiple sclerosis (MS) is a neurodegenerative disease that affects 250,000 to 350,000 people in the United States (Anderson et al. 1992; Noseworthy et al. 2000). Symptoms of MS are caused by a death of white matter in the central nervous system. White matter cells, or oligodendrocytes, provide insulation to neurons for proper signal transduction. The central nervous system malfunctions with lesions of white matter leading to several impairments including the motor symptoms of spasticity, decreased postural control, fatigue, and abnormal gait (Freal et al. 1984; Noseworthy et al. 2000; Sosnoff et al. 2011; Motl 2013). The primary cause of MS is not fully understood and no cure exists. The different types of disease course are not fully defined (Confavreux and Vukusic 2006), but the majority of people with MS start in what is considered relapsing-remitting multiple sclerosis (RRMS) (Noseworthy et al. 2000; Confavreux and Vukusic 2006). In RRMS, the symptoms can present over a periods of days, and then revert back to normal spontaneously (Noseworthy et al. 2000). A smaller portion of people with MS start in a more progressive state known as primary-progressive multiple sclerosis (PPMS) where symptoms gradually increase without relapses (Noseworthy et al. 2000; Confavreux and Vukusic 2006). As the relapses become more frequent and the increase in disease severity becomes more constant, people with RRMS can convert to secondary-progressive multiple sclerosis (SPMS). People with PPMS can also convert to progressive-relapsing multiple sclerosis (PRMS), where the constant progression of the disease is accompanied by relapses (Confavreux and Vukusic 2006). Ultimately, symptoms become more severe and mobility is increasingly limited with disease progression, resulting in decreased quality of life.

Out of all of the symptoms of MS, trouble with walking becomes more common and devastating as the disease progresses. People with MS perceive that decreases in walking ability has the largest impact on overall quality of life compared to other symptoms (Larocca 2011). Furthermore, people with MS perceive that walking is the most valuable function, followed by visual function and cognition (Heesen et al. 2008). Changes in gait associated with MS include slower steps that are shorter and wider compared to controls (Benedetti et al. 1999; Sosnoff et al. 2012). Furthermore, muscle activation can become uncoordinated in the plantarflexors and dorsiflexors of the lower leg (Benedetti et al. 1999). In terms of gait kinematics, people with MS exhibit reduced hip extension, knee extension, and ankle plantarflexion. These changes are accompanied by a decreased ability to generate propulsive forces and plantarflexor torque during terminal stance compared to healthy controls (Benedetti et al. 1999; Kelleher et al. 2010). Other symptoms, such as fatigue, spasticity, and cognitive impairment, can influence gait impairments and increase variability of gait parameters (Benedict et al. 2011; Sehle et al. 2011; Wurdeman et al. 2011; Sosnoff et al. 2012). Fatigue also leads to significantly shorter six-minute walk test (6MW) distances, a metric typically used to measure endurance (Goldman et al. 2008). The mechanisms behind these changes in gait are not fully understood. Nonetheless, it has been proposed that changes in gait associated with MS lead to decreased motor activity, leading to overall deconditioning, and increased motor disability (Motl et al. 2010).

1.3 Treatments and Rehabilitation for MS Gait Impairments

Similar to other neuromuscular disorders, the most common treatment for gait impairments in people with MS is the prescription of some type of assistive device (i.e., ankle-foot orthoses, canes, walkers, or wheelchairs) in order to maintain mobility and quality of life. Wheelchairs become increasingly necessary as gait and balance symptoms become more severe (Fay and Boninger 2002) and are the most commonly prescribed assistive device (Finlayson et al. 2001). Although helpful, there is a lack of evidence to support the prescriptions made of assistive devices for people with MS based on their mobility status (Souza et al. 2010). Furthermore, if the user does not perceive that the device is helping or if it is not meeting their needs, they can often be abandoned (Verza et al. 2006; Souza et al. 2010). Consequently, gait interventions and treatments are needed for people with MS that are perceived to be helpful and can improve mobility before it deteriorates.

Before a wheelchair is needed, passive ankle foot orthoses (AFOs) are prescribed for lower limb weakness or foot drop in MS. Usually consisting of a carbon fiber or plastic material, the main purpose of AFOs is to provide stability at the ankle joint while also providing more normative biomechanics during the swing phase of gait (Souza et al. 2010). Positive results for balance and posture (Cattaneo et al. 2002) as well as gait have been demonstrated with these passive devices. Specifically, improved gait speed, increased dorsiflexion during swing, and decreased metabolic cost have been achieved (Bregman et al. 2010). However, the positive results were only observed in a subset of their population who did not attain neutral ankle angle during swing. For the rest of the participants in that study, the AFO did not provide any further benefit to these parameters because they were less impaired. An energy storing AFO with a spring-like behavior to assist with ankle push-off was able to decrease metabolic cost by 9.8%, and increase the peak plantarflexor torque in patients with MS (Bregman et al. 2012). Work at the ankle also decreased with the AFO, but a shift towards more work produced at the hip was observed. Null results for improving gait endurance (e.g., timed 25 foot walk test) have also been observed with AFOs (Sheffler et al. 2008). Therefore, the results of passive AFOs for mitigating gait impairment of MS have been mixed and limited to specific conditions and populations.

Amongst the treatments for MS gait impairments under development, pharmacological agents and exercise have been demonstrated to improve gait and MS symptoms. Pharmacologically, many drugs have been developed to try and target the mechanisms of MS specifically and reduce the amount of lesions in the brain (Khan et al. 2008). Treatments are able to temporarily reduce disease severity, but no drug has been able to reverse the disease course. Extended-release dalfampridine, a pharmacological agent that improves signal transduction in demyelinated neurons, has been developed to improve ambulation. Results of the clinical trials suggest that gait speed can be improved with the drug as well as other symptoms (i.e., spasticity and strength) (Hersh and Rae-Grant 2012). In addition to pharmacological agents, exercise can provide modest improvement in walking endurance, speed, metabolic cost, and potentially mitigate MS severity (Motl et al. 2010). Furthermore, exercise can increase strength and boost other physiological functions, including anti-inflammatory agents within the body (Motl and Pilutti 2012; Motl and Sandroff 2015). Although not curative, exercise and pharmacological agents provide promising options for increasing mobility and mitigating MS symptoms.

Different forms of physical therapy are an option for gait rehabilitation, especially for patients with severe levels of MS (Expanded Disability Severity Scale (EDSS) ≥ 6.0). Traditional forms of physical therapy (i.e., passive or active movement facilitation by a physical therapist, task-dependent training) can improve gait measures and gait speed (Lord et al. 1998; Wiles et al. 2001). Several studies have shown that using body weight support treadmill training can improve gait (Pilutti et al. 2011) along with muscle strength and spasticity (Giesser et al. 2007).

Body weight support treadmill training requires a physical therapist to physically assist the person's legs while they walk on the treadmill. The training can focus on different components of the walking motion, such as weight acceptance or transfer during the gait cycle. Robotic treadmill training, where this process of repetitive external assistance is provided by a robotic device that attaches to the lower limbs, can induce a similar improvement in gait function as body weight support treadmill training (Lo and Triche 2008) and traditional physical therapy gait training (Schwartz et al. 2012). However, one possible limitation of both robotic and conventional training is the lack of long term positive effects on gait function if the training is stopped (Beer et al. 2008). Furthermore, both types of training are limited to the lab or clinic setting and cannot be used in the changing environments of everyday life. Overall, these studies demonstrate that physical therapy can induce improvements in gait function in people with MS; however, these types of training are limited to the lab or clinic setting, and more needs to be understood about how to induce long term positive effects in gait for people with MS.

1.4 Powered Orthoses

Powered exoskeletons and orthoses are part of a burgeoning field focused on developing daily wear and rehabilitative robotic devices to provide assistance to joints of the lower limbs. For people who are ambulatory with gait impairments, powered orthoses for the knee (Nikitczuk et al. 2007), ankle (Blaya and Herr 2004; Ferris et al. 2005; Ferris et al. 2006; Roy et al. 2009; Shorter et al. 2011a; Mooney et al. 2014; Neubauer et al. 2014), or both the knee and ankle simultaneously (Sawicki and Ferris 2009) have been developed. Using active motion control, powered orthoses can increase propulsive torque during late stance (Ferris et al. 2005), and decrease the metabolic cost of walking (Mooney et al. 2014) in healthy individuals. Recent studies have demonstrated that powered AFOs can encourage more normative biomechanics in people with gait impairments. Pilot data from post-stroke patients demonstrated an increase in peak plantarflexor torque of 16% while walking on a treadmill (Takahashi et al. 2015) and improved motor control and movement smoothness after seated rehabilitation of ankle joint movement (Roy et al. 2009). Additionally, a variable–impedance AFO was able to prevent foot slap during loading response for people with foot drop (Blaya and Herr 2004). Overall, powered orthotics have the capability of providing active motion control and assistance during the gait cycle that can help overcome gait impairments.

A portable powered ankle-foot orthosis (PPAFO) test bed has been developed to investigate the application of powered devices for the ankle-foot complex in gait disorders (Figure 1). The PPAFO is powered by a bidirectional pneumatic rotary actuator that is capable of providing both dorsiflexor and plantarflexor torque during the gait cycle (Shorter et al. 2011a). On board electronics control the flow of compressed CO_2 by way of two solenoid values. Sensors on the device inform different control schemes that determine the correct timing of actuation based on the current gait state (0-100% gait cycle) of the user (Li et al. 2011; Shorter et al. 2011a; Islam et al. 2016; Islam and Hsiao Wecksler 2016). The PPAFO is also able to control gait in multiple environments, including ascending and descending stairs or ramps (Li and Hsiao-Wecksler 2013). Pilot studies with the device have demonstrated improved propulsive ground reaction forces during terminal stance for cauda equina syndrome, a neurological condition where the person is unable to generate plantarflexor torque at the ankle (Shorter et al. 2011b). Thus, the PPAFO is a capable testbed for investigating the utility of a powered ankle foot orthosis for improving gait kinematics and kinetics in people with MS.

1.5 Goals of Thesis

As part of a larger study focused on the application of a powered ankle-foot orthosis in people with MS across multiple disability levels (Boes et al. 2015), the goal of this thesis was to investigate the potential changes in kinematics and kinetics of gait using the PPAFO compared to walking with a prescribed orthosis and shoes. Chapter 2 contains the main experimental study. Chapter 3 covers general conclusions of the study and possible future directions for this research. Two appendices were included: the first contains all of the ankle angle data for the participants (Appendix A); the second contains the description and results of a small study regarding whether to include the PPAFO weight in ankle torque calculations (Appendix B).

Figure 1: The portable powered ankle-foot orthosis (PPAFO).

2 Comparison of Gait Kinematics and Kinetics between a Passive and Powered Ankle-Foot Orthoses in People with Multiple Sclerosis

2.1 Abstract

Prescription of a passive ankle-foot orthosis is common for gait impairment in people with multiple sclerosis (MS); however, successful implementation has been limited. Furthermore, they can often be abandoned by the user if they are not perceived to be helpful. In the current study, we used a portable powered ankle-foot orthosis (PPAFO) testbed to evaluate the utility of powered ankle orthoses for improving gait biomechanics in people with MS. Five participants from a larger study focused on the efficacy of the PPAFO in people with MS performed short walking trials on a walkway with embedded force plates in three footwear conditions: (1) normal walking shoes [SHOES], (2) their prescribed AFO [AFO], and (3) the PPAFO [PPAFO]. Assistive devices were worn on the more impaired limb. Twenty-seven relevant kinetic and kinematic parameters (ground reaction force, ankle torque and angle) were analyzed for the more impaired limb. The propulsive vertical ground reaction force was found to increase with the PPAFO compared to the SHOES condition (p=0.026). Several significant changes (i.e., increased mid-stance minima of vertical ground reaction forces compared to the AFO condition (p=0.048) and decreased peak plantarflexor torque compared to the SHOES condition (p=0.008)) were consistent with the slower gait speed in the PPAFO condition. No significant changes were observed in the kinematic parameters. Overall, these results suggest that the PPAFO could be used to modify gait kinetics in people with MS. However, further investigation into increasing gait speed and the corresponding kinetics, possibly through changes in device control and extended training time with PPAFO, is needed.

2.2 Introduction

Multiple sclerosis (MS) is a neurodegenerative disease that causes several motor symptoms that become more severe as the disease progresses. Symptoms include spasticity, decreased postural control, fatigue, and abnormal gait (Freal et al. 1984; Noseworthy et al. 2000; Sosnoff et al. 2011; Motl 2013). People with MS perceive that decreases in walking ability have the largest impact on overall quality of life compared to other symptoms (Larocca 2011). Typically, gait speed is slower (Benedetti et al. 1999; Kelleher et al. 2010; Sosnoff et al. 2012) and early fatigue is common (Sehle et al. 2011; McLoughlin et al. 2015). Changes in biomechanics that are evident with increasing disease severity include decreases in ankle plantarflexion, hip extension, knee extension, plantarflexor torque, braking ground reaction forces during loading response, and propulsive ground reaction forces during terminal stance (Benedetti et al. 1999; Kelleher et al. 2010). Interventions or therapies that are able to target these changes in biomechanics could improve gait of people with MS.

The prescription of ankle foot orthoses (AFOs) for gait impairments of people with MS has had limited success. Use of passive and semi-active AFOs can improve standing balance, but also impair tasks such as rising from a chair or walking (Cattaneo et al. 2002). Positive results including decreased metabolic cost, increased dorsiflexion during swing, and increased walking speed were observed in patients with MS wearing a low-stiffness, semi-active AFO (Bregman et al. 2010). However, these positive results were only observed in a subset of their population that did not attain neutral ankle angle during swing. The population that did not benefit from the AFO were less impaired. A similar energy storing AFO with a spring-like behavior to assist with ankle push-off was able to decrease metabolic cost by 9.8% and increase the peak plantarflexor torque in patients with MS (Bregman et al. 2012). Work at the ankle also decreased with the AFO, but a shift towards more work produced at the hip was observed. In contrast, no positive results in a functional gait test (i.e., timed 25 foot walk) with the use of AFOs in people with MS have also been found (Sheffler et al. 2008). In sum, the results have been mixed and limited to specific conditions or populations for prescribed AFOs in mitigating gait impairment of MS.

Powered orthotic devices have been developed, enabling active motion control during the gait cycle. These powered orthoses have been designed to actuate the knee (Nikitczuk et al. 2007), ankle (Blaya and Herr 2004; Ferris et al. 2005; Ferris et al. 2006; Roy et al. 2009; Shorter et al. 2011a; Mooney et al. 2014; Neubauer et al. 2014), or both the knee and ankle at the same time (Sawicki and Ferris 2009). Devices for the ankle are capable of increasing plantarflexor torque (Ferris et al. 2005), and decrease the metabolic cost of walking (Mooney et al. 2014) in healthy individuals. For people with gait impairment, devices for the ankle have been developed to prevent foot slap (Blaya and Herr 2004) and improve movement capabilities of patients post-stroke (Roy et al. 2009; Takahashi et al. 2015). For example, a powered ankle foot orthosis that provided plantarflexion assistance was able to increase plantarflexor torque in people post-stroke (Takahashi et al. 2015). Other robotic gait rehabilitation devices (i.e., robotic treadmill training) in which the person is driven through the walking motion by a robotic device can also be effective in improving gait speed in people with MS (Beer et al. 2008; Lo and Triche 2008), especially in the later stages of disease progression. However, no studies have directly investigated the use of powered ankle-foot orthoses in people with MS. Therefore, devices that provide powered assistance to joints of the lower limbs during the gait cycle may be a promising new option for mitigation and rehabilitation of gait impairments of people with MS.

A portable powered ankle-foot orthosis (PPAFO, Figure 2) testbed has been developed to investigate applications of powered ankle assistance during gait (Li et al. 2011; Shorter et al. 2011a). On board electronics can be programmed to provide actuation during the gait cycle (Li et al. 2011; Islam et al. 2016; Islam and Hsiao Wecksler 2016) and changing environments (e.g. walking up/down stairs or slopes (Li and Hsiao-Wecksler 2013)). The PPAFO has been applied in populations such as mild spinal cord injury due to cauda equina syndrome (Shorter et al. 2011a) and MS (Boes et al. 2015). The PPAFO increased the propulsive vertical ground reaction forces in a single patient with cauda equina syndrome who was unable to properly generate plantarflexor torque at the ankle (Shorter et al. 2011a). In a small population (n = 16) of people with MS, use of the PPAFO decreased stride length, stride velocity, and 6-minute walk distance compared to the patient's prescribed AFO in a moderate MS severity group (Boes et al. 2015).

The research objective of this study was to compare the changes in gait kinematics and kinetics induced by a prescribed passive AFO and a powered ankle-foot orthosis in people with MS. We hypothesized that due to the added plantarflexor torque assistance of the powered ankle-foot orthosis, peak vertical and anterior-posterior propulsive forces, ankle torque, and plantarflexion angle during terminal stance/pre-swing could be increased with the PPAFO compared to the prescribed AFO or when walking with only shoes.

2.3 Methods

2.3.1 Participants

Sixteen participants (12 female, 4 male, age: 54.6 ± 5.3 yrs, weight: 83.4 ± 27.4 kg, height: 170.0 ± 8.5 cm, Expanded Disability Status Scale (EDSS): 5.1 ± 1.1, Table 1) with diagnosed MS participated in a larger protocol focused on testing the efficacy of the PPAFO for gait assistance (i.e., metabolic cost, spatiotemporal gait measurements) (Boes et al. 2015). For the current study, only five of the sixteen participants were analyzed due to reasons outlined in section 2.3.6. Each participant wore a clinician-prescribed AFO for gait assistance on their more impaired limb. One participant wore a passive orthosis on their more impaired limb for all conditions. The study protocol was approved by the Institutional Review Board of the University of Illinois at Urbana-Champaign. All participants signed informed consent forms prior to participation in the study.

2.3.2 Portable Powered Ankle Foot Orthosis

The portable powered ankle foot orthosis (Figure 2) consisted of a custom set of carbon fiber shells for the shank (small, medium and large adult male) and foot pieces (able to accommodate US men's sizes 4 to 14) joined through a bi-directional rotary pneumatic actuator (PRN30D-90-45, Parker Hannifin, Cleveland, OH, USA) aligned with the ankle joint axis. An adjustable pressure regulator for the dorsiflexor torque was included in the device. The system utilized a microcontroller (MSP430G2553, Texas Instruments, Dallas, TX, USA) to control two solenoid valves (VUVG 5V, Festo Corp-US, Hauppauge, NY, USA) that modulated the flow of compressed CO_2 to the two chambers of the actuator. The source of compressed CO_2 was a tank (JacPac J-6901-91, 20 oz capacity, Pipeline Inc., Waterloo, ON, Canada). The tank was held by one of the researchers. Force resistive sensors (30-73258, Interlink Electronics Inc., Westlake Village, CA, USA) underneath the heel and toe region of the foot piece detected contact with the ground. A Hall-effect sensor (KMA199E, NXP Semiconductors Netherlands B.V., Eindhoven, Netherlands) was utilized to detect ankle angle. Readings from the three sensors informed the microcontroller of the current gait cycle state (0-100% gait cycle (GC)) of the user (Islam et al. 2016). All sensors were sampled at 120 Hz.

2.3.3 Protocol for PPAFO Training and 6-Minute Walk Tests

The controller for the PPAFO was adjusted through a short training period (>10 minutes). First, the dorsiflexion regulator was heuristically adjusted to hold the participant's toes in the neutral position (~3 Nm at 30 psig). Plantarflexor torque assistance was set to ~10 Nm at a pressure of 100 psig. Initially, the controller provided assistance based on the gait events detected by the heel and toe sensors on the PPAFO (using a Direct Event controller, (Shorter et al. 2011a)). The participants were asked to walk for a minimum of 10 continuous steps while data were collected from the on-board heel, toe, and angle sensors. Then a custom minimization algorithm (MATLAB, Math Works Inc., Natick, MA) used the heel, toe, and angle sensor signals to train a Modified Fractional Time controller (Islam et al. 2016). Dorsiflexor and plantarflexor torque actuation timings were based on normative ankle joint torque behavior during the gait cycle (Figure 3) (Perry 1992). The participant was then asked to walk with the newly adjusted Modified Fractional Time controller and the previous minimization algorithm was run again. This procedure was repeated at least three times. After completing the training protocol, participants were instructed to walk with the participant-specific controller in multiple hallways until they felt comfortable.

As part of the larger protocol, 6MW tests were performed in three footwear conditions. The footwear conditions were wearing (1) their own walking shoes without an AFO [SHOES], (2) their prescribed AFO worn on their more impaired limb [AFO], (3) and the PPAFO worn on their more impaired limb [PPAFO]. For conditions 2 and 3, participants wore their normal walking shoe, except for the one participant who wore another passive AFO on the opposite leg for all conditions. The 6MWs were performed in counterbalance order with the SHOES condition always performed first and at least ten minutes of rest was provided between conditions.

2.3.4 Protocol for Current Study

After completing the last 6MW, participants had a ten minute rest period before performing at least three trials per footwear condition over a set of three force plates embedded in a slightly elevated walkway (Figure 4). Participants were asked to walk at a comfortable walking pace on the walkway; first in one direction and then next trial was collected walking in the opposite direction. Kinematic data from reflective markers placed on anatomical landmarks of the lower limbs were recorded (8-camera Oqus 100 system, Qualisys AB, Gothenburg, Sweden). A 23 marker set was used with reflective markers over the sacrum and ASIS, greater trochanter, thigh, lateral knee epicondyle, medial knee epicondyle, tibia, lateral malleolus, medial malleolus, fifth metatarsal head, and first metatarsal head of both legs. Ground reaction force data were recorded from three force plates (BP600900 and BP400600, Advanced Mechanical Technology Inc., Watertown, MA, USA) at 1000 Hz. Motion and ground reaction force data were filtered using 4th-order Butterworth low-pass filters (cutoff frequencies: 12 Hz for motion, 75 Hz for force data). A “clean” trial was considered as a clean heel strike and stance phase over one or two of the force plates without the contralateral limb contacting the same plate(s). Any trial that was not considered “clean” was discarded. The order of conditions was counterbalanced in the reverse order of the 6MW.

2.3.5 Data Analysis

Kinematic and kinetic measurements (i.e., joint angles, torques, and ground reaction forces) were calculated using Visual 3D v5 (C-motion Inc., Germantown, MD, USA). The built-in heel strike recognition algorithm within Visual 3D was utilized to identify heel strikes and data were verified by visual inspection. Data were exported to MATLAB and normalized to the gait cycle (0-100% GC) between heel strikes. Ground reaction forces were normalized by the participant’s body weight, and torques were normalized by body weight and leg length of the more impaired limb.

Relevant peak behaviors and their timings (as %GC) during multiple phases of the gait cycle were analyzed for the impaired limb (Figure 5). Parameter selection was determined by previous analyses in people with MS (Benedetti et al. 1999; Kelleher et al. 2010). Peak parameters were analyzed from ground reaction forces (GRFs) in the anterior-posterior and vertical directions during the loading response (AP-GRF_{pk1}, V-GRF_{pk1}) and terminal stance (AP-GRF_{pk2}, V-GRF_{pk2}). These parameters indicate the braking and propulsive force capabilities, which are known to be diminished in people with MS (Benedetti et al. 1999; Kelleher et al. 2010). Additionally, the mid-stance minima in the vertical GRFs (V-GRF_{min1}) was analyzed because it has been shown to be significantly larger in people with MS (Benedetti et al. 1999). Three peak parameters were picked in the medial-lateral direction (Benedetti et al. 1999): during loading response (ML-GRF_{pk1}), mid-stance (ML-GRF_{pk2}) and terminal stance (ML-GRF_{pk3}). Six ankle angle parameters were also analyzed to assess the overall ankle angle behavior during the gait cycle (Benedetti et al. 1999): at heel strike (θ_{Ank1}), maximum plantarflexion during stance (θ_{Ank2}), maximum dorsiflexion during stance (θ_{Ank3}), angle at toe-off (θ_{Ank4}), maximum plantarflexion during swing (θ_{Ank5}), and maximum dorsiflexion swing (θ_{Ank6}). Also, three ankle torque parameters during the gait cycle were analyzed: at heel strike torque (T_{Ank1}), max dorsiflexor torque (T_{Ank2}), and max plantarflexor torque (T_{Ank3}). Peak plantarflexor torque has been demonstrated to be diminished in people with MS (Kelleher et al. 2010). Finally, gait speed (GS) was calculated for the stride over the force plates (stride length/stride time) by using the foot center of mass position at consecutive ipsilateral heel strikes. The average value of all trials within a condition for each subject was analyzed. In some instances, only one data point was available for a participant in a given condition.

2.3.6 Reasons for Participant Exclusion in Current Analysis

The sample size for the current study was reduced from the total of 16 for various reasons (Table 1). Eight required the assistance of a walking aid (four used wheeled walkers, four used a single point cane), which interfered with the ground reaction force measurements when they walked over the force plate. The remaining eight participants did not require a walking aid. One participant did not complete the protocol because of data collection equipment malfunction the day of testing and was not rescheduled. Two of them did not have at least one "clean" trial in a footwear condition. Therefore, the data presented in this study were from the remaining five participants (three female, two male, age: 56.6 ± 3.5 yrs, weight: 79.9 ± 15.2 kg, height: 172.7 ± 8.4 cm, EDSS: 4.0 ± 0.1, Table 1). This group was considered to have moderate MS impairment (EDSS ≤ 5.5).

2.3.7 Statistical Analysis

A series of repeated measures MANOVAs were run to assess the effect of footwear condition on kinetic and kinematic data (SPSS v22, IBM Corporation, Armonk, New York). First, a MANOVA including all 16 sagittal plane kinetic parameters (AP-GRF, V-GRF, ankle torque) and gait speed was run. ML-GRF parameters were not included in any statistical analyses due to a high amount of variability in these data (Figure 7, Table 2). Eleven ankle kinematic angle parameters were analyzed in a second MANOVA with a reduced sample size ($n = 3$) due to drop out of shank segment markers of two participants (Table 1). Follow up univariate ANOVAs were performed for relevant parameters. Significance level was set at $\alpha = 0.05$. Post-hoc comparisons were made between conditions using Tukey's Honestly Significant Difference Test (HSD).

2.4 Results

Several significant differences in kinetic parameters were found while none were observed in the ankle angle parameters (Figure 6, Figure 7, Table 2, and Table 3). A significant main effect of footwear was noted for the kinetic parameter MANOVA ($p = 0.001$). Follow up univariate ANOVAs indicated significant differences between footwear conditions for two ground reaction force magnitudes (V-GRF_{min1} and V-GRF_{pk2}) and the peak plantarflexor ankle torque (T_{Ank3}). No significant difference between conditions was found for gait speed ($p = 0.139$, Table 2), but there was a trend that the PPAFO condition was slower compared to the SHOES (-8% change) and AFO (-8.1% change) conditions. There were no significant differences in ankle angle position parameters due to footwear condition ($p = 0.217$).

Modest changes were observed in the general behavior of the ground reaction forces in the PPAFO condition (Figure 6, Figure 7, and Table 2). Two amplitude parameters (V-GRF_{min1} and V-GRF_{pk2}) in the vertical ground reaction forces had significant changes due to footwear. The mid-stance minima was significantly increased (V-GRF_{min1}, $p = 0.048$) in the PPAFO condition compared to the AFO condition (5.5% change). The change in timing of this minima was borderline significant (V-GRF_{tmin1}, $p = 0.050$) with the PPAFO condition occurring earlier in the gait cycle compared to both the SHOES and AFO conditions. A significant change was observed for the propulsive vertical ground reaction force ($V\text{-}GRF_{pk2}$, $p = 0.026$). $V\text{-}GRF_{pk2}$ was significantly greater in the PPAFO condition than the SHOES condition (3% change). The average timing of the propulsive ground reaction peak in the anterior-posterior direction ($AP\text{-}GRF_{pk2}$, Figure 7) appeared to be earlier in the PPAFO condition, but no significant differences between conditions were found ($p = 0.489$).

The general behavior of the ankle torque was similar across conditions with one significant difference (Figure 6, Figure 7, and Table 2). A significant decrease in peak plantarflexor torque (T_{Ank3}, $p = 0.008$) was observed in the PPAFO condition compared to the SHOES condition (-16.8% change). Changes in peak dorsiflexor torque (T_{Ank2}) were modestly larger in the AFO condition compared to the SHOES and PPAFO conditions, but not statistically significant ($p = 0.114$).

2.5 Discussion

The objective of this study was to evaluate the changes in gait kinetics and kinematics in people with MS associated with the implementation of a powered ankle-foot orthosis compared to walking in a prescribed AFO or shoes. We hypothesized that peak vertical and anterior-posterior propulsive forces, plantarflexor ankle torque, and plantarflexion angle during terminal stance/pre-swing could be increased with the PPAFO compared to passive AFOs or walking shoes. Overall, the results of this study demonstrate that using the PPAFO for gait assistance can result in changes from baseline walking in shoes. However, two of the three significant changes associated with the PPAFO were not different from the AFO condition, indicating that the two devices did not greatly differ in modulating gait kinetics. Finally, no significant differences were found in the ankle angle parameters, probably due to the low sample size for those data (n = 3). Results of this study suggests that the PPAFO may be able to modulate the kinetics of gait in people with MS.

Application of a powered ankle-foot orthosis in people with MS induced modest changes in vertical ground reaction forces towards normative behaviors. Typically, people with lower levels of MS disability have decreased propulsive and breaking forces as well as vertical ground reaction force peaks (Benedetti et al. 1999; Kelleher et al. 2010). The PPAFO increased the vertical propulsive force (V-GRF_{pk2}) compared to baseline walking in shoes, suggesting a possible restoration of normal behavior (Figure 6, Figure 7, Table 2). This change is consistent with previous use of the PPAFO in a person with cauda equina syndrome who was unable to sufficiently generate plantarflexor torque (Shorter et al. 2011a). However, there was no concomitant increase in the propulsive anterior-posterior ground reaction forces in the current study (Figure 7, Table 2), which are also decreased in people with MS (Benedetti et al. 1999; Kelleher et al. 2010). A similar lack of increased anterior-posterior force with applied plantarflexor torque on the impaired limb of patients post-stroke has been observed (Takahashi et al. 2015). The reason for the lack of increased anterior-posterior force in these scenarios is unclear. Further investigation into the changes in ground reaction forces associated with increased plantarflexor torque from an external device is needed.

The only difference between the AFO and PPAFO conditions was the mid-stance minima in the vertical ground reaction forces was greater in the PPAFO condition ($vGRF_{min1}$, Figure 6, Figure 7, Table 2). This increase is a possible indication of slower gait speed (Neptune and Sasaki 2005). Although it was not a statistically significant change, participants walked slower with the PPAFO compared to SHOES and AFO in the current study. During the longer six-minute walk test (6MW) portion of the larger study, stride velocity in the PPAFO condition was found to be significantly slower than the SHOES and AFO conditions (Boes et al. 2015).

Interestingly, with an increase in vertical propulsive force, a concomitant increase in peak plantarflexor torque (T_{Ank3}) was not observed with the PPAFO. Instead, the peak plantarflexor torque was significantly smaller in the PPAFO condition compared to the SHOES condition (Figure 6, Figure 7, and Table 2). This change in plantarflexor torque could simply be due to slower walking speed in the PPAFO condition. Previous studies have shown that elderly adults have decreased plantarflexor torque at slower gait speeds (Kerrigan et al. 1998; DeVita and Hortobagyi 2000). The PPAFO was expected to be providing 10 Nm of torque during terminal stance/pre swing when peak plantarflexor torque should be achieved. Given the lack of increased plantarflexor torque, the participants could have re-optimized their contribution of plantarflexor torque while wearing the PPAFO. Similar adaptations to a powered ankle foot orthosis were observed in people post-stroke (Takahashi et al. 2015). In that study, the overall peak plantarflexor torque (i.e., the user’s contribution plus the device) was increased with the powered ankle-foot orthosis compared to baseline walking without the device. Further, gait speed was constrained in that study across conditions because the data were collected on a treadmill. Thus, it is possible that if gait speed was constrained in our study (e.g., with a treadmill), then the plantarflexor torque might have also increased with the PPAFO in our participants as well.

Different aspects of the form and function of the PPAFO have to be considered to improve the efficacy of the device. The current controller is based on "on-off" timings and the actuation is either all on or all off (i.e., "bang-bang" control). The nature of this actuation scheme makes it difficult to ramp of the ankle torque as it naturally does in the human body. Furthermore, if the actuation is turned on too early or late, it could potentially inhibit the desired torque output at the ankle. Future investigations should consider testing different torque and timing sequence and its effect on the kinematics and kinetics of gait (Islam and Hsiao Wecksler 2016). From the device design perspective, actuation systems that utilize proportional control of the actuator could enable a more natural ankle torque profile. Additionally, general factors such as improving the fit and reducing the weight of the device could improve the function of the PPAFO (Barnett et al. 1993).

Although significant effects were found, the results of this study should be interpreted knowing there were a few limitations. First, the reduced sample size makes it difficult to draw any conclusions of how the device could modulate gait kinetics in all people with MS. For example, the participants in this study were limited to moderate levels of MS, so very little can be interpreted about gait kinetics for people with severe levels of MS. Robotic treadmill training was able to improve gait speed in people with severe levels of MS (EDSS ≥ 6.0) (Beer et al. 2008; Lo and Triche 2008), so it is conceivable that a powered ankle foot orthosis that also provides physical assistance during the gait cycle could be beneficial for these individuals. However, it is also possible that a null result could be observed. Results from participants with severe MS (EDSS ≥ 6.0) had less of a change in gait speed and metabolic cost of walking with the PPAFO compared to the other conditions in the 6MWs of this study (Boes et al. 2015). Something else to consider is we do not know if there were even greater benefits to wearing the PPAFO over the course of multiple days or longer training sessions. In this study, the controller was trained while the participants first wore the device for 3 periods of ~10 steps each. During this time, the participants may have still been getting used to the device, so the controller could have been tuned to them walking with a slower, more conservative gait. With more training, the participants could have become more comfortable with the device and would not have had to resort to a conservative gait strategy. Future kinetic studies should focus on an increased sample size across multiple levels of MS disability with additional training.

2.6 Conclusions

Results of this study suggest that positive changes in kinematics and kinetics of gait can be induced by a powered ankle-foot orthosis in people with moderate of MS severity. Further investigation is needed into properly timing and magnitude of assistance to improve the efficacy of the PPAFO in modulating gait impairments of people with MS.

Figure 2: The portable powered ankle-foot orthosis (PPAFO). Photo is courtesy of Professor Cliff Shin in the Department of Industrial Design at the University of Illinois at Urbana-Champaign.

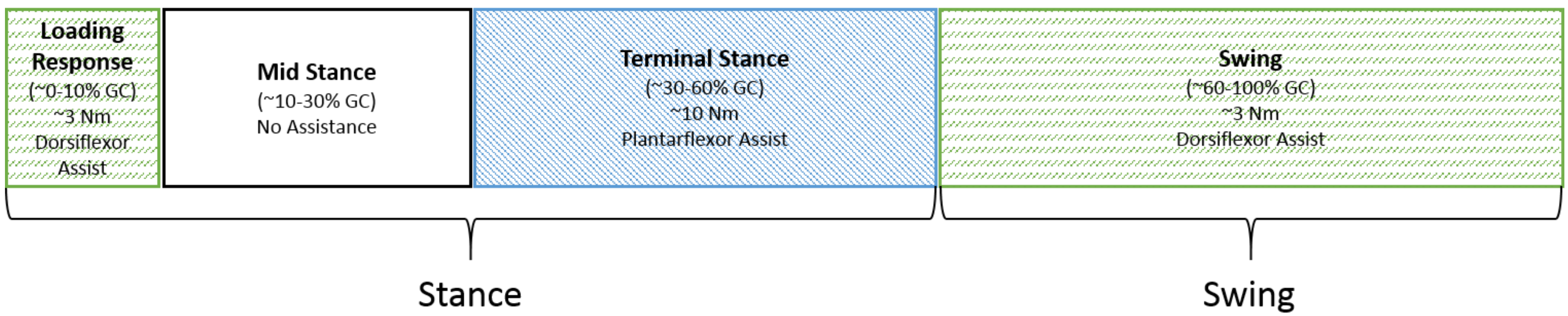

Figure 3: PPAFO actuation timings and magnitudes during the gait cycle (Perry 1992).

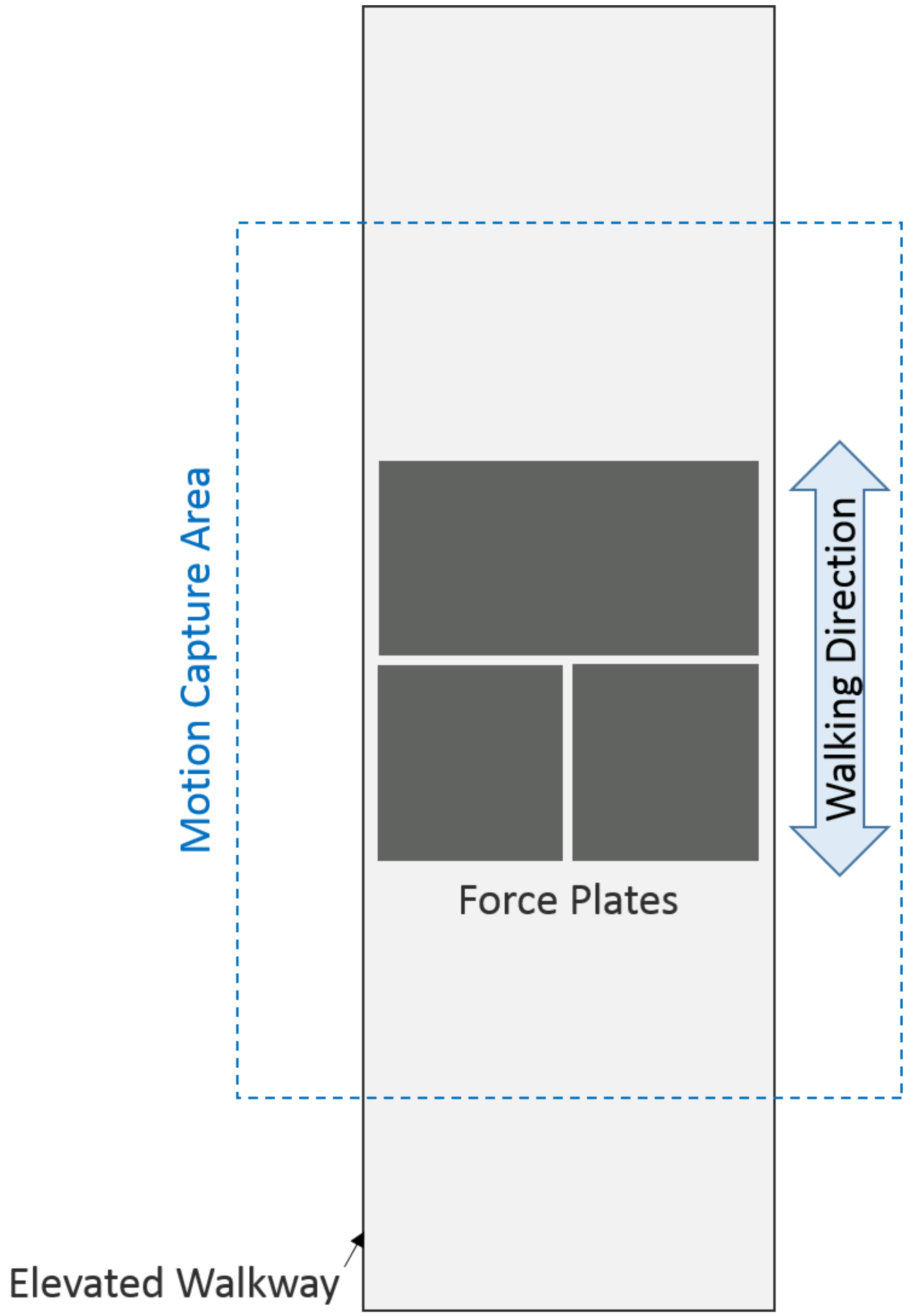

Figure 4: Data collection configuration of force plates with in the elevated walkway. The motion capture space is outlined in the blue dotted line.

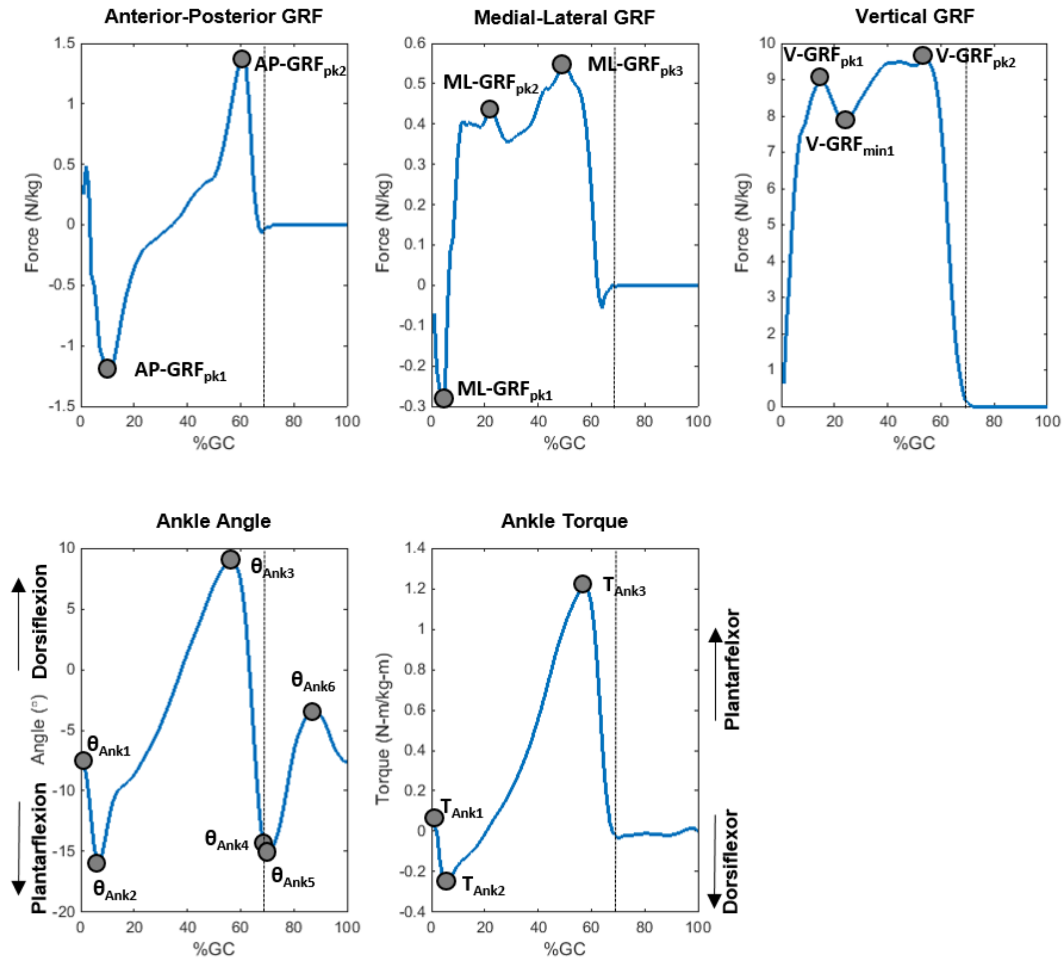

Figure 5: Data analysis parameters for ground reaction forces, ankle angle, and ankle torque. The vertical dashed line in each plot indicates the time of toe-off.

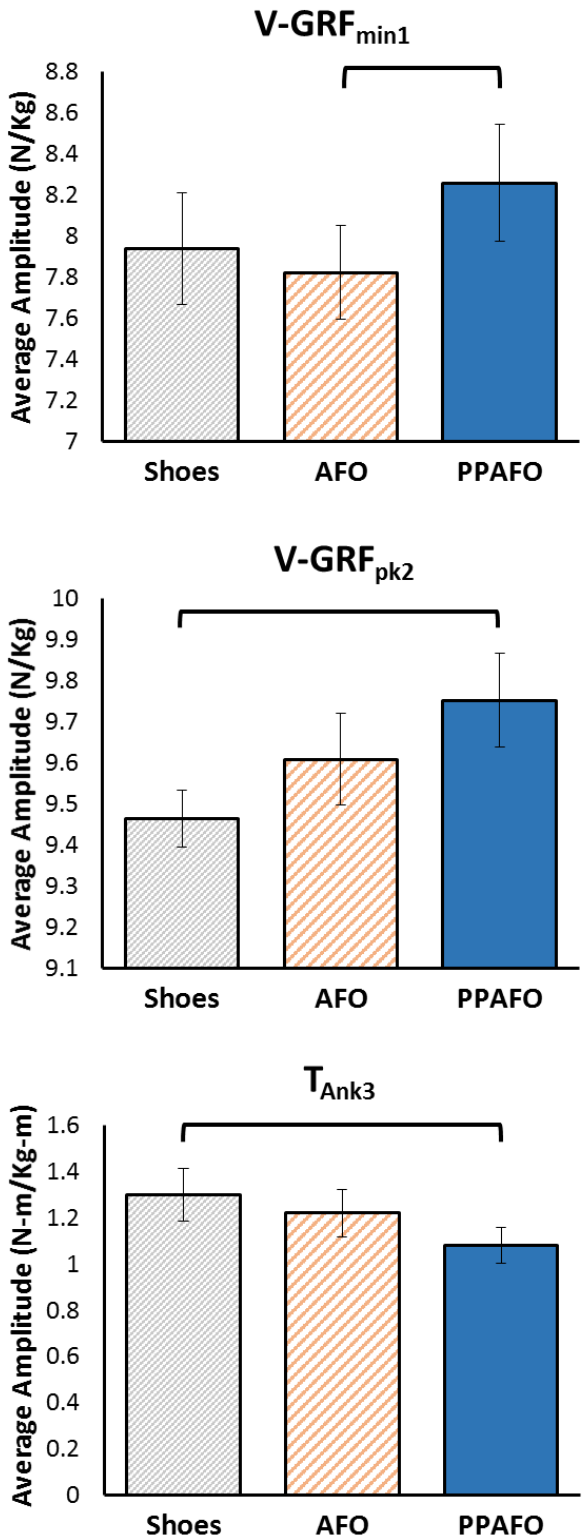

Figure 6: Average ground reaction force and torque parameters (± SEM) where significant differences were found between footwear conditions. The horizontal bar indicates a significant difference between conditions ($p < 0.05$).

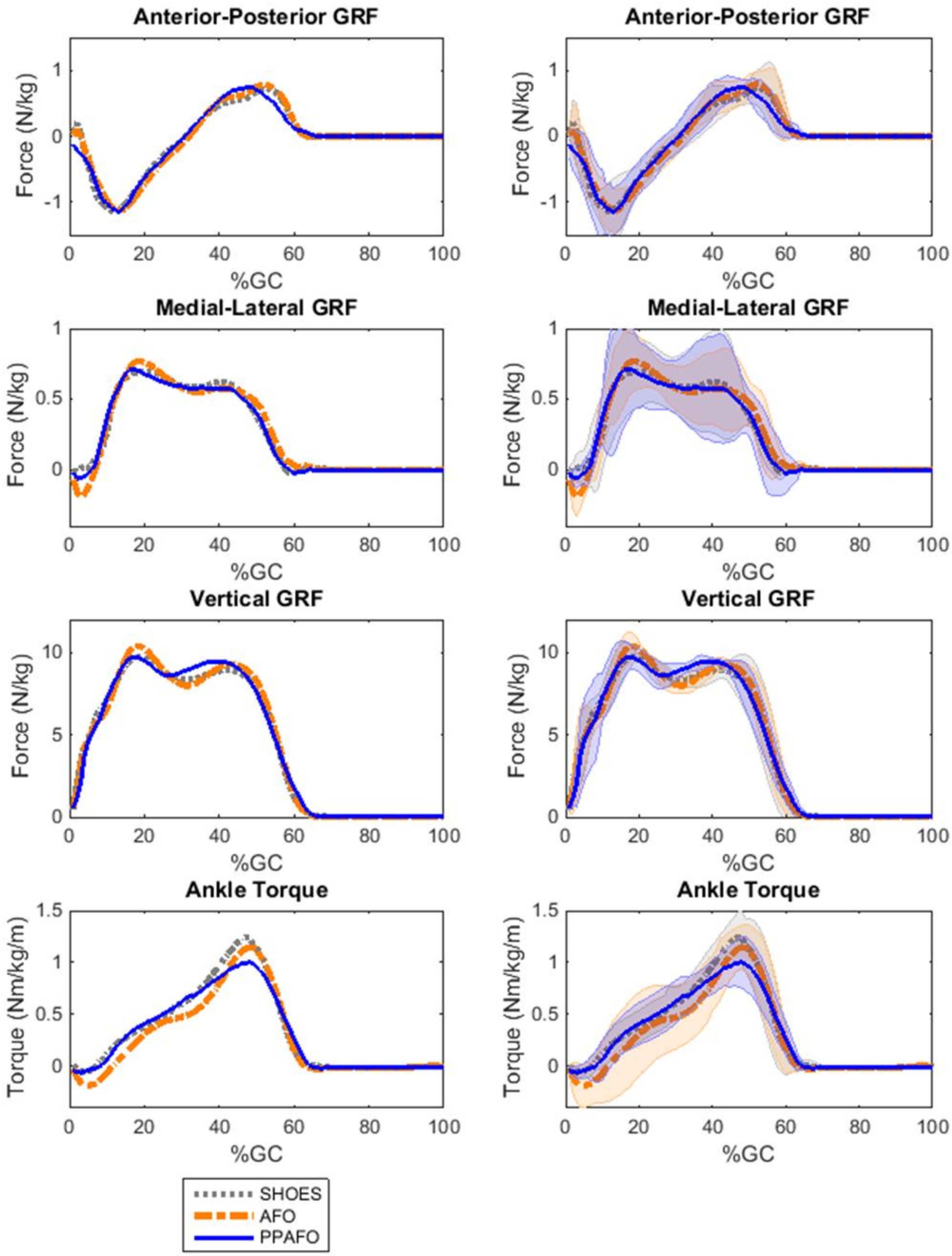

Figure 7: Average behaviors of ground reaction forces and ankle torque across conditions. The data are displayed as average ± standard deviation in the right panels.

Table 1: Demographics of all participants with reasons for exclusion from the overall analysis (assistive device required or no "clean" heel strikes) or ankle angle analysis (did not have a trial with all ankle angle parameters) highlighted.

ID	Walking Aid	Age	Gender	Weight (kg)	Height (cm)	Shoe Size (US)	AFO Leg	EDSS
001	Cane	45	F	60.6	163.4	W9.5	L	6
002	Cane	44	F	69.2	167.0	W8	R	6
003	None	51	F	117.4	171.7	W9	R	4.5
004	Walker	58	F	65.8	151.4	W6	R	6
005	AFO kept on left leg throughout	56	M	81.0	180.0	M9	R	3.5
006	Cane	53	F	45.5	166.8	W9	L	6.5
007	Walker	55	F	139.6	172.9	W11	R	6
009	Walker	62	M	109.6	172.4	M10.5	L	6.0
010	Walker	60	F	116.2	174.0	M10.5	R	6.5
011	None	59	F	68.6	170.7	W9.5	L	3.5
012	None	57	F	65.0	160.9	W8.5	L	4
013	None	51	F	65.7	176.2	W9	L	4
015	Cane	51	M	96.1	185.6	M11.5	L	6
016	None	53	F	47.0	160.4	W7	R	5.5
018	None	59	F	86.4	167.0	W8	L	4
019	None	60	M	101.4	179.3	M12	R	4

Did not collect ground reaction force data

Assistive device needed

Did not have at least 1 "clean" heel strike trial in one (or more) conditions

Did not have at least 1 trial with all ankle angle parameters in one (or more) conditions

Table 2: Average ground reaction force, ankle torque, and gait speed parameters across footwear conditions (± SD) with p-values from the univariate ANOVAs. Significant p-values are in bold ($p < 0.05$).

	Shoes	AFO	PPAFO	*p*-value
Anterior-Posterior GRF				
$AP\text{-}GRF_{pk1}$ (N/Kg)	-1.27 ± 0.36	-1.12 ± 0.35	-1.40 ± 0.34	0.186
$AP\text{-}GRF_{tpk1}$ (% GC)	10.2 ± 2.8	13.5 ± 1.5	11.2 ± 2.6	0.113
$AP\text{-}GRF_{pk2}$ (N/Kg)	0.97 ± 0.13	0.97 ± 0.18	0.87 ± 0.21	0.437
$AP\text{-}GRF_{tpk2}$ (% GC)	50.8 ± 7.0	49.0 ± 6.1	47.4 ± 4.2	0.489
Medial-Lateral GRF				
	Shoes	**AFO**	**PPAFO**	***p*-value**
$ML\text{-}GRF_{pk1}$ (N/Kg)	-0.19 ± 0.12	-0.32 ± 0.14	-0.18 ± 0.09	-
$ML\text{-}GRF_{tpk2}$ (% GC)	4.6 ± 2.3	3.8 ± 1.7	3.2 ± 2.1	-
$ML\text{-}GRF_{pk2}$ (N/Kg)	0.80 ± 0.32	0.78 ± 0.13	0.83 ± 0.29	-
$ML\text{-}GRF_{tpk2}$ (% GC)	19.1 ± 2.5	21.8 ± 2.8	19.3 ± 6.9	-
$ML\text{-}GRF_{pk3}$ (N/Kg)	0.69 ± 0.36	0.66 ± 0.19	0.687 ± 0.28	-
$ML\text{-}GRF_{tpk3}$ (% GC)	41.9 ± 4.4	43.5 ± 6.9	38.8 ± 8.8	-
Vertical GRF				
	Shoes	**AFO**	**PPAFO**	***p*-value**
$V\text{-}GRF_{pk1}$(N/Kg)	10.21 ± 0.66	10.55 ± 0.66	10.35 ± 0.55	0.264
$V\text{-}GRF_{tpk1}$ (% GC)	18.9 ± 2.7	19.2 ± 2.3	18.1 ± 3.0	0.804
$V\text{-}GRF_{min1}$ (N/Kg)	7.94 ± 0.61	7.82 ± 0.51	8.26 ± 0.64	**0.048**
$V\text{-}GRF_{tmin1}$ (% GC)	32.2 ± 3.8	32.6 ± 2.3	27.8 ± 4.4	0.050
$V\text{-}GRF_{pk2}$ (N/Kg)	9.46 ± 0.15	9.61 ± 0.25	9.75 ± 0.26	**0.026**
$V\text{-}GRF_{tpk2}$ (% GC)	43.4 ± 4.6	44.2 ± 4.7	40.8 ± 4.4	0.319
Ankle Torque				
	Shoes	**AFO**	**PPAFO**	***p*-value**
T_{Ank1} (N-m/kg-m)	0.00 ± 0.03	-0.03 ± 0.05	-0.03 ± 0.06	0.229
T_{Ank2} (N-m/kg-m)	-0.08 ± 0.07	-0.20 ± 0.19	-0.12 ± 0.06	0.114
T_{tAnk2} (% GC)	3.3 ± 1.3	5.8 ± 2.4	5.8 ± 3.1	0.195
T_{Ank3} (N-m/kg-m)	1.30 ± 0.26	1.22 ± 0.23	1.08 ± 0.17	**0.008**
T_{tAnk3} (% GC)	48.2 ± 2.6	49.1 ± 2.5	46.9 ± 2.8	0.368
Gait Speed				
	Shoes	**AFO**	**PPAFO**	***p*-value**
GS (m/s)	0.98 ± 0.11	0.98 ± 0.07	0.90 ± 0.08	0.139

Table 3: Average ankle angle of 3 participants that did not have missing motion data across footwear conditions (+/- SD). No main effect of the MANOVA for these parameters was found, so no univariate p-values are provided.

Ankle Angle (3 Participants)			
	Shoes	**AFO**	**PPAFO**
θ_{Ank1} (°)	-4.1 ± 4.4	-3.2 ± 3.0	-3.4 ± 1.4
θ_{Ank2} (°)	-11.3 ± 4.0	-8.6 ± 4.4	-9.7 ± 6.3
θ_{tAnk2} (% GC)	5.6 ± 0.4	5.3 ± 1.2	5.0 ± 2.0
θ_{Ank3} (°)	13.4 ± 0.7	12.0 ± 2.5	8.4 ± 3.7
θ_{tAnk3} (% GC)	51.8 ± 1.3	51.3 ± 1.6	50.3 ± 1.2
θ_{Ank4} (°)	-8.6 ± 5.6	-4.1 ± 4.8	-6.1 ± 2.8
θ_{tAnk4} (% GC)	64.7 ± 2.7	63.9 ± 2.5	63.8 ± 0.7
θ_{Ank5} (°)	-11.6 ± 7.0	-6.3 ± 5.8	-7.3 ± 3.4
θ_{tAnk5} (% GC)	66.8 ± 2.4	66.0 ± 2.0	65.1 ± 1.3
θ_{Ank6} (°)	-0.2 ± 5.3	0.6 ± 4.3	3.3 ± 1.4
θ_{tAnk6} (% GC)	85.4 ± 5.1	85.4 ± 7.6	86.8 ± 5.6

3 Conclusions and Future Work

In this thesis, we utilized the PPAFO to investigate the effectiveness of a powered ankle-foot orthoses for changing the kinematic and kinetics of gait compared to standard AFOs in people with MS. We hypothesized that due to active motion control in both dorsiflexion and plantarflexion that propulsive forces and torque would be increased compared to a standard prescribed orthoses or normal walking shoes. Significant increases in the vertical ground reaction force magnitude at mid-stance (compared to the AFO condition) and during terminal stance (compared to the SHOES condition) were observed in the PPAFO condition (Figure 6, Table 2). However the propulsive vertical ground reaction force was not significantly increased when compared to the AFO condition (Figure 7, Table 2). Furthermore, the peak plantarflexor torque was significantly reduced in the PPAFO condition compared to SHOES, but no differences existed between the PPAFO and AFO conditions (Figure 6, Table 2). These findings suggest that changes in gait kinetics are possible with the PPAFO. Additional investigation is needed to determine if improved gait kinematic and kinetics can be achieved with a powered ankle-foot orthosis compared to a prescribed AFO for people with MS.

3.1 Future Work

The main objective of using a powered ankle-foot orthosis, or any gait intervention, for people with MS will be keeping them ambulatory as long as possible during the disease course. Several key factors have to be investigated in order to achieve this goal with a powered ankle-foot orthosis. These include improved PPAFO control for providing assistance when it is needed during the gait cycle, optimizing the training period needed with the PPAFO to induce the desired changes in gait characteristics, and applying the device in people with MS at multiple levels of MS severity.

Optimization of the PPAFO controller is needed to induce the desired biomechanics and maximize the efficacy of the PPAFO in people with MS. In this study, the PPAFO was able to restore more normative propulsive vertical ground reaction forces. However, no change in the anterior-posterior propulsive force or ankle angle were observed. The decrease in plantarflexor torque may have been caused by the plantarflexor torque from the device being activated too soon during mid-stance. Machine learning techniques and control techniques are currently being utilized to address the problem of torque timing in our lab (Islam and Hsiao Wecksler 2016) and further investigation is still needed. Moreover, the application of proportional torque control (i.e., tracking the torque output of the user and device), instead of the "bang-bang" kinematic control of our current controller, could be beneficial to mimic the natural torque generation at the ankle. If the controller is tuned to encourage more normative behaviors, improved gait kinematics, ground reaction forces, and ankle torque may be achieved in people with MS, which would could also result in faster gait speed.

In addition to the actuation of the PPAFO, the amount of training necessary for the PPAFO to have positive changes on the kinetics and kinematics of gait is still an open question. In this study, the minimization algorithm for the controller was run when the participants first wore the device for 3 periods of ~10 steps each. During this time, the participants may have still been getting used to the device, so the controller may have been tuned to them walking with a slower, more conservative gait. Currently, it is not clear if extending this training period, or allowing the participants to retrain the device after familiarization (i.e., on the same day, or over the course of multiple days), would have resulted in improved biomechanics with the PPAFO. Optimizing the training period along with PPAFO actuation could result in an even larger benefit for the PPAFO.

Finally, it is unclear if similar changes in kinematics and kinetics would be observed in people with severe levels of MS impairment (EDSS ≥ 6.0). Previous robotic body weight support treadmill training would suggest that robotic assistance could have a positive effect on people with higher levels of MS disease severity (Beer et al. 2008; Lo and Triche 2008). One thing to consider is people with higher levels of MS severity typically walk even slower than people with low levels of MS disease severity (Kelleher et al. 2010). Thus, if the reduced peak plantarflexor torque and a lack of peak sagittal shear force during terminal stance observed in this study were because of slower gait speed, the same type of results may be expected in people with higher MS disability. At the same time, a null result could be observed. As part of this study, the results from the 6MWs of participants with higher MS impairment (EDSS ≥ 6.0) suggest there was no change in gait speed or metabolic cost of walking with the PPAFO compared to the AFO and SHOES conditions (Boes et al. 2015). Further investigation is needed, but if the PPAFO is properly tuned to encourage more normative biomechanics, it could promote independence and increase mobility for people with severe levels of MS impairment.

4 References

Anderson, D. W., J. H. Ellenberg, C. M. Leventhal, S. C. Reingold, M. Rodriguez and D. H. Silberberg (1992). "Revised estimate of the prevalence of multiple sclerosis in the United States." Ann Neurol **31**(3): 333-336.

Barnett, S. L., A. M. Bagley and H. B. Skinner (1993). "Ankle weight effect on gait: orthotic implications." Orthopedics **16**(10): 1127-1131.

Beer, S., B. Aschbacher, D. Manoglou, E. Gamper, J. Kool and J. Kesselring (2008). "Robot-assisted gait training in multiple sclerosis: a pilot randomized trial." Mult Scler **14**(2): 231-236.

Benedetti, M. G., R. Piperno, L. Simoncini, P. Bonato, A. Tonini and S. Giannini (1999). "Gait abnormalities in minimally impaired multiple sclerosis patients." Mult Scler **5**(5): 363-368.

Benedict, R. H., R. Holtzer, R. W. Motl, F. W. Foley, S. Kaur, D. Hojnacki and B. Weinstock-Guttman (2011). "Upper and lower extremity motor function and cognitive impairment in multiple sclerosis." J Int Neuropsychol Soc **17**(4): 643-653.

Blaya, J. A. and H. Herr (2004). "Adaptive control of a variable-impedance ankle-foot orthosis to assist drop-foot gait." IEEE Trans Neural Syst Rehabil Eng **12**(1): 24-31.

Boes, M. K., R. E. Klaren, R. M. Kesler, M. Islam, Y. Learmonth, M. N. Petrucci, R. W. Motl and E. T. Hsiao Wecksler (2015). Spatiotemporal and Metabolic Impacts on Gait of a Powered Ankle Exoskeleton in Persons with Multiple Sclerosis. 39th Annual meeting of the American Society of Biomechanics, Columbus, OH.

Bregman, D. J., V. De Groot, P. Van Diggele, H. Meulman, H. Houdijk and J. Harlaar (2010). "Polypropylene ankle foot orthoses to overcome drop-foot gait in central neurological patients: a mechanical and functional evaluation." Prosthet Orthot Int **34**(3): 293-304.

Bregman, D. J., J. Harlaar, C. G. Meskers and V. de Groot (2012). "Spring-like Ankle Foot Orthoses reduce the energy cost of walking by taking over ankle work." Gait Posture **35**(1): 148-153.

Cattaneo, D., F. Marazzini, A. Crippa and R. Cardini (2002). "Do static or dynamic AFOs improve balance?" Clin Rehabil **16**(8): 894-899.

Confavreux, C. and S. Vukusic (2006). "Natural history of multiple sclerosis: a unifying concept." Brain **129**(Pt 3): 606-616.

DeVita, P. and T. Hortobagyi (2000). "Age causes a redistribution of joint torques and powers during gait." J Appl Physiol (1985) **88**(5): 1804-1811.

Fay, B. T. and M. L. Boninger (2002). "The science behind mobility devices for individuals with multiple sclerosis." Med Eng Phys **24**(6): 375-383.

Ferris, D. P., J. M. Czerniecki and B. Hannaford (2005). "An ankle-foot orthosis powered by artificial pneumatic muscles." J Appl Biomech **21**(2): 189-197.

Ferris, D. P., K. E. Gordon, G. S. Sawicki and A. Peethambaran (2006). "An improved powered ankle-foot orthosis using proportional myoelectric control." Gait Posture **23**(4): 425-428.

Finlayson, M., L. Guglielmello and K. Liefer (2001). "Describing and predicting the possession of assistive devices among persons with multiple sclerosis." Am J Occup Ther **55**(5): 545-551.

Freal, J. E., G. H. Kraft and J. K. Coryell (1984). "Symptomatic fatigue in multiple sclerosis." Arch Phys Med Rehabil **65**(3): 135-138.

Giesser, B., J. Beres-Jones, A. Budovitch, E. Herlihy and S. Harkema (2007). "Locomotor training using body weight support on a treadmill improves mobility in persons with multiple sclerosis: a pilot study." Mult Scler **13**(2): 224-231.

Goldman, M. D., R. A. Marrie and J. A. Cohen (2008). "Evaluation of the six-minute walk in multiple sclerosis subjects and healthy controls." Mult Scler **14**(3): 383-390.

Heesen, C., J. Bohm, C. Reich, J. Kasper, M. Goebel and S. M. Gold (2008). "Patient perception of bodily functions in multiple sclerosis: gait and visual function are the most valuable." Mult Scler **14**(7): 988-991.

Hersh, C. and A. Rae-Grant (2012). "Extended-release dalfampridine in the management of multiple-sclerosis-related walking impairment." Ther Adv Neurol Disord **5**(4): 199-204.

Islam, M., M. T. Hagan and E. T. Hsiao Wecksler (2016). Gait State Estimation for Powered Ankle Orthosis using Modified Fraction Timing and Artificial Neural Network. Design of Medical Devices Conference. Minneapolis, MN.

Islam, M. and E. T. Hsiao Wecksler (2016). Developing a Classification Algorithm for Plantarflexor Actuation Timing of a Powered Ankle-Foot Orthosis. Design of Medical Devices Conference. Minneapolis, MN.

Kelleher, K. J., W. Spence, S. Solomonidis and D. Apatsidis (2010). "The characterisation of gait patterns of people with multiple sclerosis." Disabil Rehabil **32**(15): 1242-1250.

Kerrigan, D. C., M. K. Todd, U. Della Croce, L. A. Lipsitz and J. J. Collins (1998). "Biomechanical gait alterations independent of speed in the healthy elderly: evidence for specific limiting impairments." Arch Phys Med Rehabil **79**(3): 317-322.

Khan, O., T. Leist, T. Vollmer and S. Zamvil (2008). "Investigating multiple sclerosis: targeting therapeutic options." Int J MS Care **10**(Suppl 1): 5-20.

Larocca, N. G. (2011). "Impact of walking impairment in multiple sclerosis: perspectives of patients and care partners." Patient **4**(3): 189-201.

Li, D. Y., A. Becker, K. A. Shorter, T. Bretl and E. T. Hsiao-Wecksler (2011). "Estimating System State During Human Walking With a Powered Ankle-Foot Orthosis." IEEE-ASME T Mech **16**(5): 835-844.

Li, Y. D. and E. T. Hsiao-Wecksler (2013). Gait mode recognition and control for a portable-powered ankle-foot orthosis. 2013 IEEE International Conference on Rehabilitation Robotics (ICORR). Seattle, WA**:** 1-8.

Lo, A. C. and E. W. Triche (2008). "Improving gait in multiple sclerosis using robot-assisted, body weight supported treadmill training." Neurorehabil Neural Repair **22**(6): 661-671.

Lord, S. E., D. T. Wade and P. W. Halligan (1998). "A comparison of two physiotherapy treatment approaches to improve walking in multiple sclerosis: a pilot randomized controlled study." Clin Rehabil **12**(6): 477-486.

McLoughlin, J. V., C. J. Barr, B. Patritti, M. Crotty, S. R. Lord and D. L. Sturnieks (2015). "Fatigue induced changes to kinematic and kinetic gait parameters following six minutes of walking in people with multiple sclerosis." Disabil Rehabil: 1-9.

Mooney, L. M., E. J. Rouse and H. M. Herr (2014). "Autonomous exoskeleton reduces metabolic cost of human walking during load carriage." J Neuroeng Rehabil **11**: 80.

Motl, R. W. (2013). "Ambulation and multiple sclerosis." Phys Med Rehabil Clin N Am **24**(2): 325-336.

Motl, R. W., M. D. Goldman and R. H. Benedict (2010). "Walking impairment in patients with multiple sclerosis: exercise training as a treatment option." Neuropsychiatr Dis Treat **6**: 767-774.

Motl, R. W. and L. A. Pilutti (2012). "The benefits of exercise training in multiple sclerosis." Nat Rev Neurol **8**(9): 487-497.

Motl, R. W. and B. M. Sandroff (2015). "Benefits of Exercise Training in Multiple Sclerosis." Curr Neurol Neurosci Rep **15**(9): 62.

Neptune, R. R. and K. Sasaki (2005). "Ankle plantar flexor force production is an important determinant of the preferred walk-to-run transition speed." J Exp Biol **208**(Pt 5): 799-808.

Neubauer, B. C., J. Nath and W. K. Durfee (2014). Design of a Portable Hydraulic Ankle-Foot Orthosis. 2014 36th Annual International Conference of the Ieee Engineering in Medicine and Biology Society (EMBC): 1182-1185.

Nikitczuk, J., B. Weinberg and C. Mavroidis (2007). "Control of electro-rheological fluid based resistive torque elements for use in active rehabilitation devices." Smart Mater Struct **16**(2): 418-428.

Noseworthy, J. H., C. Lucchinetti, M. Rodriguez and B. G. Weinshenker (2000). "Multiple sclerosis." N Engl J Med **343**(13): 938-952.

Perry, J. (1992). Gait analysis : normal and pathological function. Thorofare, NJ, SLACK.

Pilutti, L. A., D. A. Lelli, J. E. Paulseth, M. Crome, S. Jiang, M. P. Rathbone and A. L. Hicks (2011). "Effects of 12 weeks of supported treadmill training on functional ability and quality of life in progressive multiple sclerosis: a pilot study." Arch Phys Med Rehabil **92**(1): 31-36.

Roy, A., H. I. Krebs, D. J. Williams, C. T. Bever, L. W. Forrester, R. M. Macko and N. Hogan (2009). "Robot-Aided Neurorehabilitation: A Novel Robot for Ankle Rehabilitation." Ieee T Robot **25**(3): 569-582.

Sawicki, G. S. and D. P. Ferris (2009). "A pneumatically powered knee-ankle-foot orthosis (KAFO) with myoelectric activation and inhibition." J Neuroeng Rehabil **6**: 23.

Schwartz, I., A. Sajin, E. Moreh, I. Fisher, M. Neeb, A. Forest, A. Vaknin-Dembinsky, D. Karusis and Z. Meiner (2012). "Robot-assisted gait training in multiple sclerosis patients: a randomized trial." Mult Scler **18**(6): 881-890.

Sehle, A., A. Mundermann, K. Starrost, S. Sailer, I. Becher, C. Dettmers and M. Vieten (2011). "Objective assessment of motor fatigue in Multiple Sclerosis using kinematic gait analysis: a pilot study." J Neuroeng Rehabil **8**: 59.

Sheffler, L. R., M. T. Hennessey, J. S. Knutson, G. G. Naples and J. Chae (2008). "Functional effect of an ankle foot orthosis on gait in multiple sclerosis: a pilot study." Am J Phys Med Rehabil **87**(1): 26-32.

Shorter, K. A., G. F. Kogler, E. Loth, W. K. Durfee and E. T. Hsiao-Wecksler (2011a). "A portable powered ankle-foot orthosis for rehabilitation." J Rehabil Res Dev **48**(4): 459-472.

Shorter, K. A., Y. F. Li, E. A. Morris, G. F. Kogler and E. T. Hsiao-Wecksler (2011b). Experimental Evaluation of a Portable Powered Ankle-Foot Orthosis. Ieee Eng Med Bio**:** 624-627.

Sosnoff, J. J., E. Gappmaier, A. Frame and R. W. Motl (2011). "Influence of spasticity on mobility and balance in persons with multiple sclerosis." J Neurol Phys Ther **35**(3): 129-132.

Sosnoff, J. J., B. M. Sandroff and R. W. Motl (2012). "Quantifying gait abnormalities in persons with multiple sclerosis with minimal disability." Gait Posture **36**(1): 154-156.

Souza, A., A. Kelleher, R. Cooper, R. A. Cooper, L. I. Iezzoni and D. M. Collins (2010). "Multiple sclerosis and mobility-related assistive technology: systematic review of literature." J Rehabil Res Dev **47**(3): 213-223.

Takahashi, K. Z., M. D. Lewek and G. S. Sawicki (2015). "A neuromechanics-based powered ankle exoskeleton to assist walking post-stroke: a feasibility study." J Neuroeng Rehabil **12**.

Verza, R., M. L. Carvalho, M. A. Battaglia and M. M. Uccelli (2006). "An interdisciplinary approach to evaluating the need for assistive technology reduces equipment abandonment." Mult Scler **12**(1): 88-93.

Wiles, C. M., R. G. Newcombe, K. J. Fuller, S. Shaw, J. Furnival-Doran, T. P. Pickersgill and A. Morgan (2001). "Controlled randomised crossover trial of the effects of physiotherapy on mobility in chronic multiple sclerosis." J Neurol Neurosurg Psychiatry **70**(2): 174-179.

Winter, D. A. (1984). "Kinematic and Kinetic Patterns in Human Gait - Variability and Compensating Effects." Hum Movement Sci **3**(1-2): 51-76.

Winter, D. A. (1989). "Biomechanics of Normal and Pathological Gait - Implications for Understanding Human Locomotor Control." J Motor Behav **21**(4): 337-355.

Wurdeman, S. R., J. M. Huisinga, M. Filipi and N. Stergiou (2011). "Multiple sclerosis affects the frequency content in the vertical ground reaction forces during walking." Clin Biomech (Bristol, Avon) **26**(2): 207-212.

Appendix A: Additional Ankle Angle Data

Two of the participants in this study were excluded from the ankle angle analysis because marker drop out in the shank segment (Table 1). The purpose of this Appendix is to provide the ankle angle data across all participants for reference. Note, in each participant figure, the separate colors delineate separate trials for that condition. Additionally, averages across all five participants and the three included in the statistical analysis are provided (Figure 13).

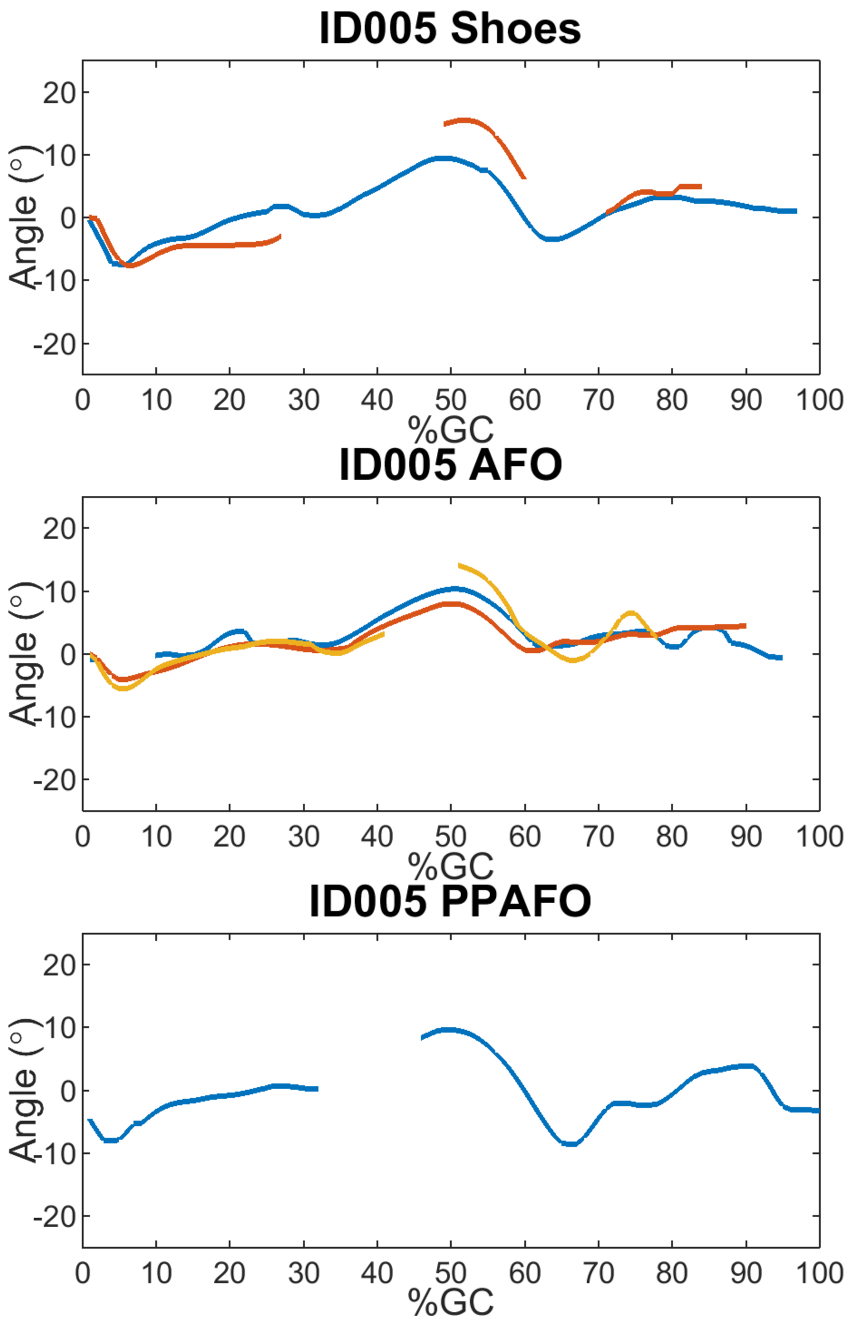

Figure 8: All ankle angle trials by condition for participant 005. Note, even though there are data missing in the middle of the PPAFO trial, all of the parameters were able to be identified.

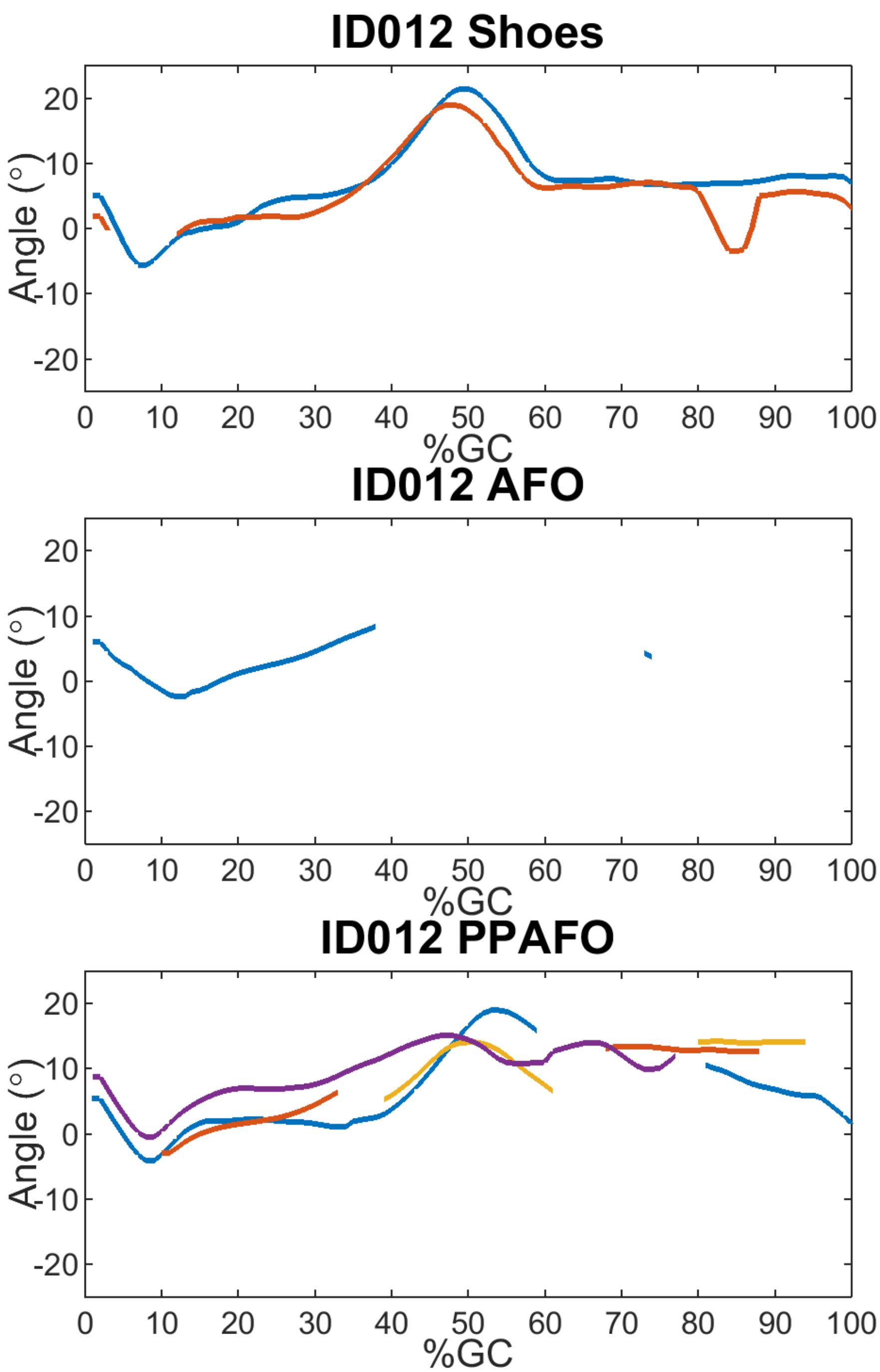

Figure 9: All ankle angle trials by condition for participant 012. Note there are not enough data in the AFO and PPAFO conditions to be able to identify every ankle angle parameter. Thus, this participant was not included in the statistical analysis.

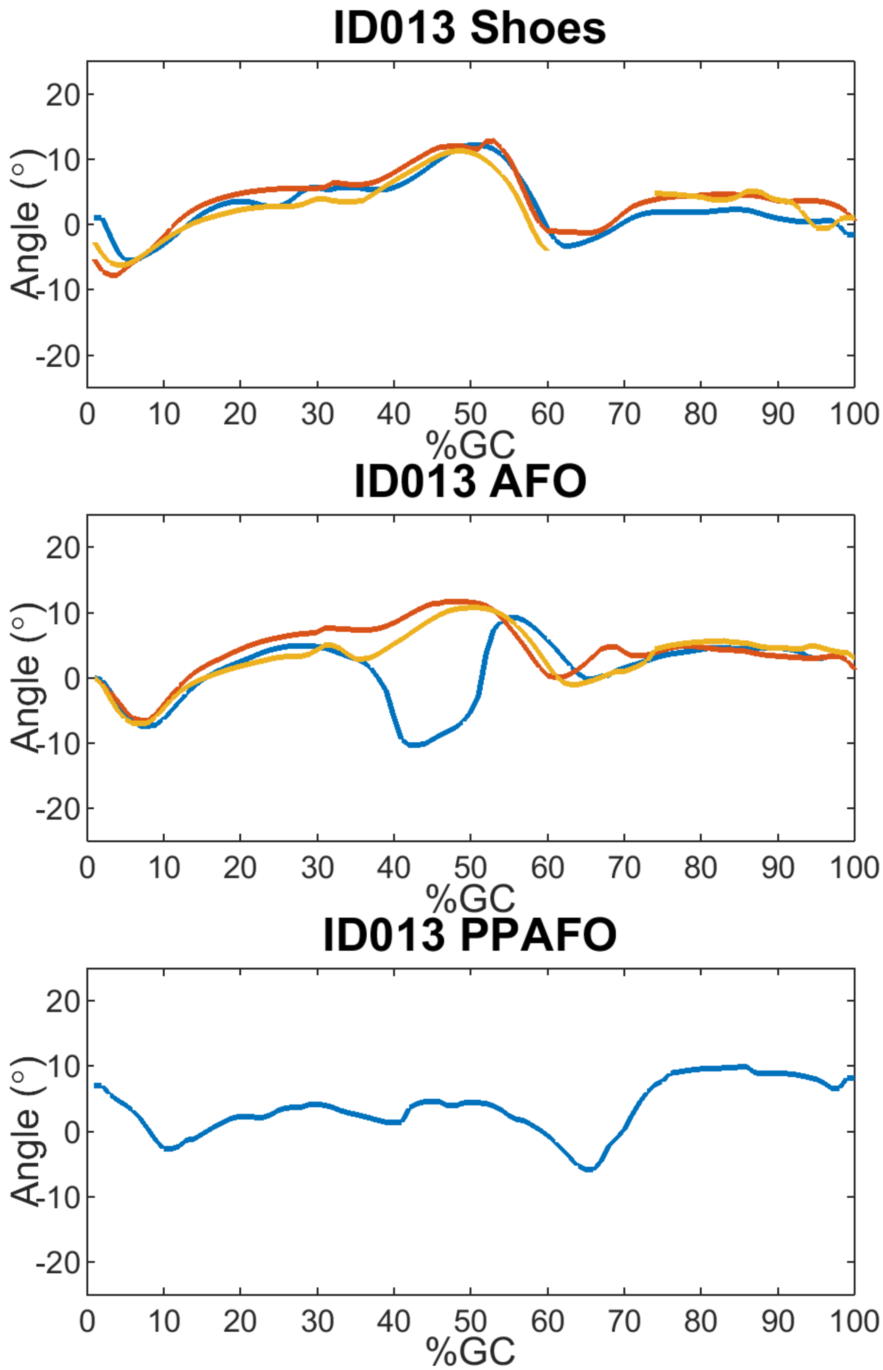

Figure 10: All ankle angle trials by condition for participant 013. Note, although there is continuous data during swing of the PPAFO trial, it was determined that the small "jump" in the data around 86% gait cycle was caused by motion markers dropping in and out of the frame. Thus, we could not include the max dorsiflexion angle during swing in the statistical analysis.

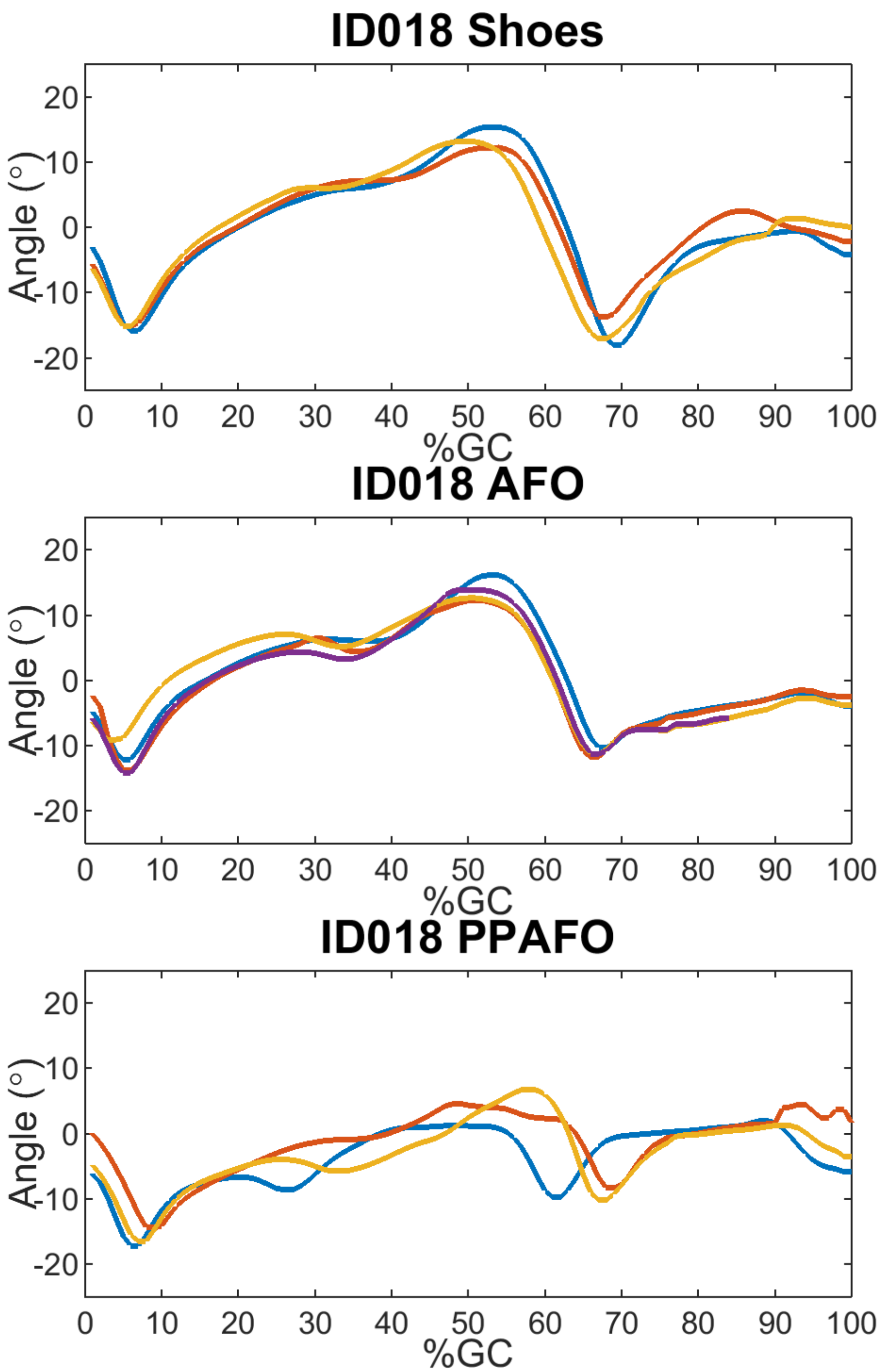

Figure 11: All ankle angle trials by condition for participant 018. Note, there were more than three trials in the AFO condition. Only the first three were included in the analysis.

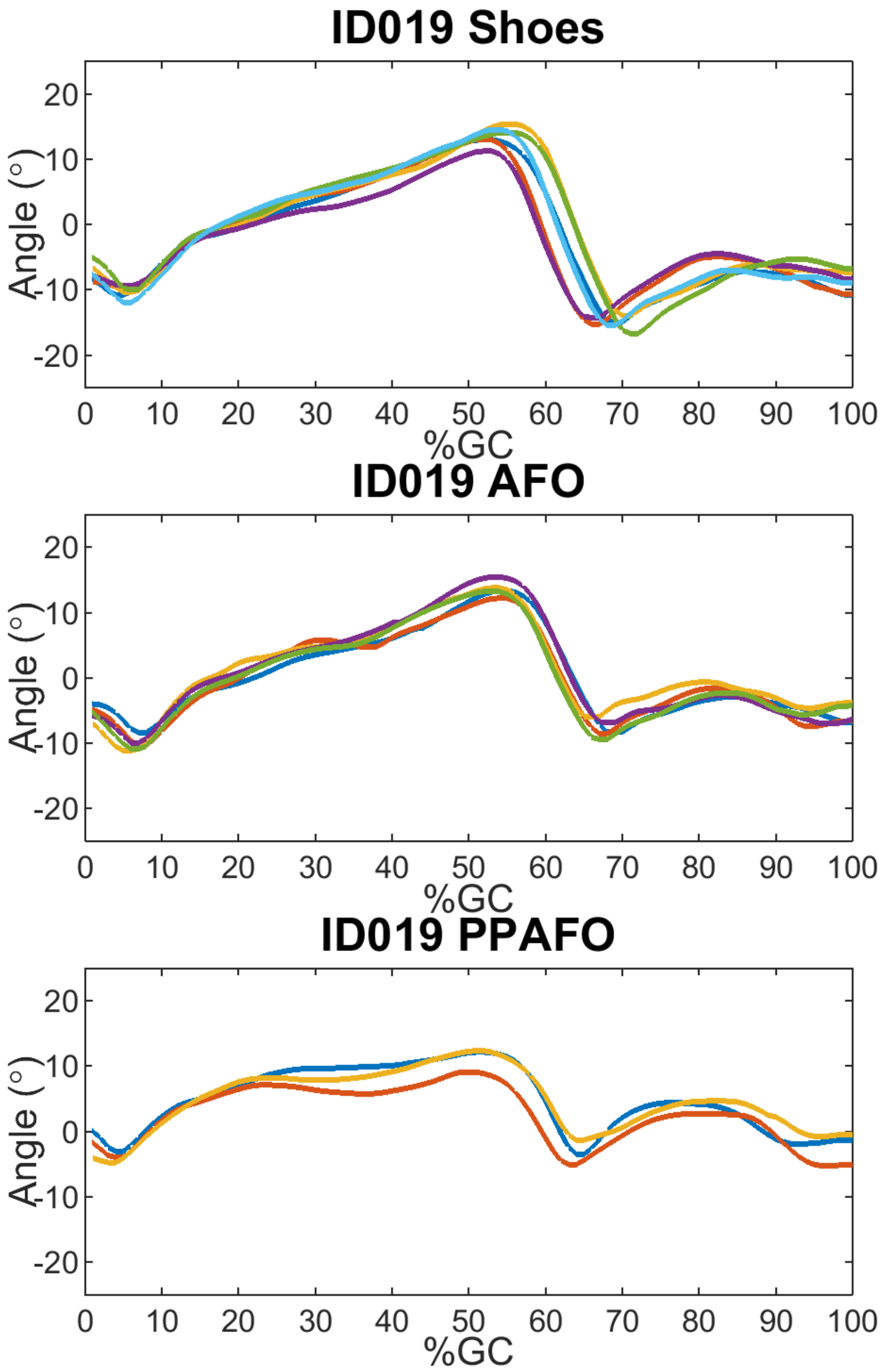

Figure 12: All ankle angle trials by condition for participant 018. Note, there were more than three trials in the SHOES and AFO conditions. Only the first three were included in the analysis.

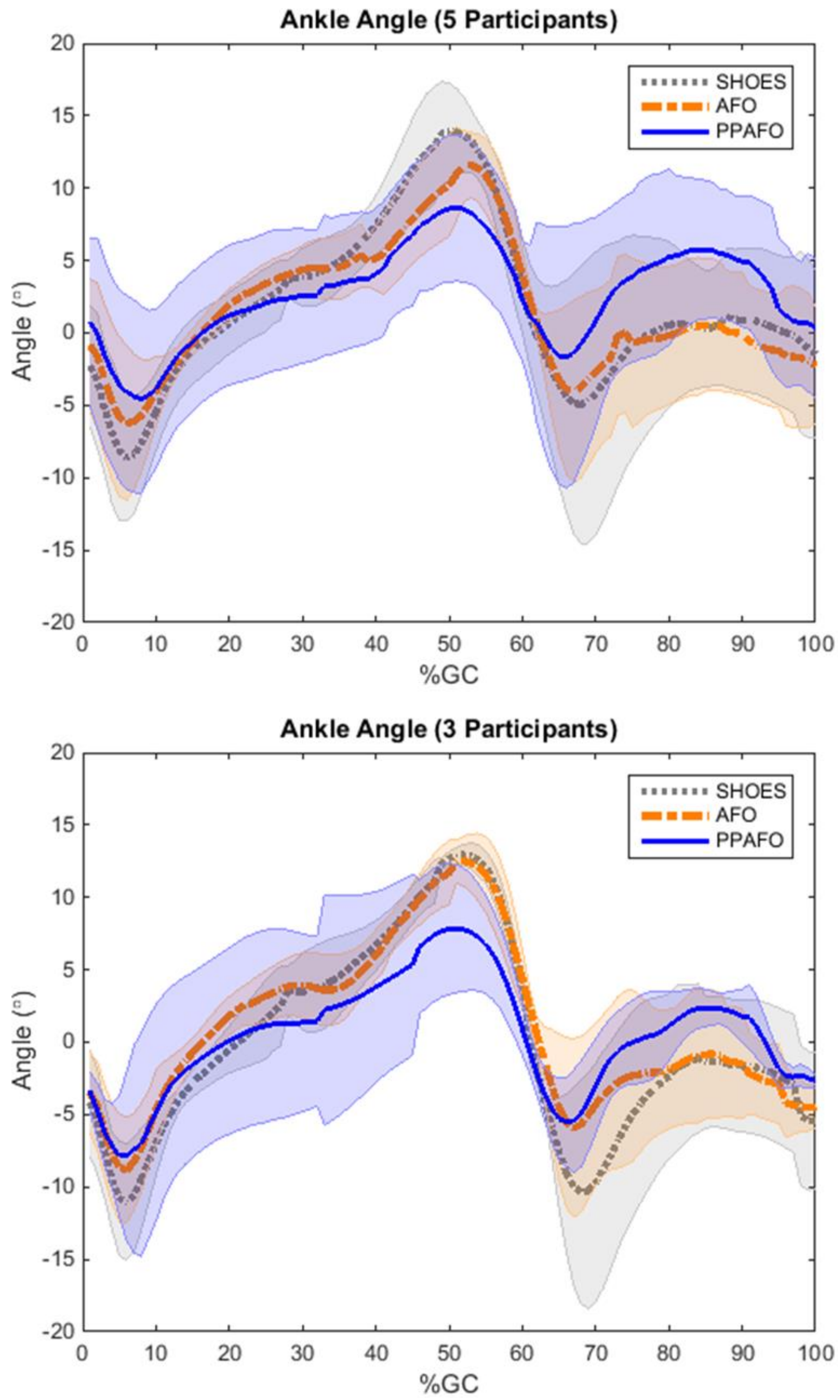

Figure 13: Average ankle angle behaviors across footwear conditions for all five participants (top) and the three participants (bottom) that did not have parameters lost due to marker dropout. Data are displayed as the average ± SD.

Appendix B: Analysis of Changing PPAFO Weight on Ankle Torque

In this study, the weight of the PPAFO was not included in the inverse dynamics calculations for ankle torque. To evaluate how much the added weight would affect the calculation of ankle torque, an increased weight of 0.2 kg was added to the model of the foot and the ankle torque was recalculated. Results demonstrate that if only the actuator was considered (0.474 kg), the difference in peak plantarflexor torque would have been between -0.008 and -0.012 (N-m/kg-m) (Table 5). It would be a safe assumption to consider the actuator weight alone because the foot shell is a similar weight to a standard walking shoe, and the shank shell and valves are not attached to the foot of the user. In the worst case, if the full weight of the PPAFO would have been included (1.176 kg, Table 4), a decrease in peak plantarflexor torque would be observed (between -0.018 to -0.022 (N-m/kg-m), Table 5). Ultimately, these difference in peak plantarflexor torque are minimal in comparison to the average ankle torque values in this study (1.08-1.30 (N-m/kg-m)). Furthermore, the significant decrease in peak plantarflexor torque in the PPAFO condition would have still been observed with these changes in foot weight.

Table 4: Weight of each component of the PPAFO.

Item	Weight
Actuator	0.474 kg
Foot Shell	0.410 kg
Shank Shell	0.246 kg
Valves	0.046 kg
Total	1.176 kg

Table 5: Differences in peak plantarflexor torque at specific increments of additional foot weight.

Additional Foot Weight	Difference at Peak Plantarflexor Torque
0.2 kg	-0.004 (N-m)/(kg-m)
0.4 kg	-0.008 (N-m)/(kg-m)
0.6 kg	-0.012 (N-m)/(kg-m)
0.8 kg	-0.016 (N-m)/(kg-m)
1.0 kg	-0.018 (N-m)/(kg-m)
1.2 kg	-0.022 (N-m)/(kg-m)