EVALUATION OF A PNEUMATIC ANKLE-FOOT ORTHOSIS: PORTABILITY AND FUNCTIONALITY

BY

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DISSERTATION

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Abstract

There are currently many challenges in creating portable human-assistive robotics and exoskeletons, although the need for robotic human assist continues to grow. These challenges span disciplines such as control, design, fuel and efficiency, user-interfaces, neuroscience, and kinesiology. Our lab has developed a pneumatically powered ankle-foot orthosis (PPAFO) to address some of these issues.

In this dissertation, we address the issue of availability of portable pneumatic power sources, and we evaluate the short-term kinematic and metabolic impact of a bilateral, bidirectional portable powered ankle-foot orthosis (PPAFO) in an able-bodied population during over-ground walking, and we evaluate the kinematic and metabolic impact of a unilateral, bidirectional portable powered ankle-foot orthosis (PPAFO) in persons with gait impairment due to Multiple Sclerosis.

First, in Chapter 2, we address the state of portable powered pneumatic power sources. Specifically, we evaluated the use of compressed gas tanks with carbon dioxide or nitrogen as fuel. A test bench model of the PPAFO and walking trials (treadmill and over-ground) were used to evaluate each tank and gas, investigating normalized run time, minimum tank temperature, and rate of cooling. We concluded that compressed gas tanks can be used to successfully power portable pneumatic robotic platforms, especially when a recycling circuit can be implemented to increase the longevity of the fuel source, but considerations need to be taken into account in order to determine the proper fuel, based on size, weight, cost, and availability.

In Chapter 3, we evaluated a bidirectional, bilateral powered ankle-foot orthosis or exoskeleton system during over-ground walking in able-bodied individuals. With the powered PPAFOs, participants were able to reduce the metabolic power needed for walking compared to the unpowered PPAFO condition, and they were able to match the minimum metabolic power needed in shoes walking. Some kinematic changes were seen while using the PPAFOs, specifically an unexpected reduction in plantarflexion during toe-off.

In Chapters 4 and 5, we evaluated the use of a bidirectional powered ankle-foot orthosis to assist persons with gait impairment due to multiple sclerosis.

Use of the current embodiment of the portable powered AFO did not improve gait performance as measured by spatiotemporal parameters of gait. Significant differences in kinematic parameters at the ankle were observed such that the PPAFO was able to provide better assistance for foot drop during swing than the AFO or a shoes condition. Changes in kinematics at the knee were found such that the changes are likely due to compensatory reactions to the changes at the ankle induced by the footwear.

Throughout this work, we have been motivated to further research the mechanical design of the device so that users can better match their natural gait pattern in regards to spatiotemporal and kinematic parameters. Improving device design and functionality will help to determine if powered orthoses can be effective at assisting and improving gait function in persons with gait impairment.

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The studies in Chapters 3, 4, and 5 were all completed at the Illinois Fire Service Institute (IFSI). I am grateful to the staff of the research laboratory at IFSI: Prof. Gavin Horn, and Richard Kesler, MS for not only letting me use their space, but also helping out to organize and run data collection when I needed an extra set of hands. These studies would also have not been possible without the team of graduate students currently working on the PPAFO test bed, as your continued research made mine better—thank you to Mazharul Islam, Ziming Wang, and Mathew Petrucci.

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Contents

Chapter 1 Introduction

Lower limb exoskeletal and powered orthotic systems are currently being developed for a variety purposes ranging from augmenting the abilities of able-bodied individuals [1-3], assisting the abilities of those with physical limitations [4, 5], and studying the fundamental biomechanics and motor control of normal and pathological gait [6, 7]. As research continues in this area, there is a need to understand the current abilities and limitations of existing exoskeletal systems [8-10].

There are many types of lower limb exoskeletons, based on the number of actuated joints, and type of power system used [9, 10]. Generally, complete lower limb devices tend to neglect the ankle even though much of the force needed for gait is transferred through the joint to the ground [11-13]. A recent review of the use of powered lower limb exoskeletons for gait assistance even mentions the need for devices with powered ankle joints [4].

Current solutions for individuals that need assistance at the ankle due to functional impairments are passive braces or ankle-foot orthoses that provide only motion control and joint stability. The ideal ankle-foot orthosis (AFO) should be adaptable to accommodate a variety of functional deficits created by injury or pathology while simultaneously being compact and light weight to minimize energetic impact to the wearer. These requirements illustrate the great technological challenges facing the development of non-tethered, powered AFOs. Passive AFO designs are successfully used as daily wear devices because of the simplicity, compactness, light weight, and durability of the designs, but have limited functionality. To date, powered AFOs have not been commercialized and exist as research laboratory devices constructed from mostly off-the-shelf components [6, 14, 15]. Fully active devices have been developed to address limitations in motion control, such as the lack of plantarflexion motion and propulsion assistance during late stance [6, 16, 17] and the need for correcting drop foot during swing [18]. A few devices have been developed to address plantarflexor and dorsiflexor deficits during gait [19-21]. All of these devices have size and power requirements of the components and control algorithms that require tethered either power supplies or control electronics.

In this field of research, the terms powered orthoses and exoskeletons are both used, commonly to refer to devices trying to achieve similar yet different goals. Most often, the term exoskeleton is used to define a device that is being used to augment or investigate the behavior of an able-bodied individual. The term orthosis, or powered orthosis, is usually used in reference to a device designed to provide assistance to persons with some type of physical deficit or limitation. Although the terms exoskeleton and powered orthosis are usually used in this way, that does not preclude exoskeletons from being used in populations with a physical impairment, and powered orthoses from being used in a able-bodied populations.

In this literature review, we give an overview of a couple of areas of research in the field of ankle exoskeletons and powered orthoses. First we start with an introduction to our exoskeleton testbed used for research, the portable powered ankle-foot orthosis (PPAFO). The next section then reviews the other currently used ankle exoskeletons in the field. As many of these exoskeletons are pneumatically powered, we follow the review of current exoskeletons with an overview of the available pneumatic power sources for portable use. After reviewing the current device research, we look at research regarding gait impairment in a population who could benefit from such technologies: persons with multiple sclerosis.

Throughout the aforementioned reviews, we noticed differences in ways that researchers have quantified the impact of their exoskeletons on walking tasks. A common goal of exoskeleton research is to reduce the amount of work done or energy used by the human; therefore many methods have been used to quantify this reduction in human energy use. A short overview of a few of the different parameters used is the final section before the direct specific aims of this dissertation are described.

1.1. INTRODUCTION TO PPAFO

A test bed of a portable powered ankle-foot orthosis (PPAFO) has been developed to explore challenges associated with mobile and/or wearable robotic devices (Figure 1.1). More specifically the PPAFO test bed can used to investigate issues related to creating mobile actively-powered orthotic devices [22]. As a research group using the PPAFO testbed, we have been addressing issues of control, runtime, weight, and bulk of pneumatically-powered human-scaled mobile robotic systems.

The PPAFO can provide modest dorsiflexor or plantarflexor torque at the ankle using a portable pneumatic power source. The bidirectional assistance is applied at the ankle as needed throughout all phases of the gait cycle including late stance and propulsion, which is unavailable in current commercially-available technologies [14]. The timing of when to apply the bidirectional assistance is determined by a user-specific tuned kinematics-based controller that uses the PPAFO's toe and heel force sensors and ankle angle to estimate the state of the limb during the gait cycle [23, 24].

On the PPAFO, power is typically delivered to the ankle by our standard pneumatic circuit (Figure 1.2), two solenoid valves (VUVG 5V; Festo Corp-US, Hauppauge, NY) which control a rotary actuator (PRN30D-90-45, Parker Hannifin, Cleveland, OH) that can be actuated in dorsiflexion or plantarflexion directions).The PPAFO was designed to provide two levels of torque based on the direction of assistance. During plantarflexion, the PPAFO is designed to provide the maximum torque allowed by the pneumatic circuit components and available from the portable power source. With the current off-the-shelf components, the PPAFO is able to provide between 10-15 Nm of torque with 100-150 psig of pressurized gas during plantarflexion. A much smaller amount of torque is needed during swing to support the foot, such that a lower pressure (30-35 psig) is used to provide a 3-4 Nm torque in the dorsiflexion direction. A second pressure regulator mounted on the PPAFO (LRMA-QS-4; Festo Corp-US, Hauppauge, NY) down-regulates the inlet pressure to achieve the much smaller dorsiflexor torque needed.

The PPAFO test bed has explored two main controller approaches (Table 1.1 & Table 1.2). The PPAFO can be controlled using a direct event (DE) controller or a state estimation (SE) controller. The DE controller supplies directional actuation based on the current activation state of the two force sensors under the toe and heel. The two states (on/off) of each sensor (heel/toe) provide four distinct states that correspond to each of the four phases of gait where assistance is needed (Table 1.1) [22].The SE controller was designed to be trained and adapted to individual gait patterns [23, 24]. The sensor data that inform the controller of the user's current state come from the two force sensors under the toe and heel, and an angle sensor on the pneumatic actuator [24]. The SE controller provides actuation during four functionally distinct gait tasks: (1) initial contact, (2) loading response, (3) forward propulsion, and (4) limb advancement. Initial timings for the SE controller are based on normative gait event values determined from pre-existing data [25, 26] (Table 1.2); subject-specific PPAFO actuation timing can then be adjusted for each user's gait pattern [24]. Generally, the DE controller requires less computational power than the SE controller, but the SE controller is more adaptive to every participant. One of the portable power sources used is a compressed gas tank as the pneumatic power supply [22, 27]. Currently, the PPAFO has a limited portable runtime (8-25 minutes) due to the fixed amount of fuel available in the supply tank (depending on tank size, 9 oz vs. 20 oz). In our previous studies, a pneumatic recycling scheme was proposed [28] to improve system efficiency by reducing fuel consumption. The PPAFO test bed was used to test the suggested recycling scheme [27]. For the recycling scheme, an additional solenoid valve (VUVG 5V; Festo Corp-US, Hauppauge, NY) and custom accumulator were added into the pneumatic power circuit (Figure 1.3). A four-phase procedure allowed the compressed exhaust gas from each plantarflexor actuation to be stored in the accumulator and released later to power the following dorsiflexor actuation.

The accumulator is a custom-constructed pneumatic elastomeric accumulator (PEA) [29-32] (Figure 1.4). The PEA is assembled from off-the-shelf pneumatic fittings, latex tubing, and a cylindrical polycarbonate sheath. The PEA concept was based on previous work completed in hydraulics where fluid power energy was stored in an elastomer that was then able to return the stored energy to the system [33]. As the compressed gas enters the accumulator, the elastomer is allowed to expand within the sheath constraint. As the accumulator powers dorsiflexion, the strain energy stored in the expanded elastomer reenters the pneumatic power system along with the stored compressed gas, as the elastomer returns to its original relaxed state.

The development of the PPAFO, both with the standard and recycling pneumatic power schemes, has created a test bed for studying the many issues of portable, wearable robotics.

1.2. POWERED ANKLE EXOSKELETONS

In the past 10 years, the field of powered lower-limb exoskeletons has expanded greatly to begin to answer questions about human adaptation to assistive exoskeletons, and about ideal design and control of the assistive exoskeletons. Part of the work being done was to determine which design choices were best suited to accomplish the desired task of the exoskeleton. Previous reviews have documented the challenges in the field of lower limb exoskeletons, as well as the currently available technologies [8, 10]. A review written by Shorter et al. [14] focused on ankle-foot devices, reviewing passive, semi-active and active devices. In this section, we review the current work that has been done with powered ankle exoskeletons so that we may use our test bed to best contribute to the ongoing research in the field.

One of the most studied powered ankle-foot orthoses (PAFO) is a design that was originally published by researchers at the University of Michigan [6] (Figure 1.5). This device provided powered plantarflexion or powered dorsiflexion through pneumatic muscles, with controllers that were based on the EMG signals from the soleus muscle during gait. The pneumatic muscles were tethered to a laboratory air compressor, and the EMG controller was tethered to a laboratory desktop computer. An updated version of this PAFO was then created to provide both dorsiflexion and plantarflexion torque during the same experiment, such that EMG of the soleus was used to control the plantarflexion torque, and EMG of the tibialis anterior was used to control dorsiflexion [34] (Figure 1.6). The design of the PAFO was also expanded to include a knee - ankle - foot orthosis (KAFO), also with EMG control [35, 36]. Initial studies with the PAFO were completed with able-bodied participants, unilateral device use, and unidirectional powered actuation, to investigate motor adaptation in gait of able-bodied persons [6, 37] (Figure 1.7)[34].

Research regarding the locomotor adaptation of able-bodied persons has continued with the University of Michigan PAFO. One aspect of adaptation that was studied is the type of controller of the PAFO [38]. Although the original PAFO used an EMG-based control algorithm, a system was also designed using a forefoot footswitch controller (kinematic controller). The two controllers were compared with two groups of able-bodied young people with a unilateral, plantarflexion-only PAFO. Both sets of participants were able to adapt to their controller over the testing period, but those using the EMG-controller showed larger reductions in muscle activation, and more normal gait kinematics.

Kao et al. [7, 39], investigated the adaptation of able-bodied persons to a modified PAFOs. First, Kao et al. [39] investigated adaptation to a powered dorsiflexion-only, tibialis anterior EMG-controlled PAFO. In this study, healthy persons walked on a treadmill with the PAFO providing dorsiflexion assistance either during swing and initial heel contact, or just during swing. Both groups adapted to the dorsiflexion assistance and adapted to walking with reduced tibialis anterior EMG and increased ankle dorsiflexion by 9^{o}. Next, Kao et al. [7], modified the PAFO to included two artificial pneumatic muscles in parallel providing the plantarflexor torque for the effect of a greater perturbation (peak positive mechanical power with double pneumatic muscle PAFO (117 W), peak torque provided by PAFO (50.09 ± 12.05 Nm) [7]; peak positive mechanical power with single pneumatic muscle PAFO (107 W), peak torque provided by PAFO (estimated from Fig. 6a: 35 Nm)[37]) (Figure 1.8). The impact of the greater plantarflexor torque was a substantially different ankle kinematic pattern during gait, and an extended adaptation time compared to the lower plantarflexor torque design; so much so that some participants did not reach a new steady state gait pattern within two 30-minute practice sessions. Participants had greater power generation and less power absorption at the ankle joint, as well as a decreased peak plantarflexion during toe-off after adapting to the PAFO with greater plantarflexor torque compared to the PAFO with less plantarflexor torque.

A complete mechanical and energy analysis of the PAFOs for both level and incline walking in able-bodied individuals was completed by Sawicki and Ferris [40, 41]. In these studies, the researchers constructed a bilateral set of PAFOs for each participant to provide plantarflexion torque during gait controlled by soleus EMG signal. The studies reported that the participants were able to adapt their gait to the PAFOs within three 30-minute sessions of level ground walking. With practice, the participants significantly reduced the soleus muscle activity to allow the PAFO to assist powering part of push-off and significantly reduce the net metabolic power used to walk compared to an unpowered PAFO condition. In inclined walking, the PAFOs helped the participants reduce their net metabolic power by 10-13%, independent of incline gradient. Sawicki and Ferris [42] continued investigating mechanisms by which the PAFOs can cause a reduction in net metabolic power by investigating varying step lengths and gait speeds at a constant step frequency. The PAFOs helped participants to reduce the net metabolic power at all step lengths.

Other researchers have adopted the University of Michigan PAFO design and have helped to expand the research field of powered ankle-foot orthoses. At Virginia Tech, Norris et al. [43, 44] adapted the PAFO pneumatic muscle design such that the powered plantarflexor actuation was timed with information from a kinematic controller based on the angular velocity of the foot section of the orthosis (Figure 1.9). First, using a bilateral set of the kinematically controlled PAFOs, Norris et al. [43] found that the metabolic cost of transport could be decreased in able-bodied young adults, and that PAFOs may reduce walking stability, as stability measures were decreased when walking with the PAFOs. Second, Norris et al. [43] found that PAFOs augmenting plantarflexion increased the preferred walking speed in young adults; however the same increased in preferred walking speed was not seen in older adults (compared to an unpowered PAFO condition). While wearing the PAFOs, young adults also reduced their metabolic cost of transport while wearing the PAFOs compared to an unpowered condition with the PAFOs.

Another team of researchers that have adopted the Michigan pneumatic muscle PAFO for gait adaption studies were from Gent University [2, 16, 45-47]. Similar to Norris et al. [43], this group adapted the PAFO for powered plantarflexion, with the plantarflexion actuation timing determined by a kinematic controller based on a footswitch located under the heel, to create a wearable assistive lower leg exoskeleton (which they have recently called WALL-X, Figure 1.10). The first study [16] aimed to determine the ideal timing of turning the powered plantarflexion on in regards to gait cycle time, to reduce the metabolic cost of walking. In Malcolm et al. [16], they reported that with actuation timing starting at 43% of the gait cycle, the metabolic cost of walking could be reduced by 6% when wearing the WALL-X compared to walking without an ankle exoskeleton. A further study, using the 43% gait cycle actuation timing for plantarflexion actuation with the WALL-X, found that metabolic adaptation can be seen after 18.5 minutes, which was a faster adaptation than when they used a EMG feedback controller [46]. Further studies indicated that the WALL-X allowed for longer durations tolerated in an exercise test including a inclined walk with load carriage [2], and that actuation timing may need to be varied for inclined walking to an earlier percentage of gait to obtain maximal reduction in metabolic cost of walking (10% reduction in metabolic cost) [47].

Recently, a novel ankle exoskeleton was developed by researchers at MIT specifically aimed at reducing the metabolic cost of walking in able-bodied individuals [3, 48, 49] (Figure 1.11). This ankle exoskeleton provided a large plantarflexor torque (approximately 120 Nm [3]) during push off using an electrical winch actuator and strut. During swing, the ankle exoskeleton allowed free movement by providing slack in the drive cord such that the user did not notice any assistance or impedance from the exoskeleton. The plantarflexor torque timing was controlled kinematically by a gyroscope on the actuator, which provided the controller data regarding the angular shank velocity. Results indicated that this ankle exoskeleton was able to help participants reduce their metabolic cost of walking by up to 10 ± 3% compared to a shoes-only condition[49]. The ankle exoskeleton was also tested during walking with extra load carriage [3]. With the exoskeleton, while wearing a 23kg weighted vest, participants were able to reduce their metabolic cost of walking by 8% compared to not wearing the exoskeleton.

In attempts to create a light-weight powered AFO that can be powered from a wearable power source, researchers at the University of Minnesota have developed a portable hydraulic ankle-foot orthosis (HAFO, Figure 1.12) [50]. The goal for the HAFO was for the power density of hydraulics to be able to supply high torque at the ankle. A design goal was to separate the actuator and power supply to best distribute the weight, with as little weight at the ankle as possible. In published works, the HAFO has only been tested on an engineering test bench, where actuator and battery testing have produced promising results.

As summarized above, previous research on powered ankle exoskeletons has focused on mostly unidirectional powered devices that apply assistive torque during plantarflexion or dorsiflexion, with majority of the work focusing on plantarflexion actuation. All of this work was also done in a tethered laboratory setting, with participants on fixed-speed treadmills. As there is little previous research on bidirectional assistance at the ankle during able-bodied gait, by developing and testing the PPAFO, we hope to learn about the importance of dorsiflexion assistance as well as plantarflexion assistance during powered gait assistance, especially as it relates to gait pathologies with dorsiflexor weakness. In addition by developing a portable system such as the PPAFO, we hope to extend the current field of research by completing studies of over-ground walking which allows for more natural variation in gait.

1.3. REVIEW OF CURRENTLY AVAILABLE PORTABLE PNEUMATIC POWER SUPPLIES

One of the current challenges of creating portable powered devices is finding light-weight and long-lasting power supplies for effective runtimes. One major field of research in need of lightweight portable pneumatic power supplies are mobile or wearable soft robotic systems and pneumatically-powered exoskeletons [10, 51, 52]. There are currently limited options available for portable pneumatic power supplies. Portable pneumatic power sources such as microcompressors, combustion systems, and chemical decomposition systems are also being developed, but each presents with its own issues such as unwanted loud noise, high local temperatures, and toxic byproducts [53-63].

The current microcompressors that are being developed include a single stage thermocompressor [55] and a miniature free-piston engine compressor [61, 62]. The thermocompressor is a Stirling thermocompressor that works by shuttling air between a heat source and a room temperature heat sink with a displacer piston (Figure 1.14). By using a series of strategically located check valves, the thermocompressor can store the heated and pressurized gas for use in a pneumatic system [63]. For the thermocompressor, a proof of concept has been achieved, but work still needs to be done to appropriately size the components and determine the necessary additional components to power the thermocompressor and store the compressed air. The miniature free-piston engine in current research is of the homogenous charge compression ignition type (HCCI), and works by using a low temperature combustion system (Figure 1.13). In the design by Tian et al. [62], there is an engine unit and a compressor unit which work together to compress room air. One of the down sides of the HCCI miniature free-piston engine compressor is the need for liquid fuel needed for combustion. Another downside is the noise produced by the combustion cycle.

A unique use of the energy from combustion was developed by Shepherd et al. [56] which was used to power a soft robot. The soft robot was made organic elastomers with a passive valving system (“pneu-nets”) that allowed for the combustion of methane inside the robot (Figure 1.15). Although a novel idea, this concept is not widely adaptable to other pneumatic powered robots.

Chemical decomposition or phase changes have been harnessed by some researchers to develop fuel systems for pneumatic robots [53, 54, 58-60]. Kim et al. [53, 54] used a compact pistons pump for the injection of H_2O_2 into a pneumatic system. The decomposition of hydrogen peroxide (H_2O_2), in the presence of a catalyst, into water and oxygen produces heated and pressurized oxygen that the compressed gas that is used for the pneumatic system (Figure 1.16). One of the limitations of this system according to Kim et al. was the slow pressure generation rate [53, 54].

Another chemical system that has been harnessed for generating pneumatic power is CO_2 as dry ice. As dry ice “melts” it creates a pressurized liquid that is in equilibrium with a gas phase at the triple point. Since it is in equilibrium, the gas phase remains at a constant pressure, even when gas is removed from the container, as long as liquid phase is present [59]. Wu et al. [59] harnessed these properties of CO_2 into a “Dry Ice Power Cell”, by controlling the heat transfer from the environment into the pressure container (Figure 1.17). Further work on the development of the Dry Ice Power Cell has continued with work by Wu et al. [58].

Compressed carbon dioxide (CO_2) tanks are commonly used for handheld power tools and paintball gaming, and are easily refilled from bulk CO_2 tanks that can be rented or refilled at sports stores. One of the main factors that impacts the use and efficiency of compressed CO_2 as a portable power source is the cooling of the fuel as the tank is emptied, due to the endothermic expansion of CO_2 [64, 65]. This cooling is known to cause situations where the regulators may freeze up, especially in the paintball industry where high frequency or continual use duty cycles are common, contributing to poorly controlled pressure output [64, 66].

Another source of portable pneumatic power is compressed air in a high pressure air (HPA) tank. Traditionally the HPA tanks are filled with either high pressure compressed air or nitrogen (N_2), which can be refilled at specialty paintball locations or from high pressure scuba tanks. HPA are not reported to freeze up like the CO_2 tanks, and are preferred among paintball enthusiasts for game play due to a more consistent tank pressure [67].

Unlike the newly developed power sources which are developed for use in soft robotics and portable pneumatic systems, compressed gas tanks have not been tested as much in the research field of powered exoskeletons. As new fuel sources are being developed and tested, there is space to test the already existing fuel sources in pneumatic robots to provide a baseline evaluation for newer fuel systems.

1.4. GAIT IMPAIRMENT IN PERSONS WITH MULTIPLE SCLEROSIS

Gait impairment is one of the major day to day issues present in persons with Multiple Sclerosis (MS) [68]. The experienced gait impairments in persons with MS include muscle weakness causing foot drop, muscle tightness or spasticity, balance impairment leading to generalized ataxia, sensory deficits in the foot preventing proper sensation of the ground, and increasing severity of these impairments with increasing fatigue [69]. The major disease process of MS is characterized by demyelination lesions of the white matter of the brain stem, cerebellum, and spinal cord [70] resulting in lower limb weakness leading to this generalized gait impairment [71]. Much research has been done and is ongoing to better understand and quantify the different aspects of gait impairment in persons with MS. By studying gait, one can possibly better identify and diagnose persons with early onset MS [72], as well as offering interventions towards improving the gait impairment as it has been noted that any gait impairment can place a significant personal and financial burden, as well as a decrease in quality of life for persons with MS [73]. Passive AFOs are often used clinically to assist with foot drop due to lower limb weakness in persons with MS in attempts to mitigate the resulting gait impairment (Figure 1.18) [74]. Mixed results have been reported when evaluating the clinical and biomechanical advantages of patient-specific, physician-prescribed, custom passive AFOs for individuals with MS [72, 75-77].

Passive ankle-foot orthoses have been studied in mixed populations of persons with MS and stroke [76-78]. Bregman et al. [76] completed an in-depth analysis of the impact of a passive AFO on gait in persons with MS and stroke. They evaluated the effect of the mechanical properties of the AFO (stiffness and neutral angle as measured by a device designed to replicate a human leg that registers joint configuration and force exerted by the AFO (a.k.a., a BRUCE device) [79]) on the energy cost of walking, walking speed, gait kinematics, and kinetics. The researchers concluded that if the mechanical properties of the AFO matched the patient's needs, the patient greatly benefited with improved outcome measures. Overall they found that walking with a passive AFO decreased cost of walking and increased walking speed; although when the participants were divided into two groups (presence or absence of foot drop during swing), statistical power was diminished. Expected differences in ankle angle (decreased plantarflexion during swing and initial contact in the foot drop group, no change in the no drop group) were observed due to the AFO, but without statistical significance [76].

Bregman et al. [77] continued to study the impact of AFOs on persons with MS and stroke, specifically by using a spring-like carbon-composite AFO. With the AFO in that study, Bregman et al. observed a decrease in energetic cost of walking, decreased range of motion of the ankle, and decreased net work at the ankle. The decreased range of motion was due to reduced plantarflexion, which was mechanically blocked by the AFO, leading to reduced push-off. Even with decreased push-off, the AFOs were able to provide enough assistance in dorsiflexion to lead to an overall reduced metabolic cost [77]

Other groups have also looked into the use of AFOs in persons with MS. Ramdharry et al.[80] found that, although the passive AFO did initially cause destabilization of standing posture, after four weeks of using an AFO, persons with MS reported fewer limitations in their mobility (e.g. walking, running, stair climbing) when assessed using the Multiple Sclerosis Walking Scale-12 [81]. Sheffler et al. [75] looked at the impact of an AFO on functional gait tasks (timed 25-foot walk, five components of the modified Emory functional ambulation profile [82]), in persons with MS and saw no significant improvement with the AFO compared to no-device trials. McLoughlin et al. [83] investigated the effect of a dorsiflexion assist orthosis during a modified 6-minute walk test. When using the dorsiflexion assist orthosis (Figure 1.19), there was no difference in distance walked or perceived fatigue, but there was reduced physiological cost of walking when normalized by gait speed [83].

Novel designs of passive devices have been used for AFOs, such as Hwang et al., who compared an AFO-shaped band (elastic band attached to the thigh and shank with a loop to hold up the forefoot (Figure 1.20)), a molded plastic AFO, and barefoot conditions in persons with stroke and MS [78]. The AFO-shaped band increased the gait velocity and cadence compared to the barefoot and AFO conditions, with mixed results in stride length differences.

Although most of these studies agree in that passive AFOs can assist persons with MS to improve gait from an unassisted condition, the existing research is limited to devices that only provide assistance with dorsiflexion. The realization that these devices do not address plantarflexion weakness in persons with gait impairment motivates further device design. There is a need for devices that can more reliably assist with impairments during all phases of gait which can be seen in persons with MS.

It is possible that these traditional passive AFOs often fail to restore normal ankle function because they lack the ability to actively modulate motion control during gait and cannot produce propulsion torque and power. One possible direction the device design could go would be in the direction of actively powered bidirectional (plantarflexion and dorsiflexion) devices for assistance at the ankle. To better inform the design of such powered devices, we propose to use our portable, powered ankle-foot orthosis (PPAFO) test bed, which can provide bidirectional externally applied torque at the ankle during gait.

1.5. REVIEW OF METABOLIC MEASUREMENTS FOR QUANTIFICATION OF ENERGY EXPENDITURE

Throughout the research reviewed above regarding gait, whether with powered exoskeleton devices or in persons with gait impairment, a metabolic energy or cost parameter during the walking tests was often quantified. There is much variety between and within fields regarding how to compute a parameter to quantify the amount of energy used by a human during walking tests. In the exoskeleton work, primarily from the lab of Ferris and his collaborators, a variation of the Brockway equation [84] was used. We researched the previously used variations of the Brockway equations and documented when each first appeared in research literature (Table 1.3).

In the work completed by researchers studying gait in persons with MS, even different metabolic measurements were used to compute parameters of energy expenditure. Some literature searches were completed to find the original sources of most of the measurements (Table 1.4). There is no analysis in this dissertation of the different methods of computing metabolically related parameters other than to acknowledge that various methods exist. The variety in methods used does challenge a researcher to compare their device within the literature and understand the basis for each measurement.

1.6. SPECIFIC AIMS AND ORGANIZATION OF THE DISSERTATION

This dissertation aimed to use the PPAFO test bed to investigate some of the broad research issues in portable exoskeleton design. One way in which we used the PPAFO test bed was to understand the abilities and limitations of portable power supplies for current and future powered orthotic and robotic platforms. Another way in which we used the PPAFO test bed was to understand the effect of powered ankle joint actuation on able-bodied gait. The proposed research with the PPAFO will expand the current literature to understand the effects of bidirectional assistive ankle torque on able-bodied gait. Lastly, we used the PPAFO test bed to further understand the functional needs and constraints of a portable powered ankle assist device in a population with gait impairment due to Multiple Sclerosis. **Therefore the specific aims of this dissertation were:**

1. Evaluate fixed-volume compressed gas tanks as power sources for a wearable powered robotic system with and without a pneumatic recycling circuit. Specifically, we investigated CO_2 and N_2 based systems, used over the entire emptying time of the tanks.
2. Evaluate the impact of an externally applied bidirectional torque at the ankle during gait in an able-bodied population.
3. Evaluate the impact of an externally applied bidirectional torque at the ankle during gait in persons with MS in ambulatory tasks that correlate to real-world ambulation.

1.7. **TABLES AND FIGURES**

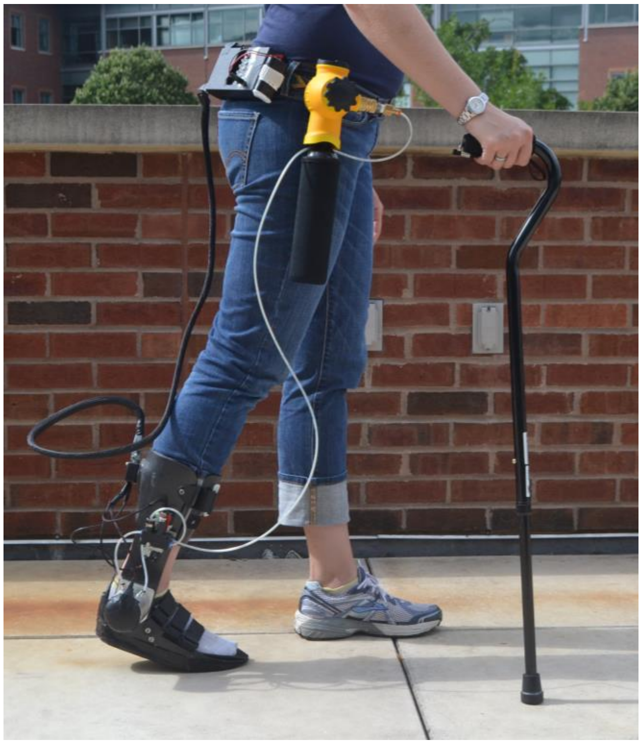


Figure 1.1: **Portable Powered Ankle-Foot Orthosis (PPAFO) with waist worn fuel tank and microcontroller.**

Table 1.1: Direct Event Controller Scheme [27] © 2013 IEEE

Functional Task	**Heel**	**Toe**	**Torque Direction**
1. Initial Contact	ON	OFF	Dorsiflexor
2. Loading Response	ON	ON	None
3. Forward Propulsion	OFF	ON	Plantarflexor
4. Limb Advancement	OFF	OFF	Dorsiflexor

Table 1.2: State Estimation Controller Scheme [27] © 2013 IEEE

Functional Task	**% gait cycle**	**Torque Direction**
1. Initial Contact	0-12	Plantarflexor
2. Loading Response	12-48	None
3. Forward Propulsion	48-62	Plantarflexor
4. Limb Advancement	62-67	None – Charging[a]
	67-100	Dorsiflexor

[a]. During SE without recycling, dorsiflexor torque starts at 60% of the gait cycle as there is no need for a charging phase
[b]. Gait cycle timings from Perry (1992) [25]

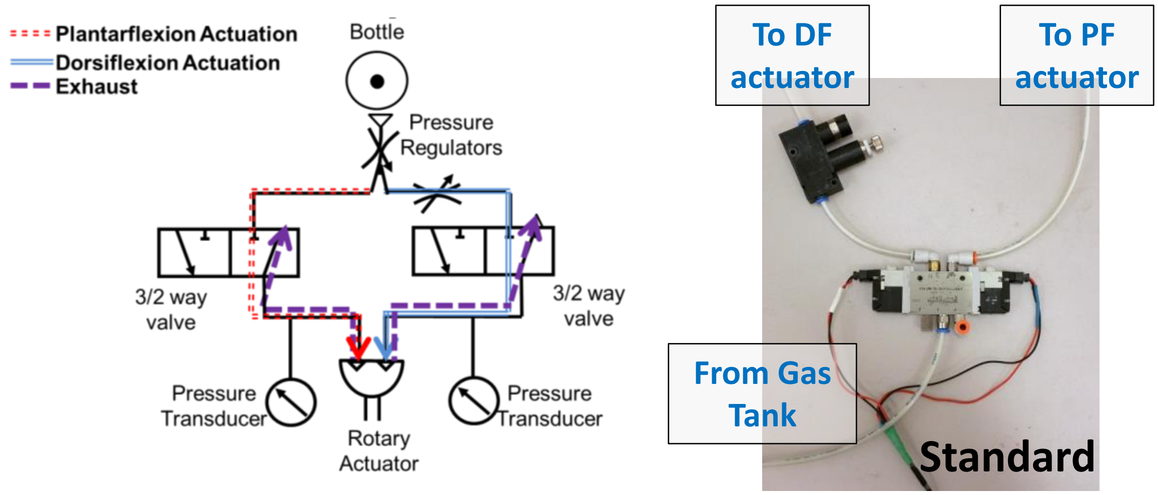


Figure 1.2: **Standard pneumatic power circuit for PPAFO.** Both the plantarflexion and dorsiflexion actuation received compressed gas from the fuel tank, through directional control valves. (Pneumatic circuit schematic © 2013 IEEE [27]).

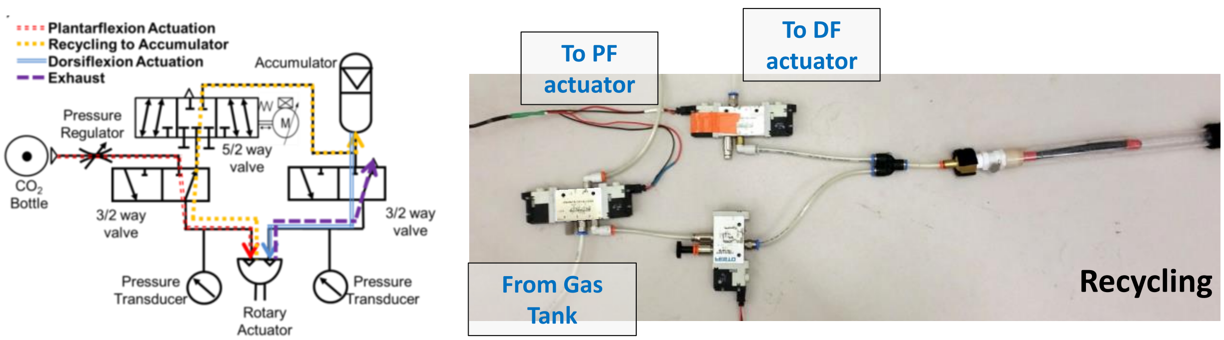


Figure 1.3: **Recycling pneumatic power circuit for PPAFO.** The plantarflexion actuation is powered directly from the compressed gas tank through a direction control valve, whereas the dorsiflexion actuation is powered by using the captured exhaust of the previous plantarflexion actuation. The exhaust is stored in a custom made elastomeric accumulator. (Pneumatic circuit schematic © 2013 IEEE [27]).

Figure 1.4: **Fully assembled custom elastomeric strain energy accumulator as to be used in the recycling pneumatic power scheme.** Top image shows uninflated accumulator, middle image shows partially inflated accumulated, bottom image shows fully inflated accumulator. The balloon expands and slides to fill the allowed space by the outer shroud.

Figure 1.5: **Original PAFO design by Ferris et al. 2005 [6]**

Reprinted with permission from [Ferris, Czernieck, and Hannaford, 2005, An ankle-foot orthosis powered by artificial pneumatic muscles, Journal of Applied Biomechanics, 21, 2, p. 189-97] © Human Kinetics, Inc.

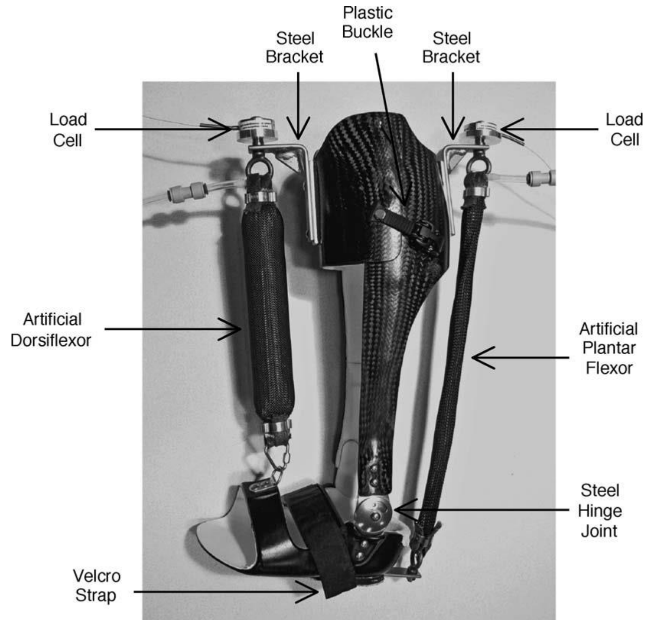


Figure 1.6: **Updated PAFO design by Ferris et al. 2006 [34]**

Reprinted from [Gait & Posture, 23(4), Ferris, D.P., K.E. Gordon, G.S. Sawicki, and A. Peethambaran, An improved powered ankle-foot orthosis using proportional myoelectric control, p. 425-8, (2006), with permission from Elsevier]

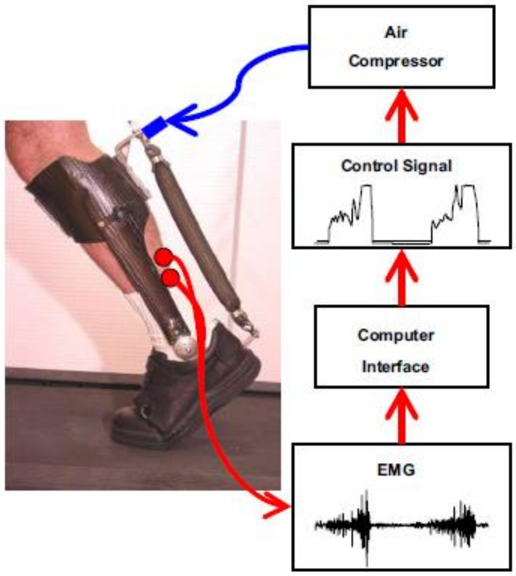
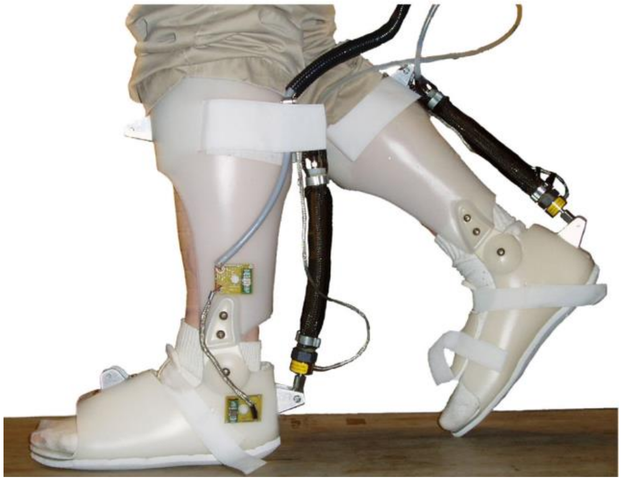


Figure 1.7: **Modified PAFO design by Gordon et al. 2007 [37].**

Reprinted from Journal of Biomechanics, 40 (12), Gordon and Ferris, Learning to walk with a robotic ankle exoskeleton, p. 2636-44, Copyright 2007 with permission from Elsevier

Figure 1.8: **Updated PAFO design with increased plantarflexion torque by Kao et al. 2010 [39].**

Reprinted from Journal of Biomechanics, 43(2), Kao, P.C., C.L. Lewis, and D.P. Ferris, Invariant ankle moment patterns when walking with and without a robotic ankle exoskeleton, p. 203-9, (2010), with permission from Elsevier.

A.


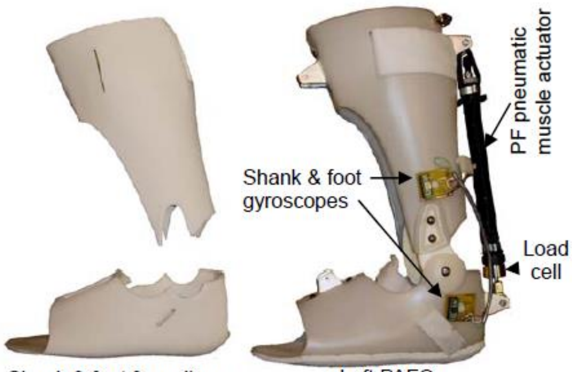
B.


Figure 1.9: **Updated PAFO design with gyroscopes used for signal input, by Norris et al. (A.[44], B.[43])**

A. Reprinted from Gait & Posture, 25(4), Norris, J.A., K.P. Granata, M.R. Mitros, E.M. Byrne, and A.P. Marsh, Effect of augmented plantarflexion power on preferred walking speed and economy in young and older adults, p 620-7, (2007), with permission from Elsevier.

B. Reprinted from ASME International Design Engineering Technical Conferences and Computers and Information in Engineering Conference, 2007, Norris, J.A., A.P. Marsh, K.P. Granata, and S.D. Ross. Positive feedback in powered exoskeletons: improved metabolic efficiency at the cost of reduced stability?, with permissions from ASME.

Figure 1.10: **WALL-X, adopted PAFO design by researchers at Ghent University [http://users.ugent.be/~ddclerc/WALL-X/]**

Image used with permission from Dr. Samuel Galle at Ghent University

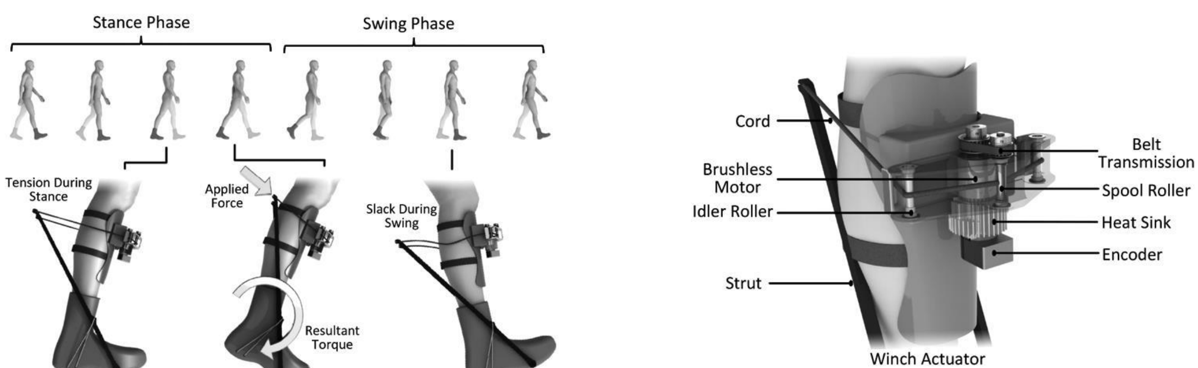


Figure 1.11: **Autonomous Ankle Exoskeleton designed by researchers at MIT [48]** © 2014 IEEE

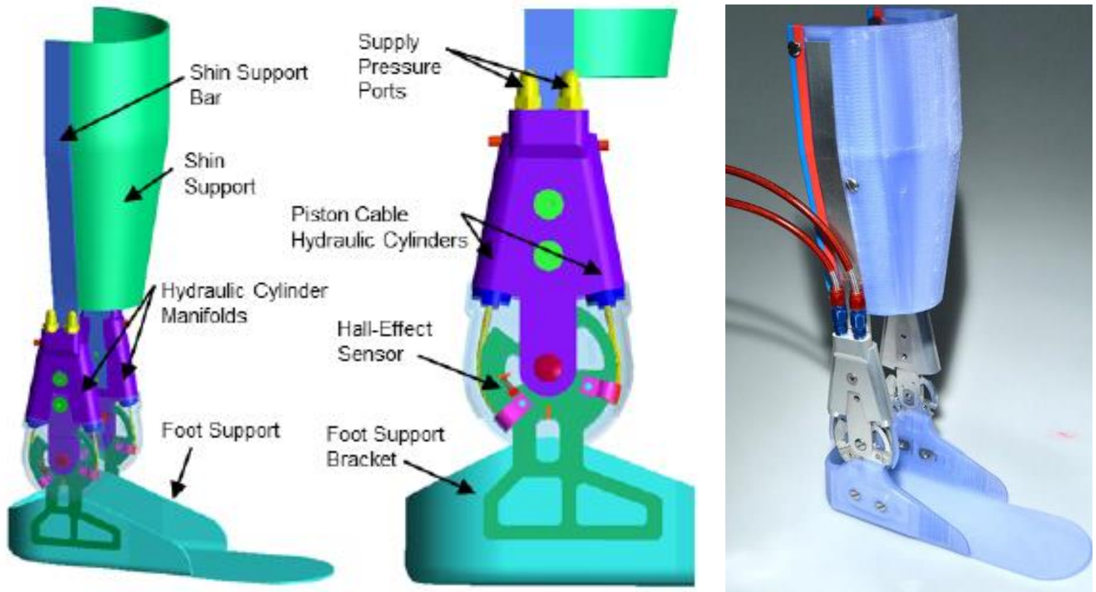


Figure 1.12: **Hydraulic Ankle-Foot Orthosis (HAFO) by Neubauer et al. [50]** © 2014 IEEE

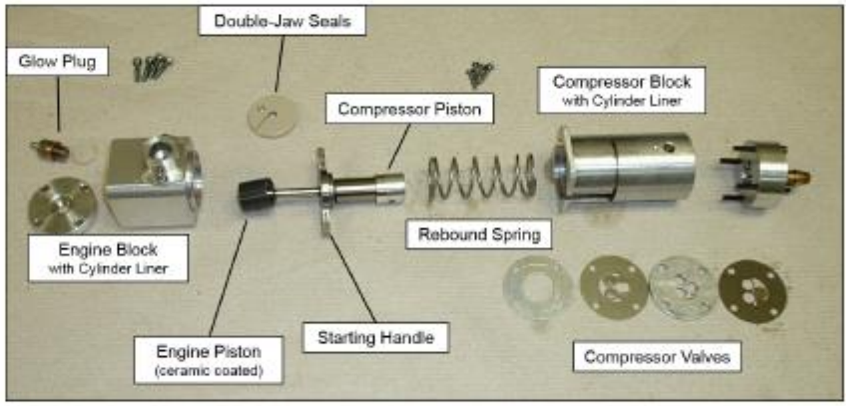



Figure 1.13: **The miniature free-piston engine compressor [61, 62]**

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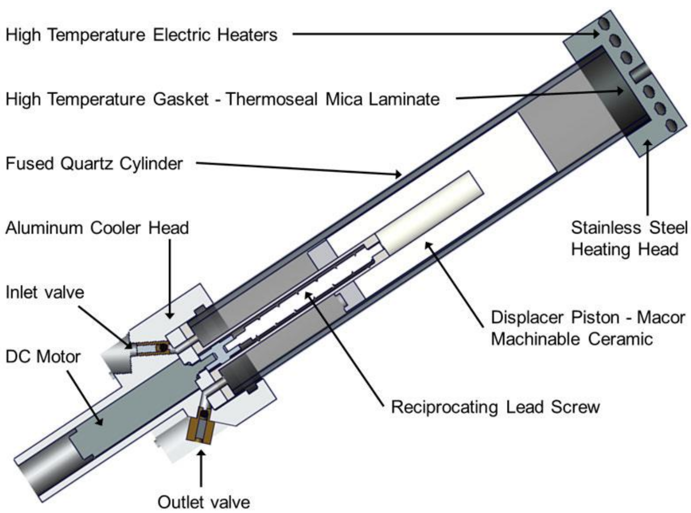

Figure 1.14: **Cross section of Stirling Thermocompressor** [63]

Reprinted with permission from ASME. [Hofacker, M.E., N.S. Kumar, and E.J. Barth, Dynamic Simulation and Experimental Validation of a Single Stage Thermocompressor for a Pneumatic Ankle-Foot Orthosis. Proceedings of the ASME/Bath Symposium on Fluid Power and Motion Control, 2013]

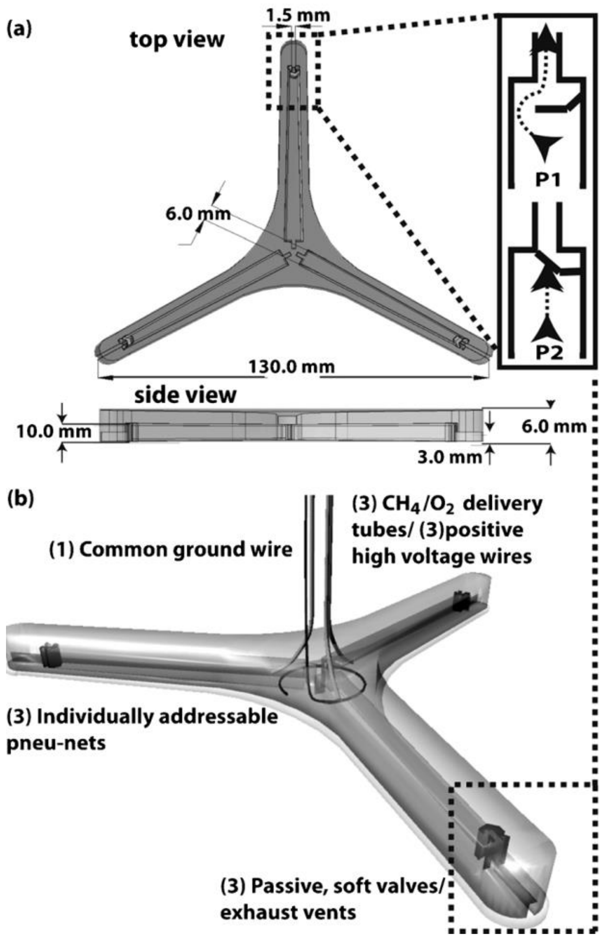

Figure 1.15: **Schematic of soft robot with combustion of methane for pneumatic fuel source** [56]

Reprinted with permission from Angewandte Chemie International Edition, 52(10), Shepherd, R.F., A.A. Stokes, J. Freake, J. Barber, P.W. Snyder, A.D. Mazzeo, L. Cademartiri, S.A. Morin, and G.M. Whitesides, Using Explosions to Power a Soft Robot, p. 2892-6, © 2013 Wiley-VCH Verlag GmbH & Co. KGaA, Weinheim

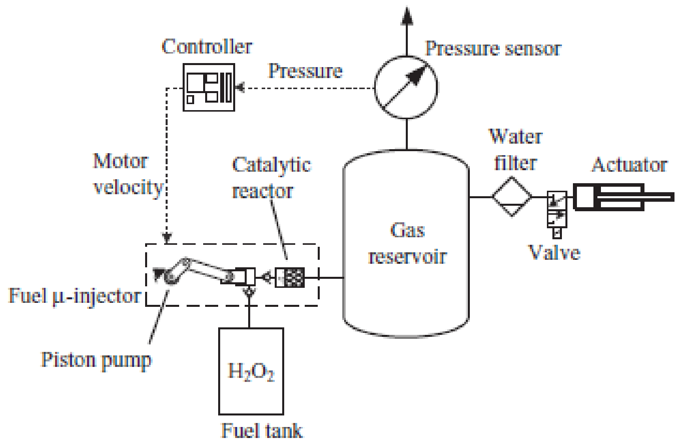

Figure 1.16: **Schematic of controllable pneumatic generator from decomposition of H_2O_2 [53]**

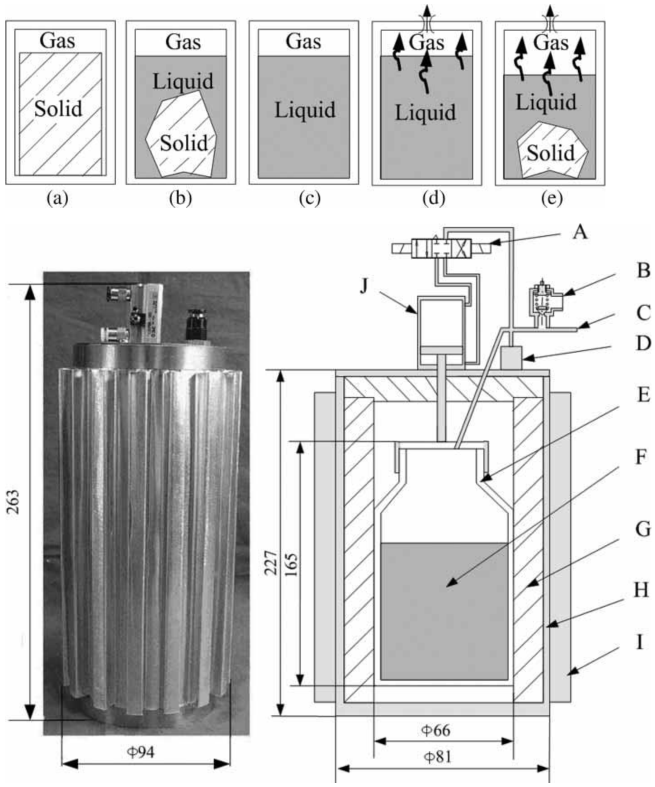

Figure 1.17: **Concept and schematic of the Dry Ice Power Cell** [58]

Reprinted with permission from Proceedings of the Institution of Mechanical Engineers Part C-Journal of Mechanical Engineering Science, 223(6), Wu, H., A. Kitagawa, H. Tsukagoshi, and S.H. Park, Development and testing of a novel portable pneumatic power source using phase transition at the triple point, p. 1425-32. © 2009 IMechE, SAGE.

Figure 1.18: **A low-profile dorsiflexion-assist AFO with free plantarflexion** [74]

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Figure 1.19: **A dorsiflexion assist orthotic** [83]

Reprinted from Archives of Physical Medicine and Rehabilitation, 96(2), McLoughlin, J.V., S.R. Lord, C.J. Barr, M. Crotty, and D.L. Sturnieks, Dorsiflexion Assist Orthosis Reduces the Physiological Cost and Mitigates Deterioration in Strength and Balance Associated With Walking in People With Multiple Sclerosis, p. 226-32, e1, © 2015, with permission from Elsivier.

Figure 1.20: **Alternatively shaped-AFO elastic band** [78]**.**

Table 1.3: Comparison of equations of metabolic parameters often used in gait studies with exoskeletons

Parameter name	**Equation**	**Notes**	**Reference**
Energy expenditure	$E(kJ) = 16.50\left(\frac{kJ}{L}\right) \times [VO_2](L) + 4.62\left(\frac{kJ}{L}\right) \times [VCO_2](L) - 9.06\left(\frac{kJ}{g}\right) \times [N](g)$	Weir equation as organized by Brockway	Weir, 1949 [85]; Brockway, 1987 [84]
Energy expenditure	$E(kJ) = 16.52\left(\frac{kJ}{L}\right) \times [VO_2](L) + 4.51\left(\frac{kJ}{L}\right) \times [VCO_2](L) - 9.22\left(\frac{kJ}{g}\right) \times [N](g)$	Weir equation coefficients modified by Brockway	Brockway, 1987 [84]
Energy expenditure	$E(kJ) = 16.58\left(\frac{kJ}{L}\right) \times [VO_2](L) + 4.51\left(\frac{kJ}{L}\right) \times [VCO_2](L) - 5.90\left(\frac{kJ}{g}\right) \times [N](g)$	Proposed by Brockway, and cited by many	Brockway, 1987 [84]
Net metabolic power	$P_{met,gross}(W) = 16.58\left(\frac{W\,s}{mL}\right) \times [\dot{V}O_2]\left(\frac{mL}{s}\right) + 4.51\left(\frac{W\,s}{mL}\right) \times [\dot{V}CO_2]\left(\frac{mL}{s}\right)$		Donelan et al., 2001 [86]
Net metabolic cost of transport	$COT_{met} = \frac{P_{met,gross}(W)}{body\ weight(N) \times walking\ velocity\left(\frac{m}{s}\right)}$	COT_{met} is dimensionless	Donelan et al., 2001 [86]
Metabolic energy expenditure rate	$\dot{E}_{met}(W) = 16.48\left(\frac{J}{mL}\right) \times [\dot{V}O_2]\left(\frac{mL}{s}\right) + 4.48\left(\frac{J}{mL}\right) \times [\dot{V}CO_2]\left(\frac{mL}{s}\right)$		Adamczyk et al., 2006 [87]

$[V\,O_2]$: Volume of oxygen consumed over period of measurement
$[V\,CO_2]$: Volume of carbon dioxide produced over period of measurement
$[N]$: Urinary nitrogen excretion over period of measurement
$[\dot{V}O_2]$: Average rate of oxygen consumed over period of measurement
$[\dot{V}CO_2]$: Average rate of carbon dioxide produced over period of measurement

Table 1.4: Comparison of equations of metabolic parameters often used in gait studies in persons with MS.

Parameter name	**Equation**	**Notes**	**Reference**
O_2 cost	$O_2\ cost = \frac{V\dot{O}_2\left(\frac{mL}{s}\right)}{body\ mass\ (kg)\ \times walking\ velocity\ \left(\frac{m}{s}\right)}$	"O_2 cost is defined as the amount of energy required to perform a task. During level walking it is the amount of oxygen consumed per kilogram body weight per unit distance traveled (mL/kg per m). Equivalent to the O_2 consumption rate divided by walking speed."	Waters et al., 1999 [88]
Energy equivalent of O_2	$"O_2 - eq." = \ 4.94\left(\frac{kJ}{L}\right) \times \left(\frac{[VCO_2]\ (L)}{[VO_2]\ (L)}\right) + 16.04\left(\frac{kJ}{L}\right)$		Garby and Astrup, 1987 [89]
Gross energy cost	No equation given, "Gross energy cost, defined as the total energy used per unit of distance, was calculated in $\frac{J}{kg*m}$ by dividing energy consumption (according to the Garby and Astrup method) during walking by walking speed."	Called "energy cost" by Bregman et al. [76];	Brehm et al., 2006 [90]

Chapter 2 : Evaluation of portable compressed gas tanks as fuel sources for portable pneumatic robots

2.1. ABSTRACT

Introduction: One of the current challenges of creating portable robots and exoskeletons is finding lightweight and long-lasting power supplies for effective runtimes. Currently available portable pneumatic power sources are limited. When choosing a portable pneumatic power source, considerations need to be taken into account in order to determine the proper fuel, based on size, weight, cost, and availability. Compressed gas tanks are a current option available as portable power sources. In wearable, portable robotics, it is important that the fuel source is analyzed for factors of the amount of added weight to the robotic system, the contact temperature for the wearer, and the runtime availability for the ideal running duration of the wearable device. This study analyzed portable tanks of compressed gas (carbon dioxide (CO_2), and nitrogen (N_2)) as power sources for a pneumatically powered ankle-foot orthosis (PPAFO). The PPAFO can be used to provide gait assistance to people with walking disability.

Methods: A test bench setup of the PPAFO system, with and without an exhaust-gas recycling power scheme, controlled to simulate walking was used to collect the temperature and mass of the gas tanks during continuous testing to determine the longevity of the gas tanks. Gas mass consumed and total runtime were also recorded during PPAFO use by healthy young adults during walking on a treadmill and over-ground.

Results/Discussion: The CO_2 tanks had much colder minimum temperatures (9 oz CO_2: -53.1 ^oC, 9 oz N_2: 8.0 ^oC) and much faster rates of cooling (9 oz CO_2: -9.9 ^oC/min, 9 oz N_2: -1.3 ^oC/min) than the N_2 tanks. With the extreme minimum temperature experienced by the CO_2 tanks, the presence of solid dry ice was noted inside the tanks. When a pneumatic recycling circuit was implemented, the (normalized) run time, regardless of fuel source, greatly increased (9 oz N_2 standard circuit: 1.42 min/oz; 9 oz N_2 recycling circuit: 1.88 min/oz). The walking trials had a longer run time than predicted by the test bench set ups (9 oz CO_2 treadmill: 1.69 min/oz, 9 oz CO_2 test bench: 1.11 min/oz), but followed similar average rate of cooling patterns (9 oz CO_2 treadmill: -6.6 ^oC/min, 9 oz CO_2 test bench: -9.9 ^oC/min).

Conclusions: This study concluded that the selection of which compressed gas tanks to use as a fuel sources for a portable pneumatic robotic, depends on one's working device constraints. When trying to extend the runtime of a compressed gas tank in a pneumatic system, a pneumatic recycling circuit always increased the run time of the gas tank. Due to the increased cooling in CO_2 and possible issues of equipment freezing, along with the possibility of extreme cold contact with the wearer in a wearable robotics platform, compressed N_2 was found to be the better fuel source. Although when total system weight and availability were taken into account, a case can easily be made for using CO_2 instead of compressed $N_{2.}$

2.2. **INTRODUCTION**

One of the current challenges of creating portable powered devices is finding lightweight and long-lasting power supplies for effective runtimes. Two major fields of research in need of lightweight portable pneumatic power supplies are mobile or wearable soft robotic systems and pneumatically-powered exoskeletons [10, 51, 52]. There are currently limited options available for portable pneumatic power supplies. Portable pneumatic power sources such as microcompressors, combustion systems, and chemical decomposition systems are also being developed, but each presents with its own issues such as unwanted loud noise, high local temperatures, and toxic byproducts [53-60].

Fixed-volume compressed carbon dioxide (CO_2) tanks are commonly used for handheld power tools and paintball gaming. One of the main factors that impacts the use and efficiency of compressed CO_2 as a portable power source is the cooling of the fuel as the tank is emptied, due to the endothermic expansion of CO_2 [57, 64]. This cooling is known to cause situations where the regulators may freeze, especially in the paintball industry where high frequency or continual use duty cycles are common, contributing to poorly controlled pressure output [64, 66]. We hypothesized that the cooling of the fuel and tank will reduce the operating pressure and increase fuel consumption rate, which are factors that may reduce the efficiency and longevity of a portable robotics platform driven by compressed CO_2.

Another source of portable pneumatic power is compressed gas in a high-pressure air (HPA) tank. Traditionally HPA tanks are filled with either high-pressure compressed air or nitrogen

(N_2). HPA tanks are not reported to freeze like CO_2 tanks, and are preferred among paintball enthusiasts for game play due to a more consistent tank pressure [65]. Some of the disadvantages of HPA tanks as a portable power source are their increased weight due to tank construction necessary for higher pressure storage, increased cost of tanks, and limited availability of refill locations or the high cost of acquiring high pressure tanks or compressors for refilling. We hypothesized that HPA tanks will provide more consistent pressure output and limited cooling of N_2, such that the efficiency and longevity of a robotics platform will be greater compared to CO_2.

In this study, the portable pneumatically-powered system utilized as the test platform was the portable powered ankle-foot orthosis (PPAFO) [22, 27]. The PPAFO uses pneumatic power to generate torque at the ankle via a rotary actuator (Figure 2.1). To be a portable system, the PPAFO has been operated with fixed-volume CO_2 tanks. The PPAFO provides two levels of torque based on the direction of assistance needed during the gait cycle (Figure 2.2). During late stance, a plantarflexor torque is provided by the PPAFO to help with push-off and forward propulsion. A much smaller dorsiflexor torque is needed by the PPAFO to support the ankle-foot complex during leg swing and initial contact with the ground during the next step. These plantarflexor and dorsiflexor torques are generated with operating pressures of approximately 100 psig and 30 psig, respectively. To improve runtime without increasing the size of the power source, a pneumatic recycling scheme can be implemented which captures exhaust gas from the high-pressure plantarflexor part of the cycle and then uses that exhaust to power the low-pressure dorsiflexor part of the cycle [27, 28].

The aim of the current study was to evaluate fixed-amount compressed gas tanks as power sources for a wearable powered robotic system with and without a pneumatic recycling circuit. Specifically, we investigated CO_2 and N_2 based systems, used over the entire emptying time of various sized tanks in test bench trials and when used to assist with walking on a treadmill and over-ground. We hypothesized that an increased rate of cooling due to continuous usage of the CO_2 will directly lead to shorter runtimes compared to N_2. Secondly, we hypothesized that the different sized CO_2 bottles would have no difference in normalized run time, but would show difference in cooling rates. Lastly, we hypothesized that there would be no difference in performance between simulated walking behavior on a test bench and actual walking on a treadmill or over-ground, such that the test bench configuration was an appropriate representation of actual device operation during walking.

2.3. METHODS

We completed the study using the PPAFO testbed powered by compressed gas tanks. The primary gas used in the study was CO_2, with a secondary analysis of using N_2 gas. Due to availability in the area, only N_2 was evaluated in this study, where compressed air (~78% N_2) was not evaluated. This analysis of compressed gas was conducted on a test bench created to emulate the wearable PPAFO system. The test bench included all of the same pneumatic components as the wearable PPAFO, as well as sensors for data collection. To confirm that our findings from the test bench were a reliable measure of the intended use of the compressed gas tanks in portable powered orthoses, the wearable PPAFO operated with CO_2 was tested on four test subjects during treadmill walking and seven test subjects during over-ground walking.

2.3.1. PPAFO Testbed

The PPAFO can provide modest dorsiflexor or plantarflexor torque at the ankle via an untethered pneumatic power source [22]. Power is typically delivered to the ankle by our standard pneumatic circuit: two solenoid valves (VUVG 5V; Festo Corp-US, Hauppauge, NY) control a bi-directional rotary actuator (PRN30D-90-45, Parker Hannifin, Cleveland, OH). The plantarflexor torque (high pressure) is powered directly from the gas tank with a bottle top regulator (JacPac J-6901-91, Pipeline Inc., Waterloo, ON, Canada). Dorsiflexor torque (low pressure) is powered from the gas tank with a bottle top regulator and a secondary in-line regulator (LRMA-QS-4, Festo Corp-US, Hauppauge, NY) to down regulate the pressure further (Figure 2.3 and Figure 2.5). With the current off-the-shelf components, the PPAFO can provide between 10-15 Nm of torque with 100-150 psig of pressurized gas during plantarflexion and 3-4 Nm of torque with 30-35 psig during dorsiflexion. For the recycling power scheme, an additional solenoid valve (VUVG 5V; Festo Corp-US, Hauppauge, NY) and custom accumulator were added into the pneumatic power circuit (Figure 2.4 and Figure 2.5).

The accumulator was a custom-constructed pneumatic elastomeric accumulator (PEA). The PEA was assembled from off-the-shelf pneumatic fittings, latex tubing, and a cylindrical polycarbonate sheath (Figure 2.7). The PEA concept was based on previous work completed in hydraulics where fluid power energy was stored in an elastomer that was then able to return the stored energy to the system [33]. As the compressed gas enters the accumulator, the elastomer expands within the sheath constraint. As the accumulator powers dorsiflexion, the strain energy stored in the expanded elastomer reenters the pneumatic power system along with the stored compressed gas, as the elastomer returns to its original relaxed state.

The recycling power scheme was a four-phase procedure that allowed compressed exhaust gas from each plantarflexion actuation to be stored in the accumulator and released later to power the following dorsiflexion actuation [27] (Figure 2.4). The recycling power scheme started with phase one where the plantarflexion actuation was powered directly from the gas tank. Phase two captured the exhaust from the plantarflexion power into the PEA. Phase three allowed for dorsiflexion actuation, by connecting the PEA to the dorsiflexion side of the actuator while venting the plantarflexion side of the actuator to atmosphere to allow for the PEA to power the actuator. The final phase exhausted the dorsiflexion side of the actuator, preparing for another cycle to begin.

The PPAFO was controlled by a state estimation controller (SE) [23, 91], where the current state of the gait cycle was estimated by sensor input from heel and toe sensors on the sole of the PPAFO (force sensitive resistor, SEN-09376, SparkFun Electronics, Niwot, CO), as well as the current ankle angle from the rotary actuator position (programmable angle sensor, KMA199E, NXP Semiconductors, San Jose, CA). The SE controller provided actuation during four key regions of gait associated with four functional tasks: (1) initial contact, (2) loading response, (3) forward propulsion, and (4) limb advancement (Figure 2.2). Suggested timings for the SE controller were based on normative gait event values determined from pre-existing data [25, 26] (Table 2.1).

2.3.2. Data Collection

For all studies below (test bench, treadmill, and over-ground), trials were completed on different days based on availability of equipment, availability of participants, and refill of compressed gas tanks. Efforts were made to ensure that environmental conditions were the same for every trial, and gas tanks were allowed to equilibrate to room temperature after being refilled before being tested again.

2.3.2.1. Test bench study

To study the use of each of the gas sources and the effect of the recycling circuit on their use, a tabletop version of the PPAFO system was created. The test bench included all of the same pneumatic components as the PPAFO (Figure 2.5). For the test bench trials, the pneumatic valves were controlled using a custom Simulink (v 7.13.0.564, the Mathworks, Waltham, MA) data collection model, and programmed to represent one gait cycle per second with appropriately timed plantarflexion and dorsiflexion actuation based on the temporal percentage of the gait cycle, emulating the SE controller (Table 2.1) [27]. To mimic the inertial properties of the foot ($I = 44.0$ kg*cm^2, average of 100 males (p.304, Table 4.4, [92])) and the range of motion limits of the ankle (dorsiflexion 10^o, plantarflexion 20^o [25]), an uniform 304L stainless steel bar (91.4 cm x 3.18 cm x 0.32 cm, 0.72kg) was attached to the shaft of the rotary actuator. By using a test bench PPAFO, the longevity of each power source was evaluated under multiple controlled test conditions without the need for numerous able-bodied participants. A longevity test consisted of running the PPAFO test bench, as if a person was walking continuously, until a gas tank was emptied. The run time was defined as the amount of time it took from starting the trial with a full compressed gas tank until the tank was no longer able to provide enough pressure to rotate the actuator.

The longevity testing evaluated three different fuel tanks. The tanks used in the testing were: (1) 9 oz fuel capacity CO_2 tank; (2) 20 oz fuel capacity CO_2 tank (both from Catalina Cylinders, Hampton, VA) and; (3) 4500 psig 68 cubic inch HPA tank (Ninja Paintball, Crystal Lake, IL). The CO_2 tanks (9 oz and 20 oz) were filled by weight to their rated fill capacity from a bulk CO_2 refill tank in our facility. The HPA tanks were filled with high pressure N_2 to ~4300 psig based on the availability of the refill station at the paintball supplier (Firemark Paintball, Dewey, IL), which corresponded to ~9 oz of N_2 gas. The bottle top regulator was placed on a full gas tank and set to 100 psig at the beginning of each trial. Data were collected until the trial ended, defined by the actuator no longer being able to rotate through the intended range of motion (visual inspection), and no longer hearing audible exhaust from the pneumatic valves (auditory inspection). Longevity trials were run for both the standard and recycling pneumatic circuits with each gas tank. For the 9 oz CO_2 tanks, three full tanks were run per pneumatic circuit. For the 20 oz CO_2 tanks and 9 oz N_2 tanks, two full tanks were run per pneumatic circuit (Table 2.4 and Table 2.5).

During each longevity test, temperature and mass were measured (Figure 2.5). The temperatures of the gas tank and regulator were recorded using a digital thermometer with thermocouples (HH303, Type K, Omega Engineering, Stamford, CT) attached to the bottom surface of the bottle and the output valve of the regulator. The minimum tank temperature was determined to evaluate the possibility of freezing conditions around the regulator. The gas tank and regulator were suspended such that their combined mass was recorded by a uniaxial force gauge (DFG35-10, Omega Engineering, Stamford, CT). The pneumatic pressure at the bottle top regulator outlet was recorded by a pressure transducer (AST4000A00150P3B1000, American Sensor Technologies, Mount Olive, NJ). The pneumatic pressure at the input to the plantarflexion side of the rotary actuator was recorded by another pressure transducer (AST4000A00100P3B1000, American Sensor Technologies, Mount Olive, NJ). The pneumatic pressure at the input to the dorsiflexion side of the actuator was recorded by a third pressure transducer (2091050PG1M2405P, Setra, Boxborough, MA). Pressures and angle were collected using a data acquisition device (Q8-USB, Quanser, Markham, Ontario, Canada), and were directly imported to the testing computer using Simulink. Mass and temperature were collected via USB ports and drivers on the computer and read directly into MATLAB (v2013a).

2.3.2.2. Treadmill walking testing

In order to compare the well-controlled test bench scenario for the CO_2 gas source to results collected while walking with the PPAFO, four healthy young participants (1 female/3 male, age: 23.1 ± 3.3 years) were tested while walking on a treadmill. Approval for the study was granted by the Institutional Review board and participants gave informed consent. Participants performed continuous walking on a treadmill while wearing a PPAFO on the right leg, with a comfortable walking shoe on the left leg. The treadmill was set to a self-selected comfortable speed, while asking the participant to achieve the same gait cycle timing as the test bench (i.e., 1 Hz step rate created with a metronome set to 60 bpm). The PPAFO actuation timing was controlled by a waist-worn microcontroller programmed with the SE controller, using each participant's real-time signals to estimate the phase of gait, and apply proper directional actuation at the timings used in the test bench study (Table 2.1)[23].

Each participant completed two treadmill trials, one with the PPAFO in the standard pneumatic circuit, and one with the PPAFO in the recycling pneumatic circuit. Each trial lasted the length of time that a 9 oz CO_2 tank was able to power the PPAFO. Trials were ended when the pressure gauge on the bottle top regulator dropped below 30 psig (visual inspection), and there was no longer audible exhaust from the pneumatic valves (auditory inspection). During each trial, as with the test bench trials, temperature and mass of the PPAFO system were collected through USB computer interfaces into MATLAB. The CO_2 tank was hung from the same testing rig as the test bench to measure the mass, such that the participant did not have to carry the gas tank while walking.

2.3.2.3. Over-ground walking testing

To investigate the run time and use of the CO_2 power source during more natural unconstrained over-ground walking, an additional testing condition was performed on seven healthy young adult participants (2 female/5 male, age: 21.1 ± 1.1 years). Approval for the study was granted by the Institutional Review Board, and participants gave informed consent.

Participants wore a PPAFO on each leg (bilateral) in the standard pneumatic circuit. Participants first wore the orthoses for an adaptation period (>20 min) where their gait pattern was programmed into the PPAFO's modified SE controller [24]. The additional SE controller programming allowed for greater adaptive power of the controller, to better estimate the phase of gait based on how each participant independently activated the heel and toe sensors, and each participant's ankle angle during gait, as opposed to using normative gait values [24, 93].

Participants performed one continuous over-ground walk in a 9 m x 12 m hallway loop with four 90-degree turns. The trial began with one full 20 oz CO_2 bottle per PPAFO, and the trial ended when there was no longer audible exhaust from the pneumatic circuit (auditory inspection), and the participant no longer felt any assistance from the PPAFO on one side. Each participant carried two CO_2 tanks (one per PPAFO) on a tool belt at their waist during the over-ground walking trials. The total consumed mass of CO_2 was recorded for each condition, as well as the total distance walked, and the total walking time until the first CO_2 tank was empty.

2.3.3. Data Analysis

For the test bench, the outcome measures of normalized run time (min/oz), mass consumed (oz), rate of cooling (^oC /min), and minimum temperature (^oC) were analyzed to determine the relationship of these factors to longevity, and to compare these values across tank sizes (9 oz, 20 oz), and fuel types (CO_2, N_2). Normalized run time was computed as time per amount of fuel consumed, such that the total run time was divided by the total mass consumed. Rate of cooling was computed from the initial to minimum temperatures of a tank and the amount of time needed to reach the minimum temperature in a trial. A linear approximation was determined to represent the experimental data (Figure 2.8).

2.3.4. Statistical Analysis

Outcome measures of normalized run time, mass consumed, minimum temperature and rate of cooling were compared to assess compressed gas fuel types (CO_2 and N_2), experimental testing conditions (test bench, treadmill walking and over-ground walking), tank sizes (9 oz and 20 oz), and pneumatic circuit type (standard and recycling) . Due to only certain trials performed per fuel type, experiment condition, and tank size, the collected data did not allow for full analyses of test condition (3) × fuel type (2) × tank size (2). First, the 9 oz tank trials were compared with a MANOVA, using normalized run time, minimum tank temperature, and rate of cooling, as the dependent variables with type-experiment trials (9 oz CO_2 test bench, 9 oz N_2 test bench, and 9 oz CO_2 treadmill walking) and pneumatic circuit type (standard vs. recycling) as fixed factors. Second, the different CO_2 tank sizes were compared with a MANOVA, to investigate differences in normalized run time, minimum tank temperature, and rate of cooling as the dependent variables with tank size (20 oz CO_2 and 9 oz CO_2 test bench) and pneumatic circuit type (standard vs. recycling) as fixed factors. For these MANOVAs, Bonferroni post-hoc comparisons were performed when appropriate. Lastly, an independent-samples T-test was used to compare only experiment conditions (20 oz CO_2 test bench to the 20 oz CO_2 over-ground walking) for the parameter of normalized run time. For all tests, the level of significance was set to $\alpha = 0.05$.

2.4. RESULTS

The first statistical comparison was to directly compare the different types of fuel, experimental testing conditions, and pneumatic circuits. The 9 oz CO_2 test bench, 9 oz N_2 test bench, and 9 oz CO_2 treadmill conditions were compared with both standard and recycling pneumatic circuits. The MANOVA found overall significance for differences in type-experiment trials ($p < 0.001$) and pneumatic circuit type ($p < 0.001$), as well as a significant interaction ($p < 0.001$). For the parameter of normalized run time, the 9 oz CO_2 treadmill trials were significantly longer than the 9 oz CO_2 test bench and the 9 oz N_2 test bench conditions ($p < 0.001$, Figure 2.9). There was no significant difference between fuel type (N_2 vs. CO_2) for normalized run time. The recycling trials had significantly longer normalized run times than the standard pneumatic circuit trials ($p < 0.001$, Figure 2.9). For the rate of cooling (Figure 2.10), all trials were significantly different from one another ($p < 0.001$), and the standard and recycling pneumatic circuits were significantly different ($p < 0.001$), with a significant interaction between circuit and trial type ($p < 0.001$). The parameter of minimum tank temperature (Figure 2.11) showed a significant difference between trial type, with the 9 oz N_2 test bench having a significantly warmer minimum tank temperature than both the 9 oz CO_2 test bench, and 9 oz CO_2 treadmill conditions ($p < 0.001$).

The second statistical comparison was to compare the different sized CO_2 tanks with the standard and recycling pneumatic circuits on the test bench for the parameters of normalized run time, rate of cooling, and minimum tank temperature. The MANOVA showed significant effects for bottle size ($p < 0.001$) and pneumatic circuit type ($p < 0.001$) with a significant interaction effect ($p = 0.005$). The normalized run time was significantly different between the standard and recycling pneumatic circuits ($p < 0.001$,Figure 2.12), with no significant differences based on tank size (9 oz vs. 20 oz). The rate of cooling (Figure 2.13) was significantly different for bottle size ($p < 0.001$), pneumatic circuit ($p < 0.001$), with a significant interaction effect ($p < 0.001$). The minimum tank temperature (Figure 2.14) also was significantly different for bottle size ($p < 0.001$), pneumatic circuit ($P < 0.001$), with a significant interaction effect ($p < 0.001$).

The last statistical comparison was to compare the test bench to over-ground walking, both with 20 oz CO_2 tanks. An independent samples T-test showed no significant difference between the normalized run time for the 20 oz CO_2 test bench and 20 oz CO_2 over-ground walking trials (Figure 2.15).

2.5. DISCUSSION

This study evaluated fixed-volume compressed gas tanks as power sources for powered robotic systems. In this study, the implementation of a pneumatic recycling circuit was used to further evaluate the compressed gas tanks in robotic systems. The robotic system used in this study was the portable powered ankle-foot orthosis (PPAFO). Two different compressed gas systems (CO_2 and N_2) with various sized tanks were analyzed on a test bench as well as treadmill and over-ground walking. We hypothesized that the tanks of CO_2 would have shorter normalized run times than the tanks of N_2. When comparing different sized tanks of CO_2, we hypothesized that the normalized run times would be the same regardless of tank size, but a slower rate of cooling would be demonstrated in a larger volume tank. Lastly, we hypothesized that the performance of compressed gas tanks as fuel on the test bench (with simulated walking behavior) would act similarly to actual over-ground and treadmill walking with the PPAFO.

To address our first hypothesis, we compared a 9 oz tank of N_2 on the test bench to 9 oz tanks of CO_2 on the test bench and while walking on the treadmill. The CO_2 tanks (regardless of testing condition), had significantly colder minimum temperatures compared to the 9 oz N_2 condition (Figure 2.11). As we hypothesized, the CO_2 tanks had significantly faster rates of cooling than the N_2 tank (Figure 2.10), although we did not see the expected difference in normalized run time between the 9 oz CO_2 and 9 oz N_2 test bench conditions (Figure 2.9). The relationships between runtime and temperature were thermodynamically-based, with each gas type having its own unique characteristics. The gas characteristics such as those commonly depicted in a phase diagram (i.e. phase transition curves), where the temperature and pressure of the gas inside the tank determine whether the fluid exists as a gas or as a combination of gas, solid, and liquid are responsible for some of the differences in behaviors between gases that were observed in this study. The characteristics of CO_2 are especially worth noting; as with the pressures and temperatures experienced in running down a CO_2 tank, it is possible to go through multiple phase transitions. The difference in minimum temperatures of the two gasses (CO_2: -53.1 ^oC, N_2: 7.9 ^oC) is important to note, as the temperature of the working fluid can have impact on the functional of pneumatic components. Most pneumatic components are rated for working fluid temperatures, such that mechanical and electrical parts inside the components will not freeze up or be damaged by extreme temperatures.

Interestingly, the 9 oz CO_2 treadmill condition had the longest normalized run time compared to the 9 oz CO_2 test bench condition (Figure 2.9, Table 2.4), likely due to the variability in actuating against a biological active ankle in treadmill walking, as opposed to the more constant set up of the test bench. It is possible that while wearing the PPAFO, participants went through smaller ranges of motion at the ankle during walking than expected, using less volume of fuel per actuation.

As expected from previous work [27], the recycling circuit provided an increased normalized run time compared to the standard pneumatic circuit (Figure 2.9). The recycling circuits also provided for a reduced rate of cooling in the CO_2 trials, as the tanks still eventually reached the same minimum temperature as the standard circuit but did so over a longer run time (Figure 2.8 & Figure 2.10).

Our second hypothesis involved comparing different sized CO_2 tanks with both the standard and recycling pneumatic circuits on the test bench. As we hypothesized, there was no difference due to CO_2 tank size for normalized run time (Figure 2.12) and there was a significant difference in cooling rate (Figure 2.13). Additional findings in this comparison included the expected increase in normalized run time due to the recycling pneumatic circuit.

Lastly, we hypothesized that there would be no difference between test bench and treadmill or over-ground walking. The comparison between 20 oz CO_2 test bench and 20 oz CO_2 over-ground walking showed no significant differences for normalized run time (Figure 2.15). There was a significant difference between normalized run time for 9 oz CO_2 test bench and 9 oz CO_2 treadmill, with the treadmill having a greater normalized run time (Figure 2.9). It is not clear-cut to draw strong conclusions from these differences as the over-ground and treadmill studies differed in a few ways. Specifically, the over-ground study was completed with bilateral PPAFOs, and the treadmill study was completed with a unilateral PPAFO. Also in the over-ground study, participants were allowed to walk at their preferred speed (average 0.9 Hz gait cycle), whereas in the treadmill study, the participants were encouraged to walk with a 1 Hz gait cycle to match the test bench controller settings. Generally, the test bench is a reasonable approximation of the PPAFO use during walking, and should be continued to be used for device design and development, likely with validation for each type of study.

2.5.1. Presence of dry ice

One qualitative observation that was recorded during the trials was the evidence of presence of dry ice inside the tank at the end of each standard pneumatic circuit CO_2 trial. Presence of dry ice was determined by shaking the tank and listening for a solid rattling noise on the test bench and treadmill walking tests. It is plausible based on the data collected that the formation of dry ice could be related to the rate of cooling. The total mass consumed values per each trial though did not vary between the trials that had dry ice formation (standard) and those without (recycling), indicating that the amount of dry ice was unlikely to be a usable amount of fuel from the tank.

2.5.2. Other Considerations

Other than the length of time that a given amount of compressed gas will power a portable robotic device, some other considerations might need to be taken into account, such as the overall weight of the filled tank, and the overall price of the system. A 4500 psig 68 cu. in. fiberglass HPA tank filled will 9 oz of N_2 can weigh ~55 oz, where as a filled 9 oz CO_2 tank weighs ~24.5 oz, almost half the total weight of the N_2 system. The cost of HPA tanks is also more than that of CO_2 tanks (approximately \$100 HPA vs. \$30 CO_2), and it can be more difficult to find available retailers or to set up one's own system to fill the HPA tanks for regular use [65].

An important proof of concept in using a pneumatic recycling scheme was demonstrated here. In order to have an application for a recycling scheme, a system must have at least two different pressure needs, in that it has a higher pressure actuation from which the exhaust gas can be captured to power a lower pressure actuation. If a control system can be designed to operate portable pneumatic robotics in this way, then a substantial improvement in any power source longevity can be expected.

2.5.3. Limitations

This study was not without limitations. As with any pneumatic system, there is always the possibility for undetected leaks within the system that can affect outcome measures. This work could have benefited from a few more trials on any condition to improve the statistical power, but the repeatability of the test bench was proven by just two or three reliable trials completed on different days. As with any human subjects testing, there is natural variability in human motion, so it is not expected that a test bench will exactly replicate the human study results.

2.6. **CONCLUSION**

This study concluded that the selection of which compressed gas tanks to use as a fuel sources for a portable pneumatic robot, depends on one's constraints. If a long run time is needed, a recycling scheme to reuse pressurized exhaust gas will greatly extend the lifetime of any fuel source. The downside to using CO_2 is the extreme cooling that happens with constant use that can cause possible component freezing, and the possible formation of dry ice. The HPA tank filled with compressed N_2 does not get as cold as the CO_2 tanks, although the HPA tank is heavier than the CO_2 tanks. Consideration of higher cost and greater difficulty with refilling may limit the use of HPA tanks versus CO_2 tanks. From this study, we are able to conclude that if cost and availability are concerns for the fuel source, than CO_2 is the better option; if pneumatic components or external robot components experiencing a very cold temperature is of significant concern then N_2 or compressed air is the better option. When powering portable pneumatic robots, a more compact and long-lasting fuel source would be ideal, but until such a power supply is developed, fixed volume compressed gas tanks have the ability to provide portable power for extended periods of use.

2.7. **TABLES**

Table 2.1: State Estimation Controller Scheme [27]

Functional Task	**% gait cycle**	**Torque Direction**
1. Initial Contact	0-7	Dorsiflexor
2. Loading Response	7-48	None
3. Forward Propulsion	48-62	Plantarflexor
4. Limb Advancement	62-67	None – Charging[a]
	67-100	Dorsiflexor

[a.] Without recycling, dorsiflexor torque starts at 62% of the gait cycle as there is no need for a charging phase

[b.] Gait cycle timings adapted from Perry [25]

Table 2.2: Participants in walking studies (Treadmill and Over-ground)

	Participants	Age	Height	Weight
Treadmill	N = 4 (3 Male/1 Female)	23.1 ± 3.3 years	179.6 ± 2.0 cm	77.8 ± 3 kg
Over-ground	N = 7 (5 Male/2 Female)	21.1 ± 1.1 years	178.4 ± 5.5 cm	71.5 ± 6 kg

Table 2.3: Data Collection Sensors

Sensor	Model #	Company
Pressure Transducer – 50 psig	2091050PG1M2405P	Setra, Boxborough, MA
Pressure Transducer – 100 psig	AST4000A00100P3B0000	American Sensor Technologies, Inc, Mount Olive, NJ
Pressure Transducer – 150 psig	AST4000A00150P3B1000	American Sensor Technologies, Inc, Mount Olive, NJ
Force Gauge	DFG35-10	Omega Engineering, Stamford, CT
Thermometer	HH303, Type K	Omega Engineering, Stamford, CT
Angle sensor – programmable angle sensor	KMA199E	NXP Semiconductors, San Jose, CA
Data acquisition device	Q8-USB	Quanser, Markham, Ontario, Canada

Table 2.4: Mean (standard deviation) of all longevity experimental trials

	Number of Trials	Run Time (min)	Mass consumed (oz)	Normalized Run Time (min/oz)	Minimum bottle temperature (^oC)	Average rate of cooling ($^oC/min$)
Standard Circuit						
20 oz CO_2 Test bench	n = 2	20.42 (0.24)	17.92 (0.02)	1.14 (0.01)	-49.3 (0.4)	-3.97 (0.40)
9 oz CO_2 Test bench	n = 3	9.26 (0.10)	8.32 (0.46)	1.11 (0.07)	-53.1 (0.4)	-9.87 (0.09)
9 oz (~4300psig) HPA (N_2)	n = 2	12.96 (0.12)	9.12 (0.11)	1.42 (0.03)	8.0 (0.9)	-1.32 (0.05)
9 oz CO_2 Treadmill Walking	n = 4	13.73 (2.55)	8.12 (0.73)	1.69 (0.17)	-48.5 (7.5)	-6.63 (0.86)
20 oz CO_2 Over-ground Walking	n = 7	25.32 (2.65)	19.58 (0.71)	1.29 (0.19)	Data not collected	
Recycling Circuit						
20 oz CO_2 Test bench	n = 2	30.15 (0.15)	17.44 (0.11)	1.73 (0.02)	-44.0 (0.6)	-2.25 (0.02)
9 oz CO_2 Test bench	n = 3	11.46 (0.39)	7.04 (0.18)	1.63 (0.01)	-44.1 (0.2)	-5.78 (0.18)
9 oz (~4300psig) HPA (N_2)	n = 2	16.25 (2.45)	8.64 (0.34)	1.88 (0.21)	7.8 (0.3)	-0.91 (0.17)
9 oz CO_2 Treadmill Walking	n = 4	20.86 (3.40)	7.84 (0.41)	2.66 (0.36)	-44.6 (2.7)	-3.75 (0.37)

Table 2.5: Individual trial values for each testing condition and circuit

Data Source	Pneumatic Circuit	Time (min)	Mass consumed (oz)	Normalized Run Time (min/oz)	Dry Ice formed? (Y/N)	Minimum temp. (°C)	Rate of cooling (°C/min)
Test bench 20 oz CO_2	Standard	20.25	17.95	1.13	Y	-49.5	-4.26
		20.59	17.98	1.15	Y	-49.0	-3.69
	Recycling	30.25	17.28	1.75	N	-44.4	-2.23
		30.04	17.44	1.72	N	-43.5	-2.26
Test bench 9 oz CO_2	Standard	9.35	8.00	1.17	Y	-53.3	-9.93
		9.28	8.00	1.16	Y	-53.3	-9.77
		9.15	8.80	1.04	Y	-52.6	-9.91
	Recycling	11.74	7.20	1.63	N	-43.9	-5.66
		11.02	6.88	1.60	N	-44.3	-5.99
		11.63	7.20	1.62	N	-44.0	-5.70
Test bench N_2	Standard	13.04	8.96	1.46	N	8.6	-1.28
		12.87	9.12	1.41	N	7.3	-1.35
	Recycling	17.99	8.80	2.04	N	7.6	-0.79
		14.52	8.32	1.75	N	8.0	-1.02
Treadmill Walking CO_2	Standard	13.30	8.00	1.66	Y	-51.1	-7.36
		10.44	7.20	1.45	N	-37.4	-5.55
		14.68	8.32	1.76	Y	-53.9	-7.29
		16.49	8.96	1.84	Y	-51.6	-6.33
	Recycling	20.53	8.00	2.57	N	-46.8	-3.84
		23.29	8.32	2.80	N	-46.8	-3.77
		16.17	7.36	2.20	N	-43.4	-4.14
		23.43	7.68	3.05	N	-41.3	-3.26
Over-ground Walking CO_2	Standard	25.0	19.84	1.26	N	Data not collected	
		30.51	17.95	1.70	N		
		24.52	19.68	1.25	N		
		21.92	19.82	1.11	N		
		26.30	19.85	1.32	N		
		24.30	19.99	1.22	N		
		24.63	19.53	1.26	N		

2.8. **FIGURES**

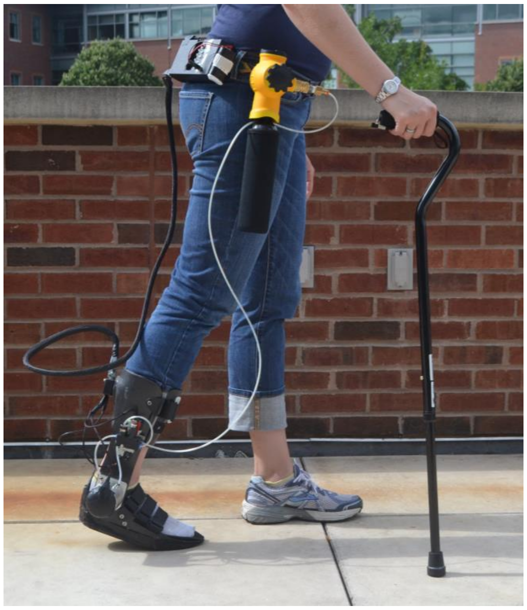

Figure 2.1: **Portable Powered Ankle-Foot Orthosis (PPAFO) with waist worn fuel tank and microcontroller.**

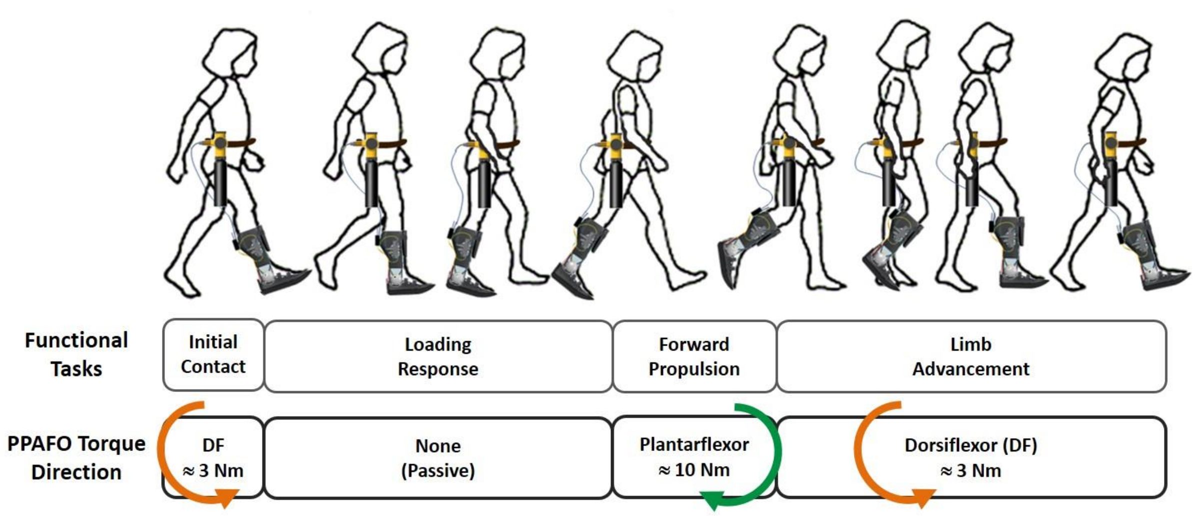

Figure 2.2: **Assistance during a gait cycle provided by the PPAFO.** Torque values are based on 100 psig during plantarflexor assistance and 30 psig during dorsiflexor (DF) assistance.

Adapted and reprinted from Clinical Gait Analysis, Kirtley, C., Introduction, p. 201-22, © 2006, with permission from Elsevier.

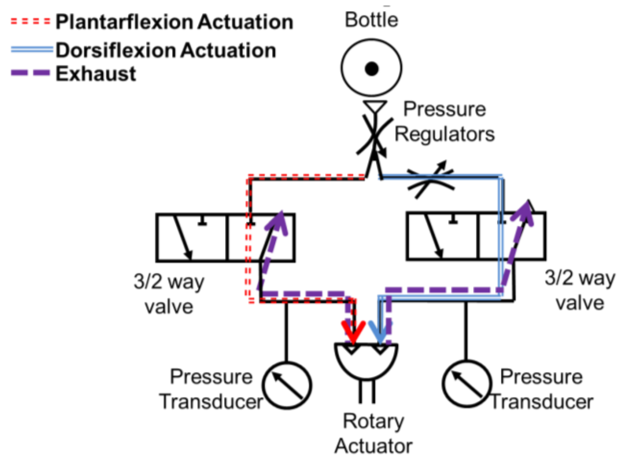

Figure 2.3**: Standard pneumatic power scheme for PPAFO** [27] © 2013 IEEE

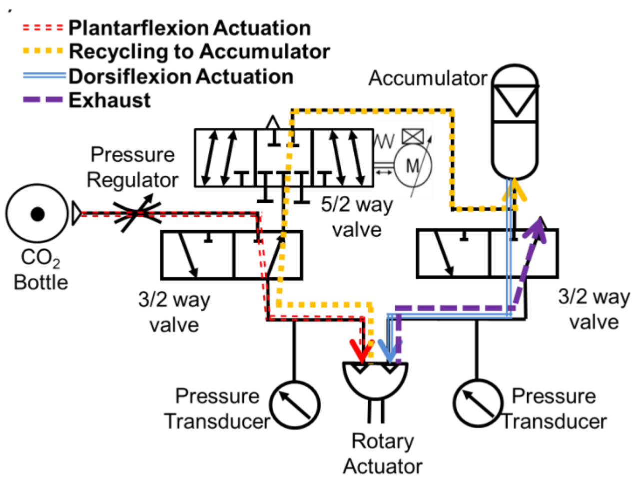

Figure 2.4: **Recycling pneumatic power scheme for PPAFO** [27] © 2013 IEEE

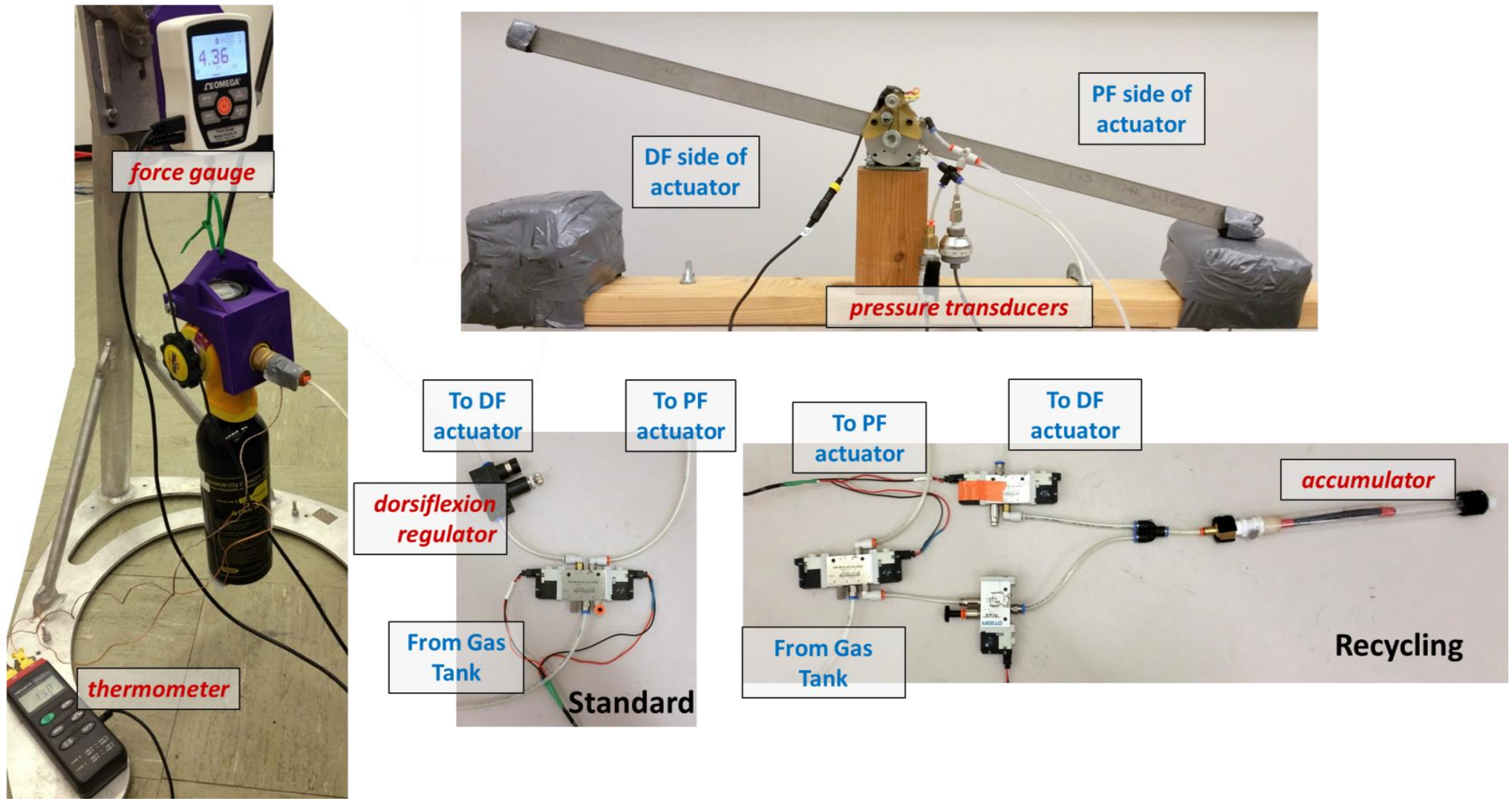

Figure 2.5: **Test bench setup of PPAFO**

Figure 2.6: **PPAFO with accumulator and pressure transducers for longevity testing.** All components were attached to the PPAFO for both pneumatic circuits during treadmill testing, so only pneumatic connections needed to be adjusted between trials. For the over-ground walking trials, the pressure transducers on the back and the accumulator were not attached to the PPAFO.

Figure 2.7: **Fully assembled custom elastomeric strain energy accumulator** used in the recycling circuit. Top image: uninflated accumulator. Middle image: partially inflated accumulated. Bottom image: fully inflated accumulator. The balloon expands and slides to fill the allowed space by the outer shroud.

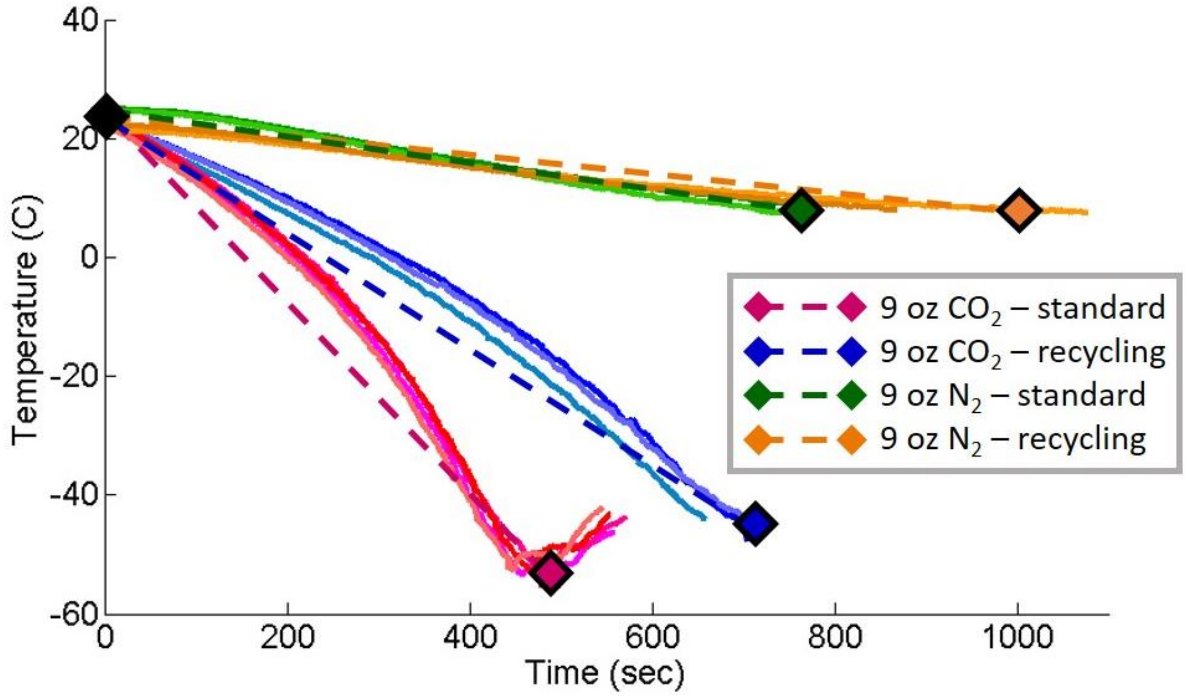

Figure 2.8: **Bottle temperature throughout longevity trial of 9 oz CO_2 and N_2 tanks on test bench.** Rate of cooling was computed by finding slope of the curve between the average starting temperature and the average minimum temperature (dashed lines). Each solid colored line represents an independent trial for a given condition.

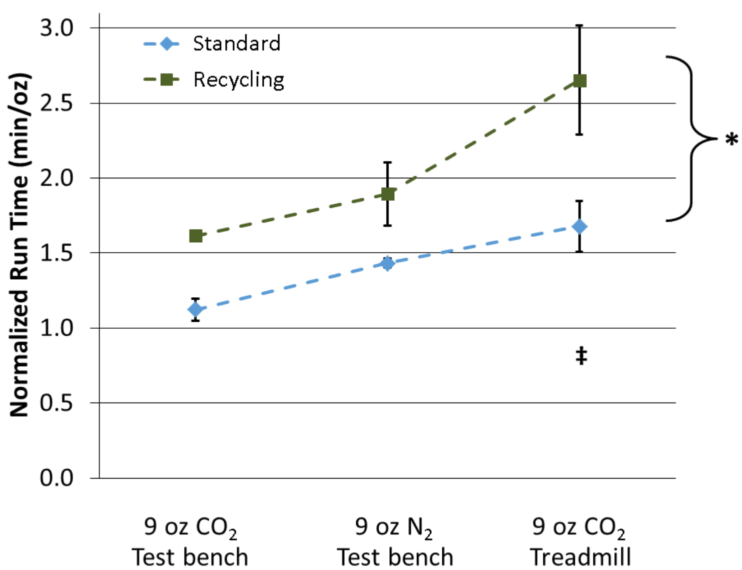

Figure 2.9: **Normalized run time for each 9 oz trial type.** The 9 oz CO_2 treadmill trial had a significantly longer normalized run time than either test bench trial (‡). There was also a significant difference between pneumatic circuits: recycling and standard (*).

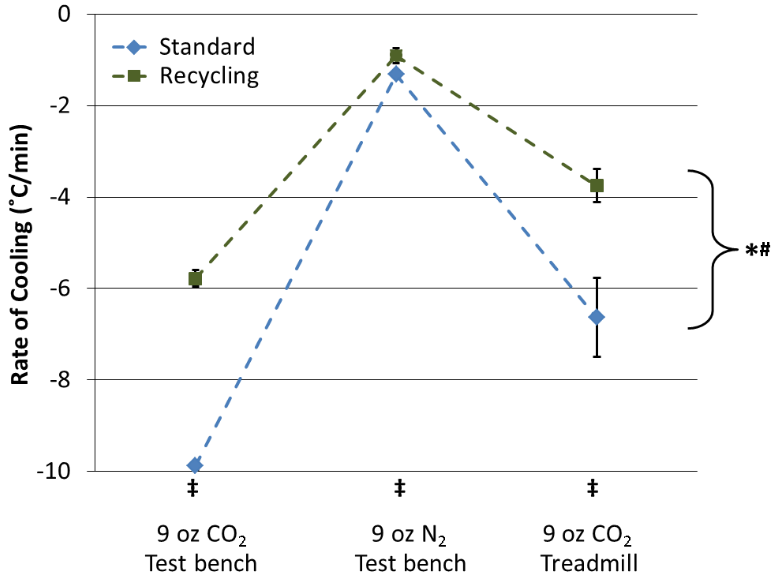

Figure 2.10: **Rate of cooling for each 9 oz tank trial.** All trial conditions (9 oz CO_2 test bench, 9 oz N_2 test bench, and 9 oz CO_2 treadmill) were significantly different from one another (‡). The standard and recycling circuits were also significantly different from one another (*), with a significant interaction effect (#).

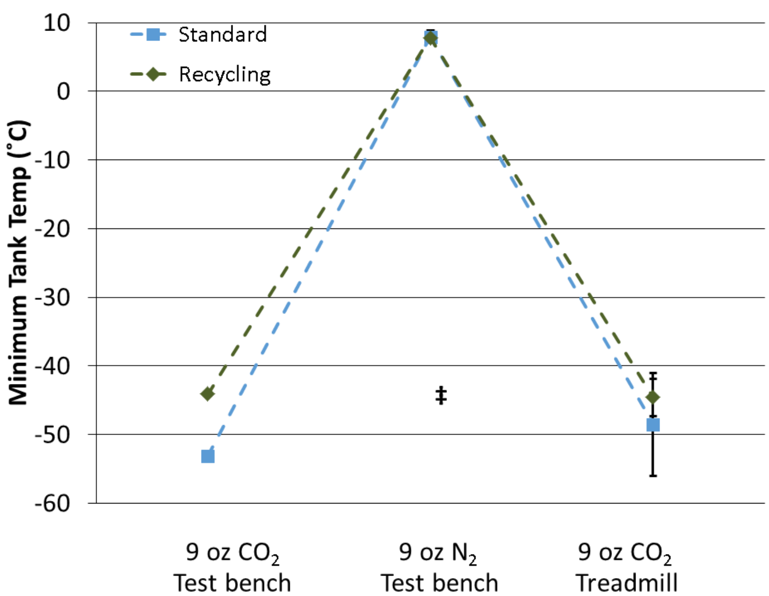

Figure 2.11: **Minimum tank temperature comparing the 9 oz fuel experiments.** The N_2 tank had a significantly warmer minimum tank temperature than the 9 oz CO_2 tank test bench and treadmill trials. There were no significant differences between standard and recycling pneumatic circuits.

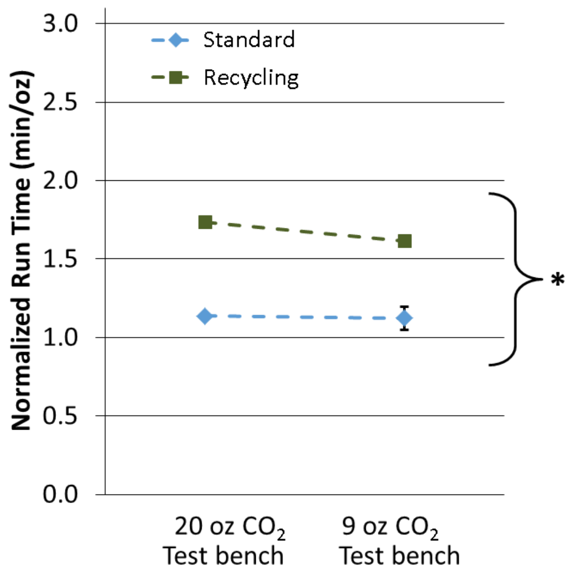

Figure 2.12: **Normalized run time for comparing CO_2 tank sizes.** There is a significant difference between pneumatic circuits (standard and recycling) (*).

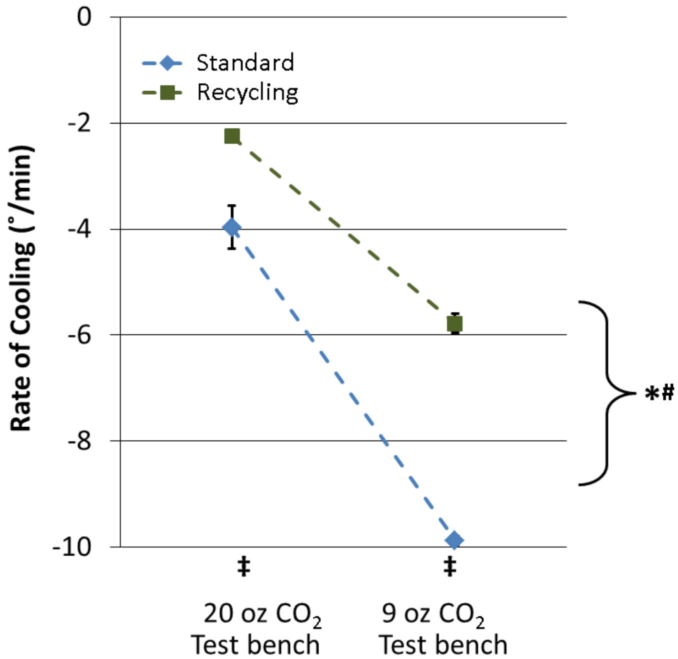

Figure 2.13: **Rate of cooling compared between CO_2 tank sizes.** There are significant differences between tank sizes (‡) and between pneumatic circuits (*), as well as a significant interaction (#).

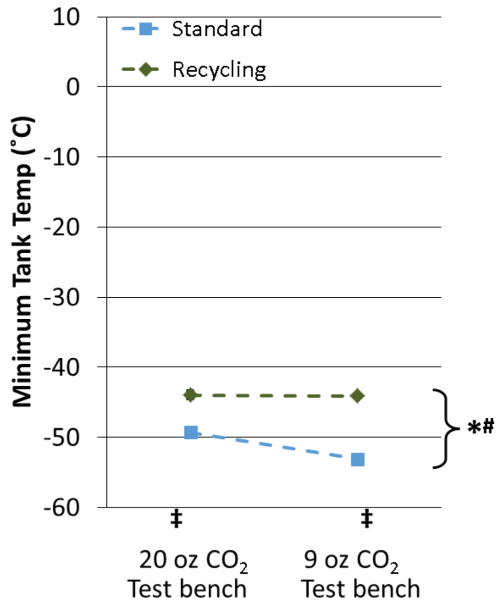

Figure 2.14: **Minimum tank temperature comparing CO_2 tank sizes.** There is a significant difference between tank sizes (‡) and pneumatic circuits (*), as well as a significant interaction (#).

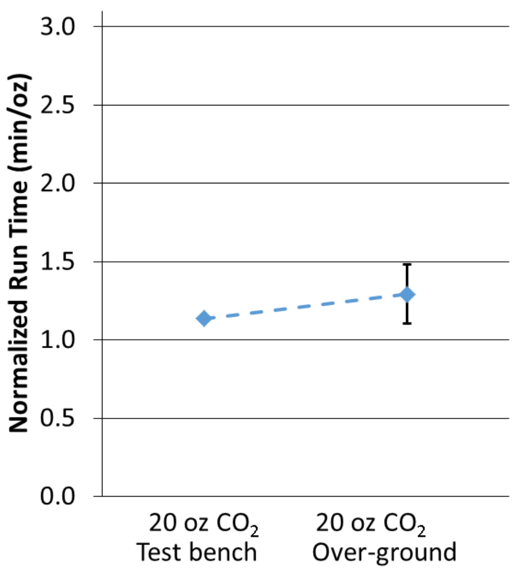

Figure 2.15: **Normalized run time comparing the test bench to over-ground walking.** There was no significant difference between test bench and over-ground conditions. The standard deviation for the test bench is smaller than the marker used.

Supplemental Figures

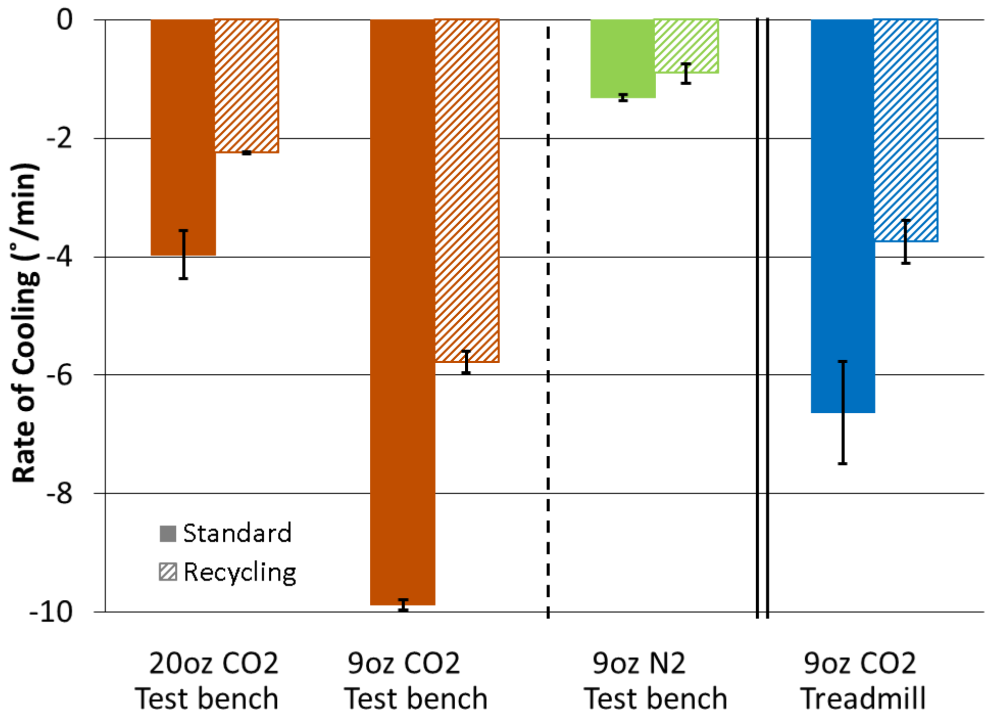

Figure 2.16: **Average linear approximation of rate of cooling for each gas tank type on the test bench and during able-bodied treadmill walking.**

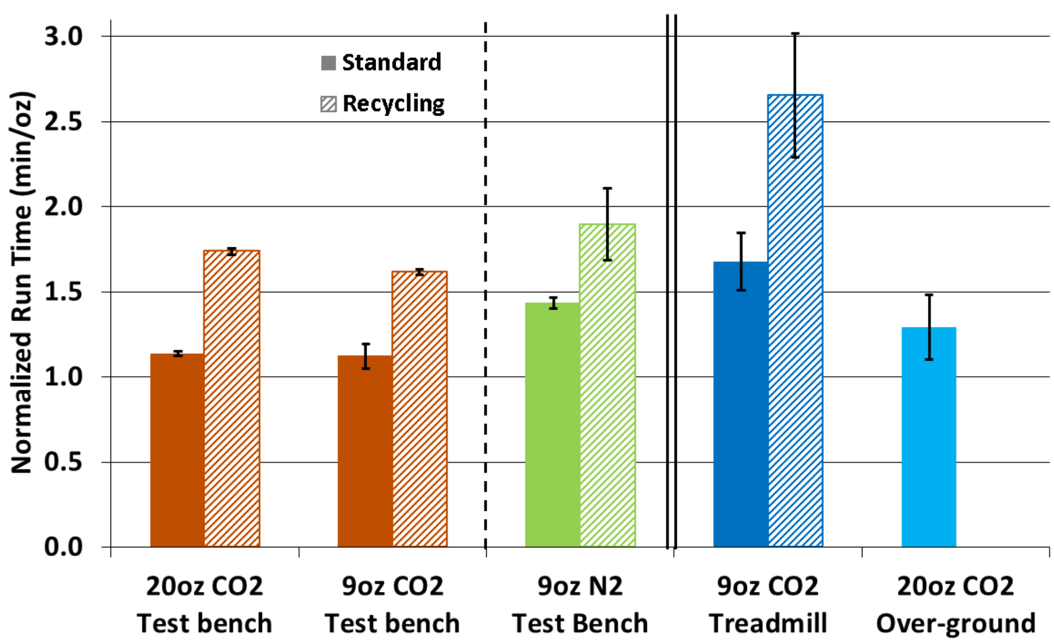

Figure 2.17: **Normalized run time for each longevity trial type.**

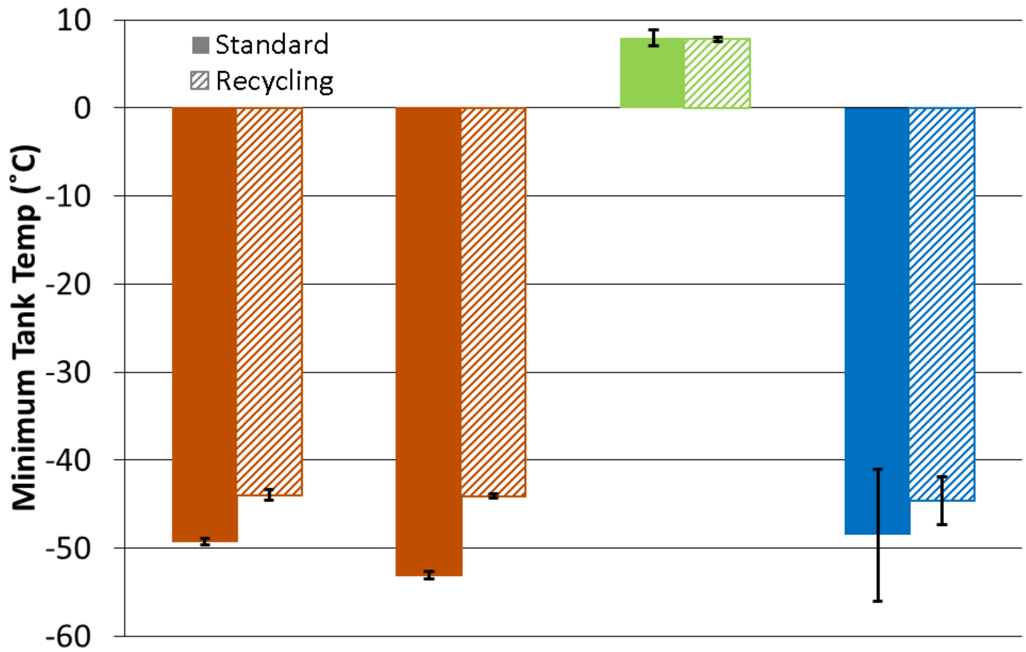

Figure 2.18: **Minimum tank temperature reached for each longevity trial type.**

Chapter 3 : Spatiotemporal, kinematic, and metabolic evaluation of a bilateral, bidirectional powered ankle-foot orthosis in able-bodied over-ground gait

3.1. ABSTRACT

Introduction: Previous research on powered ankle-foot orthoses and exoskeletons have focused mainly on powered devices that apply assistive torque during plantarflexion while walking on a laboratory treadmill. The purpose of this study was to evaluate the short-term kinematic and metabolic impact of bilateral, bidirectional powered ankle assistance via a powered ankle-foot orthosis in an able-bodied population during over-ground walking.

Methods: The portable powered ankle-foot orthosis (PPAFO) device was used in this study. Eight healthy young adults completed three 6 minute walk (6MW) tests during which they wore a portable metabolic unit. During a portion of the path, the participant walked over a gait mat and through a 3D motion capture space. Three footwear conditions were compared: shoes-only, bilateral unpowered ankle orthoses, and bilateral powered ankle orthoses. Outcome measures of 6MW distance, stride velocity, stride length, stride width, metabolic cost of transport, and sagittal-plane knee and ankle angles (peak magnitudes in four subphases of gait) were determined and compared between footwear conditions.

Results and Discussion: With the powered PPAFOs, participants were able to reduce the metabolic cost of transport for walking compared to the unpowered PPAFO condition, and they were able to match the metabolic cost of transport for shoes walking (Powered: 0.30 ± 0.08, Unpowered: 0.36 ± 0.07, Shoes: 0.32 ± 0.03). Kinematic changes were found while using the

PPAFOs, specifically an unexpected reduction in plantarflexion angle during toe-off (Powered: -12.9 ± 7.2°, Unpowered: -18.6 ± 5.6°, Shoes: -27.8 ± 6.6°).

Conclusions: Being able to match the metabolic cost of transport while using the powered PPAFO in over-ground walking compared to shoes is a valuable starting place for being able to further research augmenting gait with new technologies. Yet, some of the unexpected results kinematic changes while using the PPAFO motivate us to further investigate the mechanical design of devices so that users can better match their natural gait pattern in regards to spatiotemporal and kinematic parameters.

3.2. **INTRODUCTION**

Lower limb powered orthoses and exoskeletal systems are currently being developed for a variety purposes ranging from augmenting the abilities of able-bodied individuals, e.g., [1-3], assisting the abilities of those with physical limitations, e.g., [4, 5], and studying the fundamental biomechanics and motor control of normal and pathological gait, e.g., [6, 7]. There are many types of lower limb exoskeletons, based on the number of actuated joints, and type of power system used, e.g., [9, 10]. Generally, complete lower limb devices tend to neglect the ankle even though much of the force needed for gait is transferred through the ankle joint to the ground [11-13]. A recent review of the use of powered lower limb orthoses for gait assistance even mentions the need for devices with powered ankle joints [4].

Some of the most studied powered ankle-foot orthoses (PAFO) designs are those from the University of Michigan (e.g., [5-7, 34, 36-42, 94, 95]). These devices generally provided powered plantarflexion and/or powered dorsiflexion through pneumatic muscles, with EMG-based controls during gait. These tethered designs use a laboratory air compressor and desktop computer to power and control the pneumatic muscles while participants walk on a laboratory treadmill at constant speeds. Studies have investigated unilateral device use and uni- (or bi-) directional powered actuation to investigate motor adaptation in gait of able-bodied persons [6, 37].

Other researchers have adopted the Michigan PAFO design with a pneumatic muscle to create plantarflexor torque and have explored the effect of added plantarflexor torque to the metabolic cost of walking in young able-bodied participants. At Virginia Tech, a team of researchers used a bilateral set of kinematic controlled PAFOs [43, 44]. The data for the timing of the custom control algorithm were based on the angular velocity of the foot section of each orthosis during walking. In maintaining a tethered design, Norris et al. [44] investigated the impact of providing additional plantarflexion via the PAFO while walking on a treadmill on the metabolic cost of transport and preferred walking speeds in young and older adults. Researchers at Gent University used a kinematic controller based on a footswitch located under the heel in a series of bilateral PAFO studies on their Wearable Assistive Lower Leg eXoskeleton (WALL-X) design [2, 16, 45-47]. One study aimed to determine the ideal timing of turning on the powered plantarflexion in regards to gait cycle time based on reducing the metabolic cost of walking [16].

Recently, a novel ankle exoskeleton was developed by researchers at MIT specifically to reduce the metabolic cost of walking in able-bodied individuals [3, 48, 49]. This ankle exoskeleton provided a large plantarflexor torque (approximately 120 Nm [3]) during push off using an electrical winch actuator and strut. The ankle exoskeleton allowed free movement during swing by providing slack in the drive cord such that the user did not notice any assistance or impedance from the exoskeleton. The plantarflexor torque timing was controlled kinematically by gyroscopes on the actuator, which provided angular shank velocity data to the controller. With onboard controllers at the waist, this untethered system was still only tested on a treadmill at fixed speeds [3, 49].

Although much work has been done in the area of powered ankle-foot orthoses and exoskeletons, so far the studies done with these devices have been limited to laboratory environments with treadmill walking. As most of human walking occurs over-ground, it is necessary to begin to study the impact of these devices in less strict walking environments.

Researchers at the University of Illinois developed the first untethered PAFO that provides actuation in both plantarflexion and dorsiflexion using a bidirectional pneumatic rotary actuator at the ankle (e.g., [22-24, 27, 91, 96, 97]). Pneumatic power can be provided in a portable system by a waist-worn tank of compressed gas. Modest plantarflexor torque of 9-12 Nm can be generated using an input pressure of 90-120 psig, and a dorsiflexor torque of ~3 Nm was heuristically tuned to support the foot during swing by down regulating the input pressure to ~30 psig [91, 98, 99]. The Portable Powered Ankle-Foot Orthosis (PPAFO) can been kinematically controlled with an on-board microcontroller using foot switches and ankle angle sensor. The PPAFO testbed explored multiple approaches to determine the current phase of gait to control plantarflexor and dorsiflexor torque timing with either direct measurement of gait events [22] or algorithms to estimate the state of the system as a function of gait cycle [23, 24, 96]. As a portable gait assistance device, studies have also been conducted to understand control while walking in multiple environments (level-ground, stairs, ramps) [91] and methods to improve fuel efficiency [27].

The purpose of this study was to evaluate the short-term kinematic and metabolic impact of a bidirectional powered ankle-foot orthosis in an able-bodied population during over-ground walking that is not restricted to walking on a treadmill. The PPAFO was worn bilaterally in young healthy adult participants. We hypothesized that use of the powered PPAFO would reduce steady-state metabolic cost compared to the unpowered PPAFO and shoes only, due to the bi-directional powered assistance provided by the PPAFO. We also hypothesized that there would be greater dorsiflexion during swing and a greater plantarflexion at toe off with the powered condition compared to unpowered condition and shoes only. Finally, we hypothesized that participants would have greater over-ground gait speed (stride velocity) when walking in the powered condition compared to the unpowered condition, and a slower gait speed with the unpowered condition as compared to shoes only. As there is little previous research on bidirectional assistance at the ankle during able-bodied gait, we hope to learn about the importance of powered dorsiflexion as well as plantarflexion assistance, especially as related to gait pathologies with lower limb weakness.

A secondary exploration within the study was completed which specifically addressed adaptation during walking with the PPAFO. The amount of training and adaptation time needed to optimize the use of a powered exoskeleton device is an area of current research in the field [37, 38, 46, 47, 94, 95, 100]. Much of the research has shown mixed results in the amount of time needed for neuromuscular adaptation, with values ranging from 5 to 25 minutes of controller training or walking practice needed. Higher powered and EMG controlled devices seem to require more time to adaptation than lower powered and kinematically controlled devices [7, 38]. We conducted a preliminary investigation into kinematic or metabolic adaptation changes throughout 20 minutes of walking, which occurred after a controller training period. As our device is lower powered and kinematically controlled compared to previously published exoskeleton designs, we hypothesized, that a shorter (less than 20 minutes) amount of time will be necessary for adaptation.

3.3. METHODS

3.3.1. Portable Powered Ankle Foot Orthosis

The PPAFO can provide dorsiflexor or plantarflexor torque via an untethered pneumatic power source; thus allowing for gait assistance away from confined spaces or treadmill walking (Figure 3.1). Torque is delivered to the ankle through a standard pneumatic circuit: two solenoid valves (VUVG 5V; Festo Corp-US, Hauppauge, NY) control a rotary actuator (PRN30D-90-45, Parker Hannifin, Cleveland, OH). The rotary actuator was powered directly from a portable compressed CO_2 tank (20oz, Catalina Cylinders, Hampton, VA) with a bottle top regulator (JacPac J-6901-91, Grainger, Inc). The valves and controller were powered by a portable battery pack.

The PPAFO was designed with interchangeable sizes in the shank and foot bed fiberglass shells to create a custom fit with padding for every participant. The PPAFO also had an orthotic rocker sole to aid in natural gait with a hard sole orthotic [101-103]. With the pneumatic valves, actuator, and other necessary hardware, each PPAFO has a total weight of near 1.8 kg. The electronics controller and battery as well as the portable pneumatic tank were worn at the waist and weigh approximately 1.5 kg. The design of the PPAFO was intended to keep at much weight as possible centrally located on the body, near the wearer’s center of mass [104, 105].

The applied torque of the PPAFO can provide directional assistance at the ankle as needed throughout all phases of the gait cycle (Figure 3.2). The PPAFO was controlled by a modified fractional time state estimation controller (SE) [24], where the current state of the gait cycle was estimated by sensor input from heel and toe contact sensors on the sole of the PPAFO, as well as a Hall effect sensor that measured ankle angle. The SE controller provided actuation during four functional gait tasks: (1) initial contact, (2) loading response, (3) forward propulsion, and (4) limb advancement (Figure 3.2).

For this study, a bilateral PPAFO system was used (Figure 3.3). Each PPAFO was run from its own controller and fuel source. The PPAFO controllers in the bilateral system were programmed and run independently from one another.

3.3.2. Participants

The study included eight healthy young adult participants (Table 3.1) with no previous experience testing the PPAFO. Inclusionary criteria included wearing a men's shoe size 5-14 and having no severe or recent lower limb injuries. Exclusionary criteria included significant cardiac or respiratory problems, and the inability to walk without assistive devices. Approval for the study was granted by the Institutional Review Board and participants gave informed consent.

3.3.3. Testing Protocol

Participants first wore the PPAFOs for a training period where the PPAFO controller was trained for an individual's gait pattern. The training period took approximately 20 minutes; during that time, the participant walked sporadically with the PPAFO for a cumulative walking time of ~10 minutes. To start the training period, the PPAFO controllers were set to baseline assistance timing based on normative gait timing values [25]. The PPAFOs were trained one at a time, such that the PPAFO (side) that was not being trained was powered with the most recently update program timings (initially, the baseline assistance was used). The participants were instructed to walk as naturally as possible with the powered PPAFOs. For each PPAFO's controller training, the participants walked for more than 10 continuous test steps during which heel, toe, and angle sensor data from that PPAFO were collected. The PPAFO sensor data were then evaluated to determine the ideal timing for each participant based on the participant's pattern of sensor activation (gait pattern) [24]. The PPAFO controller was then updated with the new actuation timings. After one PPAFO controller was initially updated, then the other side PPAFO was programmed in the same manner as the first. The participant was asked to repeat the aforementioned process such that the programming would be repeated for each side. The controller adjustment process was repeated at least three times for each PPAFO, alternating between the controller for each side. Verbal feedback from each participant was also used to confirm comfortable actuation timings. After the controller timings were set, each participant finished the training period by walking until he/she felt comfortable with the actuation of the PPAFO; at a minimum, each participant completed one loop around the 9 m x 12 m hallway while acclimating to walking with the PPAFOs.

Participants then completed three 6-minute walk (6MW) tests that were performed over-ground in a 9 m x 12 m hallway loop with four 90-degree turns (Figure 3.4). Instructions were given to the participant to walk as fast as possible within the limits of their safety for the entirety of the 6MW. With the PPAFO trials (unpowered and powered), participants were also instructed to walk as naturally as possible. One 6MW was completed per footwear condition: personal shoes, unpowered bilateral PPAFOs, powered bilateral PPAFOs. The trials were always completed in the same order such that the shoes trial was always completed first for a baseline measurement, and the powered PPAFO condition was always complete last so that the participants could continue walking past the 6MW until the portable power sources for the PPAFOs ran out of fuel (determined by cessation of pneumatic exhaust sounds from control valves) as part of another study (Chapter 2). For both the powered and unpowered trials, participants donned both the PPAFOs as well as the fuel tanks and controllers that were worn at the waist.

3.3.4. Outcome Measures/Data Analysis

Spatiotemporal, kinematic, and metabolic parameters of gait were measured during each 6MW. During each 6MW, spatiotemporal parameters of gait were measured as the participant passed over an 8.9 m pressure sensitive walkway (GAITRite, CIR Systems Inc., Sparta, NJ) with each lap (≥ 4 passes per 6MW test). Bilateral values of stride length (cm), stride width (cm), and stride velocity (cm/s) for each participant were determined from the software (GAITRite 4.0). Averaged values were then computed from the values recorded during the final 3 minutes of the 6MW. Total distance traveled during the 6MW was recorded with a measuring wheel (RT312, Rolatape, Watseka, IL).

During each 6MW, rates of oxygen consumption ($\dot{V}O_2(\mathrm{mL/min})$) and carbon dioxide production ($\dot{V}CO_2(\mathrm{mL/min})$) were measured breath-by-breath using a commercially available portable metabolic unit (K4b2, Cosmed S.r.l., Rome, Italy). Net values for $\dot{V}O_2$ and $\dot{V}CO_2$ were computed using 30-second moving averages for 1 minute before the 6MW (resting) and over the entire 6MW (walking) using the Cosmed software. Steady-state values then were computed from averaging walking values for the final 3 minutes of the 6MW. Net values were computed by subtracting average resting values from the steady-state values (Eqns. 1 and 2) [86].

$$\left[\dot{V}O_2\right]_{\mathrm{net}}(\mathrm{mL/min}) = \left[\dot{V}O_2\right]_{\mathrm{steady-state}} - \left[\dot{V}O_2\right]_{\mathrm{resting}} \tag{1}$$

$$\left[\dot{V}CO_2\right]_{\mathrm{net}}(\mathrm{mL/min}) = \left[\dot{V}CO_2\right]_{\mathrm{steady-state}} - \left[\dot{V}CO_2\right]_{\mathrm{resting}} \tag{2}$$

Metabolic cost of transport (COT_m) was computed to quantify the energy expenditure during walking [40, 41, 43, 86]. A modified Brockway equation [84] with updated coefficients by Adamczyk et al. [87] was used to calculate COT_m from the net $\dot{V}O_2$ and $\dot{V}CO_2$ as suggested by Donelan et al. [86] (Eqn. 3). Gait speed was computed as the average gait speed over the last three minutes of the 6MW, by dividing the distance covered in that time by 3 minutes.

$$\text{Metabolic Cost of Transport } (\mathrm{COT_m}) = \frac{16.477\ (\mathrm{W \cdot s/mL}) * \left[\dot{V}O_2\right]_{\mathrm{net}}(\mathrm{mL/min}) + 4.484\ (\mathrm{W \cdot s/mL}) * \left[\dot{V}CO_2\right]_{\mathrm{net}}(\mathrm{mL/min})}{60\ \left(\frac{\mathrm{s}}{\mathrm{min}}\right) * \text{body weight (N)} * \text{gait speed (m/s)}} \tag{3}$$

COT_m was normalized by body weight. In the PPAFO conditions (unpowered and powered), an additional 6.6 kg was added to the participant's body weight (1.8 kg for each PPAFO and 1.5 kg for each compressed gas tank and controller).

Lower body movements were recorded using a motion capture system for a portion of each loop around the 6MW path (Oqus Motion Capture Camera System, Qualisys, Highland Park, IL) (Figure 3.4). A lower body 23 static marker set was used with markers on the ASIS, greater trochanter, mid-thigh, lateral knee epicondyle, medial knee epicondyle, tibia, lateral malleolus, medial malleolus, heel, toe 1, and toe 5 for both the right and left limbs, as well as a sacral marker. In the trials where the PPAFOs were worn, an extra marker was placed on the lateral point of the actuator in line with the plantarflexion/dorsiflexion axis of the ankle (Figure 3.3).

The sagittal plane joint angles (ankle, knee) were determined from the motion capture data. The marker data were filtered (4^{th} order Butterworth filter, 12Hz cutoff frequency), and then using inverse kinematics subject-specific models calibrated to a static standing pose, the joint angles were computed (Visual3D v5, C-Motion, Germantown, MD). Heel strike and toe off events were determined for each gait cycle by visual inspection of the motion data. Each participant completed at least one gait cycle for each pass through the motion capture space. Data analysis was completed on the passes through the motion capture space during the final three minutes of each 6MW; each participant completed at least 4 passes per final three minutes. To compare ankle and knee angles between footwear conditions, for each gait cycle, the local minimum or maximum peak magnitude and peak timing were found during four subphases of gait: loading response, terminal stance, pre-swing (near toe-off), and terminal swing [25].

A secondary exploration was completed which specifically addressed adaptation during walking with the PPAFOs. All participants were asked to complete an additional 14 minutes of walking after completing the 6MW, for a total of 20 minutes of continuous over-ground walking. The final three minutes (minutes 18-20, “Powered-Late”) of walking were compared to three last minutes from the 6MW test (minutes 4-6, “Powered”). All of the spatiotemporal, metabolic, and kinematic parameters were collected and analyzed as described above during this extended walking time period. One of the eight participants chose to end the powered PPAFO walking trial after completing the 6MW due to uncomfortable fit of the PPAFOs (Table 3.1).

3.3.5. Statistical Analysis

A series of repeated measures MANOVAs were run to assess the effect of footwear condition on metabolic, spatiotemporal, and kinematic data (SPSS v22, IBM Corporation, Armonk, New York). First, spatiotemporal and metabolic measures of COT_m, stride velocity, stride length, stride with, and 6MW distance were compared across footwear conditions (shoes, unpowered, powered) with a repeated measures MANOVA. Then the same parameters were evaluated for adaptation (powered, powered-late) with a repeated-measures MANOVA. Kinematic parameters of ankle angle peak magnitudes and timing and knee angle peak magnitudes and timing were also compared across footwear conditions (shoes, unpowered, powered) and for adaptation (powered, powered-late) using four repeated measures MANOVAs. Follow-up post-hoc tests (Bonferroni) were examined on significant variables to identify significantly different footwear conditions. Level of significance was set to $\alpha = 0.05$.

3.4. **RESULTS**

The first statistical comparison found differences in COT_m, stride velocity, stride length, stride width, and 6MW distance dependent on footwear condition. The MANOVA for the metabolic and spatiotemporal parameters found overall significance for differences in footwear conditions ($p < 0.001$). Metabolic cost of transport was significantly greater during unpowered PPAFO walking than powered PPAFO walking ($p = 0.012$, Table 3.2, Figure 3.5). Stride velocity ($p < 0.001$), stride length ($p = 0.004$) and 6MW distance ($p < 0.001$) were significantly greater in the shoes condition than in the unpowered or powered PPAFO conditions. Both the stride velocity and 6MW distance were greater in the unpowered condition than the powered condition (Table 3.2, Figure 3.5). For stride width, the shoes condition was significantly narrower than the unpowered or powered conditions ($p = 0.005$).

Lower limb kinematics during the final 3 minutes of the 6MW tests were compared across footwear conditions (Figure 3.6). For ankle angle, differences were seen in both peak timing and magnitude between footwear conditions (MANOVA, $p < 0.001$). The plantarflexion peak during loading response was sooner ($p = 0.022$) in the unpowered condition than either the shoes (1% GC later) or powered conditions (2 %GC later). At the dorsiflexion peak at terminal stance, it was observed that the shoes condition peak occurred significantly sooner ($p < 0.001$) than the unpowered (4.1 %GC later) or powered (6.4 %GC later) conditions, and the unpowered condition peaked significantly sooner (2.3 % GC earlier) than the powered condition. The plantarflexion peak magnitude at pre-swing was significantly greater ($p < 0.001$) in the shoes condition than the unpowered (9.2^o less) or powered conditions (14^o less). The dorsiflexion peak during terminal swing was significantly later ($p = 0.002$) in the shoes condition than the unpowered (5.5 %GC earlier) and powered conditions (7.9 %GC earlier) , and the dorsiflexion peak magnitude at terminal swing was significantly greater ($p = 0.008$) in the powered condition than the shoes (6^o less) or unpowered condition (6.8^o less) (Figure 3.6, Table 3.3).

For knee angle, only timing of the peak magnitude were significantly different between footwear conditions (MANOVA, $p = 0.003$) . For the peak at loading response, the unpowered PPAFO condition was significantly earlier than the powered or shoes conditions ($p = 0.033$). For the terminal stance ($p < 0.001$) and mid-swing peaks ($p < 0.001$), the powered PPAFO condition was significantly later than the shoes or unpowered PPAFO conditions (Figure 3.6, Table 3.3).

When examining the effect of adaptation to the PPAFO over 20 minutes of walking in the spatiotemporal and metabolic parameters of gait, no significant differences were seen between minutes 4-6 (Powered) and minutes 18-20 (Powered-Late) of the powered PPAFO walking condition (MANOVA, $p = 0.653$, Table 3.2, Figure 3.5). There were also no significant differences found in ankle (MANOVA, $p = 0.344$) or knee (MANOVA, $p = 0.139$) angles between minutes 4-6 or 18-20 of powered PPAFO walking (Figure 3.6).

3.5. DISCUSSION

The purpose of this study was to evaluate the short-term kinematic and metabolic impact of bilateral ankle exoskeletons in an able-bodied population during over-ground walking. We evaluated the impact of the bilateral PPAFOs in a 6-minute walk (6MW) test environment where each participant completed one 6MW per footwear condition: personal shoes, unpowered PPAFOs, and powered PPAFOs. Each participant also was asked to complete continued over-ground walking for the preliminary adaptation study with the powered PPAFOs past the 6-minute time if they were able. Seven out of 8 participants completed 20 minutes of walking with the powered PPAFOs (Table 3.1).

We hypothesized that in the powered PPAFO condition, participants would have a reduced metabolic cost of transport, compared to the unpowered PPAFO and shoes conditions, due to the bi-directional powered assistance provided by the PPAFOs (Table 3.2). The results of this work indicate that as hypothesized, in the powered PPAFO condition, participants used less net metabolic power than in the unpowered condition. There was no statistically significant difference between the shoes and powered PPAFO condition in this study. These results indicate that the powered PPAFOs provide some energetic assistance and are able to reduce the COT_m from the unpowered PPAFOs, but they are not better than able-bodied walking with shoes. Given the weight (6.6 kg) and design (rocker bottom) of the PPAFOs, it is substantial to point out that in the powered condition, being as good as a normal walking shoes condition in regards to metabolic power is an achievement. A device that does not require more energy to walk is a good starting point for future research regarding optimizing the PPAFOs for human augmentation, whether in able-bodied or impaired individuals.

We hypothesized that participants would have a faster over-ground gait speed when walking with the powered PPAFOs compared to the unpowered PPAFOs, and a slower gait speed with the unpowered PPAFOs compared to shoes. Contrary to our hypothesis, the results of this study indicated that the powered PPAFOs caused participants to walk with a slower stride velocity than the unpowered PPAFO condition, and the shoes condition allowed for a faster stride velocity than both the unpowered and powered PPAFO conditions (Table 3.2, Figure 3.5). The stride velocity was correlated to the total distance walked in the 6MW. The participants walked the furthest in the shoes condition, and the participants walked further in the unpowered PPAFO condition than in the powered PPAFOs condition. When comparing stride length values, the shoes condition had a significantly longer stride length than either the unpowered or powered PPAFO conditions. The final spatiotemporal parameter that was investigated was stride width, in which the shoes condition had a significantly smaller stride width than either the unpowered or powered PPAFO conditions. All of the changes above, generally walking with smaller and slower length steps with a wider base (stride length), point to the participants reacting to feeling unstable while walking with the PPAFOs. It is possible that in a study with a unilateral PPAFO, participants would likely have felt more stable, compensating with their unassisted limb in learning how to best walk with the PPAFO. In future studies, one could possibly measure lower limb electromyography on the plantarflexor and dorsiflexor muscles to see if muscle coordination was timed with PPAFO actuation. It is possible that some of the unexpected differences, such as slower velocity with the PPAFO could be explained by oppositional muscle coordination, where the participant is acting against the PPAFO action.

We hypothesized that participants would have a larger peak dorsiflexion angle during swing and larger peak plantarflexion angle at toe off in the powered condition compared to the unpowered and shoes conditions. As hypothesized, the powered condition provided the greatest dorsiflexion angle during swing (Table 3.3, Figure 3.6). Contrary to our hypothesis, the powered condition provided a smaller plantarflexion ankle at toe off than the shoes condition, and there was no significant difference between the powered and unpowered conditions at toe-off. It is unclear at this time why the powered PPAFO did not produce a larger plantarflexion peak than the other footwear conditions, especially the unpowered condition.

In terms of timing of the peak dorsiflexion and peak plantarflexion angles, the powered and unpowered conditions changed the timing of the ankle angle peaks from the shoes condition (range from 1-6 %GC difference, Table 3.3). It is likely that these timing differences were due to multiple factors including the programmed actuation timings of the PPAFOs (for the powered condition), the added distal mass of the PPAFOs, and the rocker-bottom sole of the PPAFOs. Even though the controller was programmed specifically for each participant's gait, it is still possible that the participant became part of a feedback system in which, as we changed the controller to match the participant's gait, the participant subconsciously changed their gait pattern to match the actuation timing of the PPAFO. At a certain point in training, the controller timing and the participant's timing converged so that the participant was able to feel comfortable walking, but it is impossible to determine from this study how much the controller timing dictated the participant's gait pattern timing. Other researchers have put time and effort into determining when the best time is to actuate a powered exoskeleton during treadmill walking [16], with increasing attention to varying the gait speeds [96], it is unclear if their timing scheme would be applicable in over-ground walking.

We also investigated sagittal plane knee kinematics to see if changes were seen in the knee between footwear conditions. No significant changes were observed in peak knee angles, but statistically significant changes were found in the timing of the peaks throughout the gait cycle (< 3 %GC differences, Table 3.4). Although these changes in timing were statistically significant, the changes were not very large and not likely clinically significant. These timing changes may be just compensatory responses to the changes in ankle kinematics.

In regards to net metabolic power, it is worth noting that, in our preliminary adaptation investigation to compare minutes 4-6 to minutes 18-20 of walking with the PPAFO, no significant difference was found at the later time point. The first possibility to explain the lack of difference is that the true metabolic steady-state with the powered PPAFOs was reached in minutes 4-6 and no further metabolic adaptation could occur. One could argue an alternate interpretation such that metabolic adaptation was still occurring, but 20 minutes of walking was not long enough to elicit a significant change in metabolic power and a longer bout of continuous walking is needed to elicit a further reduction in metabolic power needed to walk. Previous research shows that it takes anywhere from 7-18 minutes to metabolically adapt to walking with a powered exoskeleton [43, 46]. Previous research regarding the type of controller used and the force provided by the ankle exoskeleton also comments on the time needed to adapt, with a lower-powered, and kinematically controlled exoskeleton both taking a shorter amount of time to adapt [7, 38]. Since our device is lower powered than those previously used in research and it uses a kinematically based controller, it is reasonable that it fits in with previous research such that participants were able to reach a steady state by the end of 6 minutes of walking.

There were no significant changes in ankle or knee kinematics in the powered PPAFOs condition between minutes 4-6 and minutes 18-20 of over-ground walking. Similar to the COT_m, this result may be due to that the participants had already reached a kinematic adaptation to the PPAFOs by the 4-6 minute mark, such that there was no significant change at 18-20 minutes. Another possibility is that the participants had yet to adapt their kinematics completely and were still slightly adjusting their movement, just not enough for statistical significance.

One confounding factor of the study design that was not taken into account in the analysis of the spatiotemporal, kinematic, and metabolic changes was the ordering of the footwear conditions. Each participant completed the shoes condition first, followed by the unpowered condition, and then the powered PPAFO condition. Although each participant was given at least 10 minutes rest between 6MW tests, there still could have been fatigue or other factors from the previous walks that influenced the later walks.

It is possible that some of these spatiotemporal, kinematic, and metabolic changes were due to two attributes of the PPAFOs: the rocker bottom and the mass of the device (1.8 kg per PPAFO). Previous research has commented on the impact of adding mass to the distal end of the leg during gait [104, 105], such that additional mass as the ankle increases the metabolic parameter of choice (oxygen consumption rate [104] and net metabolic rate [105]). Barnett et al. [104] reported that adding 1.82 kg unilaterally did not changed the rate of oxygen consumption or the walking speed. Adding 3.73 kg bilaterally (1.82 kg per leg) resulted in an increased swing time. A linear regression analysis showed that oxygen consumption rate versus added weight was a significant relationship. Browning et al. [105] found similar trends, but evaluated 4 kg (2 kg/limb) as the smallest added mass, which is greater than the mass of our PPAFOs. The rocker-shape of the PPAFO sole also possibly had some impact on the gait of these participants as a rocker bottom sole is known to change the kinematics of gait, and possibly impact spatiotemporal and metabolic parameters of gait as well [87, 101-103].

One unique aspect of this study was the use of over-ground walking assessment. Most lower body exoskeletal work has been evaluated on treadmills. Completing this study in an over-ground walking environment provided a more realistic use scenario, which had the added advantage or perhaps disadvantage of introducing greater variability into the study. Participants were allowed to easily change gait speed whenever they desired, whereas gait speed is generally controlled on a motorized treadmill. Over-ground walking also allowed for any natural changes in gait kinematics to happen at the preferred speed based on footwear as opposed to at a specified, fixed speed. Previous research has indicated that there are significant differences between over-ground and treadmill walking [106], and it is important that lower-limb devices are evaluated in the task for which their end use is designed.

3.5.1. Limitations

This study was limited by a few factors, including the small and limited participant number. Some of the participant's gait might have been impacted by the fit and comfort of the PPAFO devices. Even though the devices were sized and fit to the participant, they were not custom fit and designed for each participant. Also the participants were asked to carry the controllers and fuel bottles for the PPAFOs during the unpowered and powered PPAFO conditions, whereas exoskeletons in treadmill studies are tethered to off-board power supplies. This adding weight at the waist (3.0 kg, two full CO_2 fuel tanks and regulators and control boxes) may have impacted the gait parameters evaluated here. Also as with any human subject study, there is always inherent variability in how each participant completed the task. For this study, it is possible that our results would have been more consistent if each participant would have had more time training with the device and practice walking with the PPAFOs. Although participants were given training time, it is also possible that more familiarity with the device would have produced different outcomes.

3.6. **CONCLUSIONS**

In conclusion, this study evaluated a bilateral powered ankle exoskeletal system in an over-ground walking environment with able-bodied young adults. The bilateral PPAFO system was programmed to match subject-specific gait patterns. With the powered PPAFOs, participants were able to reduce the metabolic power needed for walking compared to the unpowered PPAFO condition, and they were able to match the minimum metabolic power needed in shoes walking. Some kinematic changes were found while using the PPAFOs, primarily at the ankle such that there was an unexpected reduction in plantarflexion during toe-off. Some of the unexpected results motivate us to further investigate the mechanical design of the device so that users can better match their natural gait pattern in regards to spatiotemporal and kinematic parameters. In summary, being able to match the metabolic cost of transport while using the powered PPAFOs in over-ground walking to that used while wearing shoes is noteworthy. The differences in COT_m, while still impacting spatiotemporal and kinematic parameters is a good starting point for continually improving and developing new technologies for gait augmentation.

3.7. **FIGURES**

Figure 3.1: **Portable Powered Ankle-Foot Orthosis**

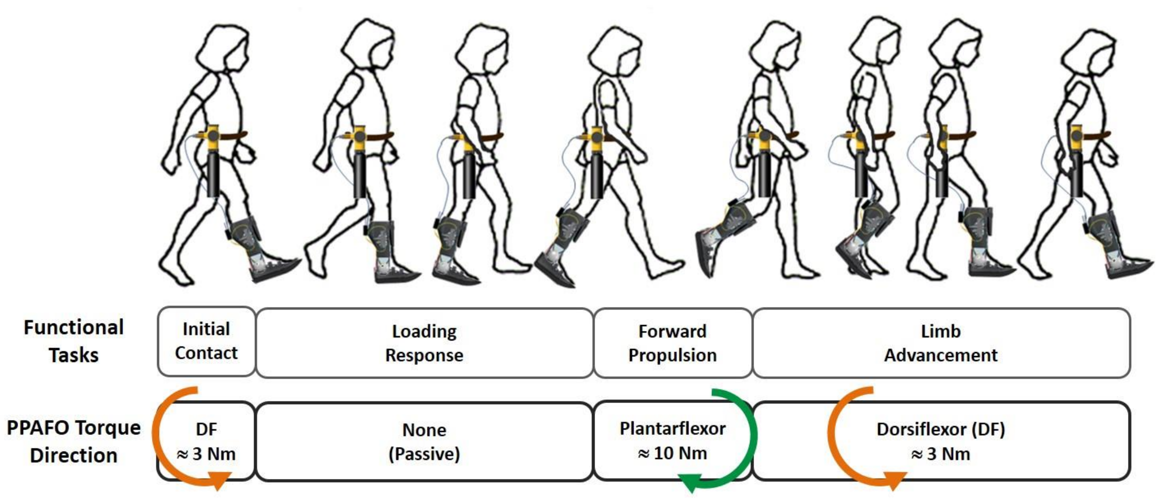

Figure 3.2: **Assistance during a gait cycle provided by the PPAFO.** Torque values are based on 100 psig during plantarflexor assistance and 30 psig during dorsiflexor (DF) assistance. Figure adapted from Kirtley (2006) [26]

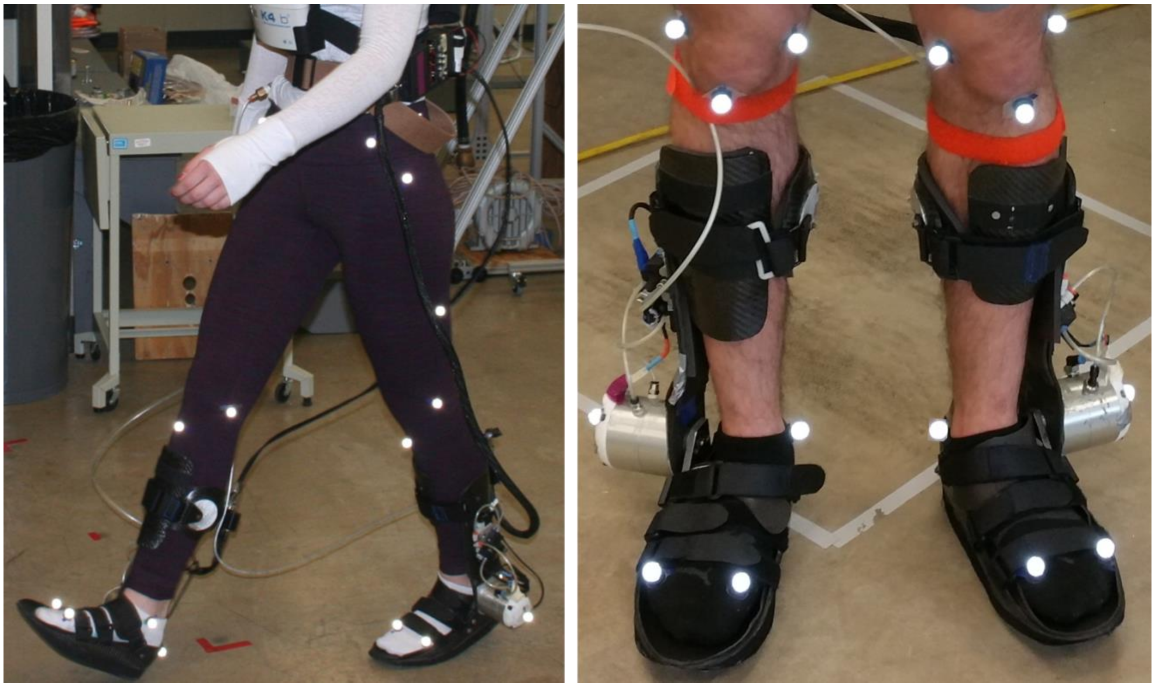

Figure 3.3: **Lower body motion capture marker sets used with the bilateral PPAFO testbed**.

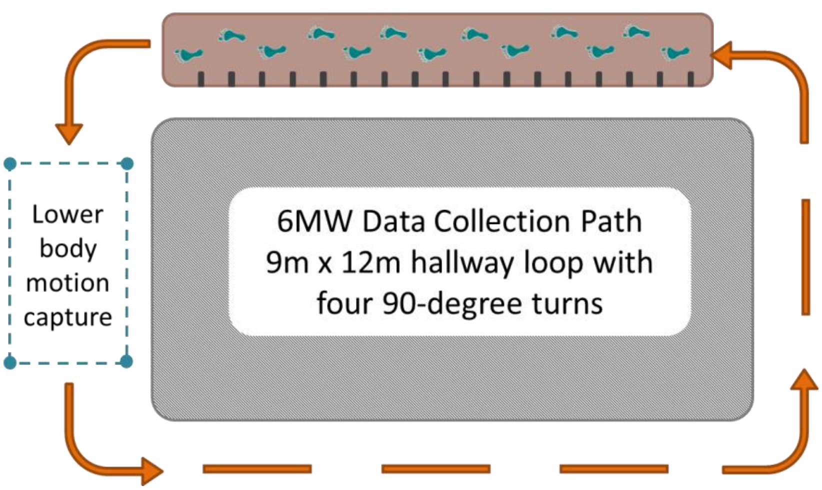

Figure 3.4: **6-minute walk test path with motion capture space, and gait mat in the loop.**

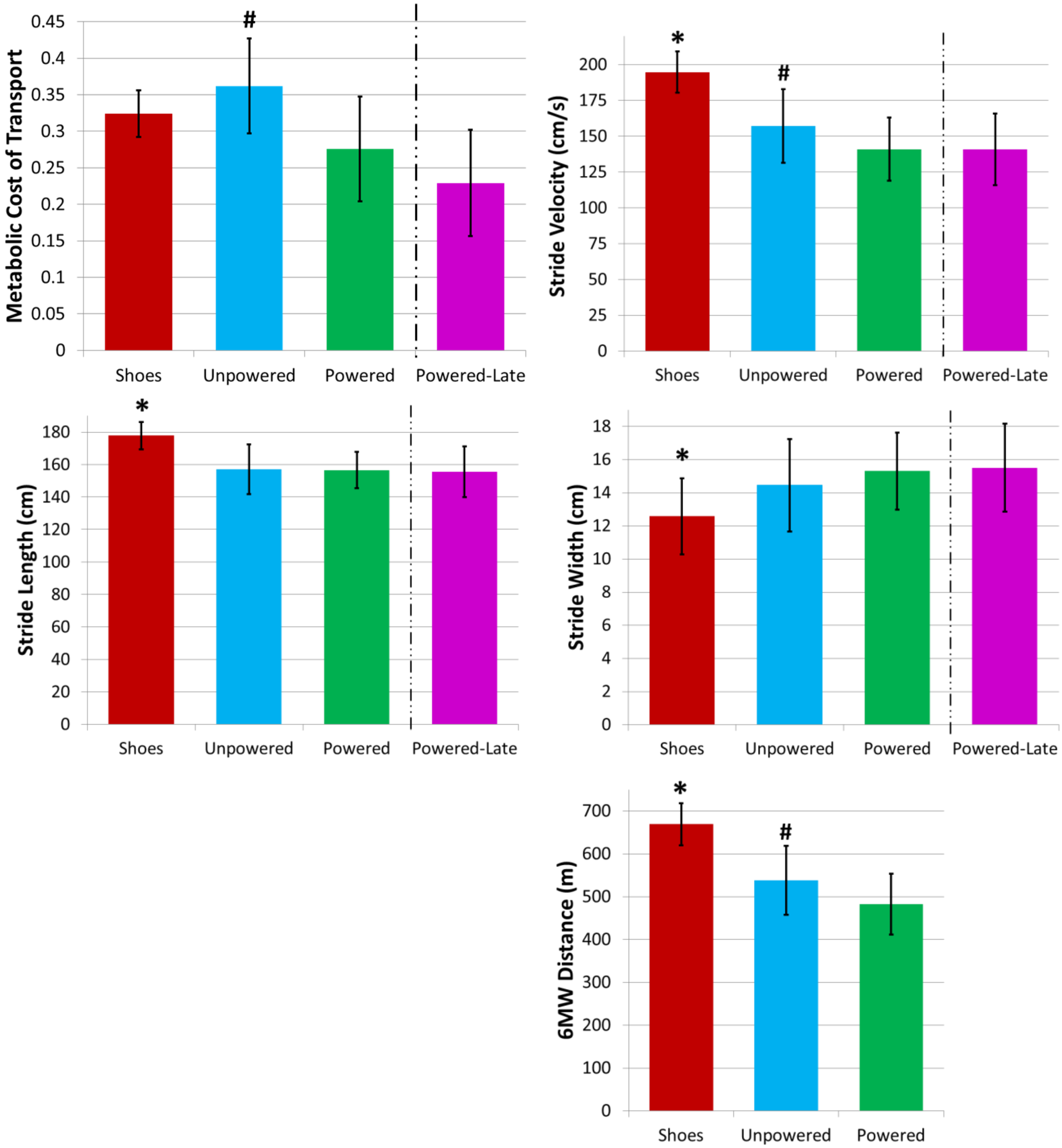

Figure 3.5: **Spatiotemporal and metabolic parameters of gait during the final 3 minutes of the 6MW.**

* indicates significantly different from other two footwear conditions

\# indicates significantly different from powered condition

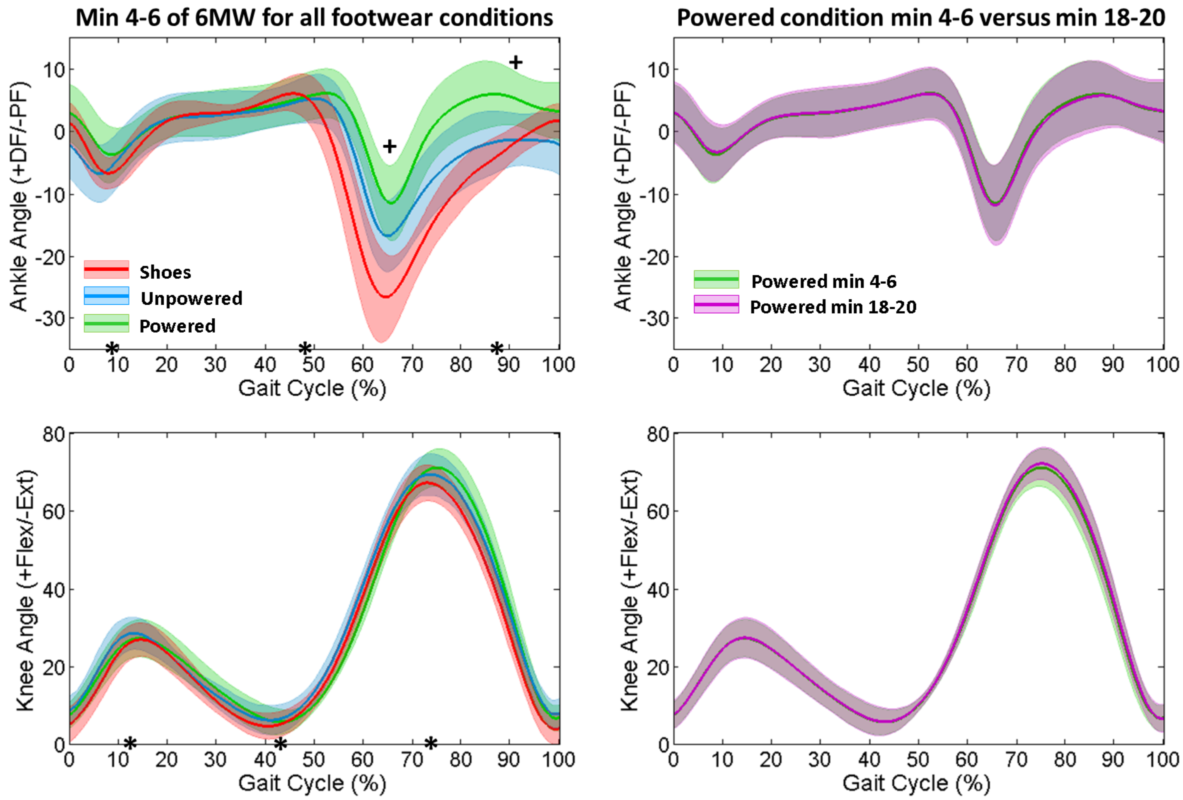

Figure 3.6: **Lower limb kinematics during the final 3 minutes of the 6MW test compared across footwear conditions at 4-6 and 18-20 minutes of walking.**

* indicates significant timing differences on the x-axis,
\+ indicates significant peak magnitude differences near the peak magnitude.
significance was set to $p < 0.05$

3.8. **TABLES**

Table 3.1: Participant demographics for 6MW and > 20 minutes of walking

Participant Groups		Males/Females	Age (years)	Height (cm)	Weight (kg)
Completed 6MW	n = 8	6 / 2	20.9 ± 1.2	178.7 ± 5.1	71.7 ± 5.7
Completed > 20 minutes of walking	n = 7	5 / 2	21.1 ± 1.1	178.4 ± 5.5	71.5 ± 6.1

Table 3.2: Mean (± standard deviation) Metabolic and Spatiotemporal parameter values for eight participants

	Shoes	**Unpowered**	**Powered**	**Powered-Late**	**p-value** (shoes, unpowered, powered)
Metabolic Cost of Transport	0.32 ± 0.03	0.36 ± 0.07 #	0.30 ± 0.08	0.25 ± 0.08	$p = 0.012$
Stride Velocity (cm)	194.9 ± 14.4 *	157.1 ± 25.7 #	141.1 ± 22.1	140.9 ± 24.9	$p < 0.001$
Stride Length (cm)	177.9 ± 8.4 *	157.1 ± 15.4	156.5 ± 11.2	155.6 ± 15.6	$p = 0.004$
Stride Width (cm)	12.6 ± 2.3 *	15.5 ± 2.8	15.3 ± 2.3	15.5 ± 2.7	$p = 0.005$
6MW Distance (m)	669.3 ± 49.6 *	538.9 ± 80.7 #	482.5 ± 70.8		$p < 0.001$

* indicates significantly different from other two footwear conditions

\# indicates significantly different from powered condition.

Table 3.3: Mean (± standard deviation) Ankle peak kinematic magnitudes and times

	Shoes	Unpowered	Powered (min 4-6)	Powered – Late (min 18-20)	p-value (shoes, unpowered, powered)
Ankle Angle Loading Response	-8.2 ± 3.1 deg	-6.3 ± 5.7 deg	-5.1 ± 4.7 deg	-4.7 ± 4.7 deg	p = 0.468
Ankle Angle Timing Loading Response	8.1 ± 1.3 %GC	7.2 ± 0.8 %GC #	9.3 ± 1.6 %GC	9.3 ± 2.0 %GC	p = 0.022
Ankle Angle Terminal Stance	6.7 ± 3.2 deg	5.5 ± 4.1 deg	6.1 ± 4.6 deg	7.1 ± 4.2 deg	p = 0.859
Ankle Angle Timing Terminal Stance	46.3 ± 2.5 %GC *	50.4 ± 1.1 %GC #	52.7 ± 0.7 %GC	51.2 ± 1.7 %GC	p < 0.001
Ankle Angle Pre-Swing	-27.8 ± 6.6 deg *	-18.6 ± 5.6 deg	-13.8 ± 6.0 deg	-12.9 ± 7.2 deg	p < 0.001
Ankle Angle Timing Pre-Swing	64.7 ± 1.5 %GC	64.8 ± 1.4 %GC	65.6 ± 0.6 %GC	66.2 ± 1.6 %GC	p = 0.129
Ankle Angle Terminal Swing	1.0 ± 3.3 deg	-0.2 ± 5.1 deg	7.0 ± 5.1 deg *	6.9 ± 5.6 deg	p = 0.008
Ankle Angle Timing Terminal Swing	98.5 ± 0.9 %GC *	93.0 ± 5.0 %GC	90.6 ± 5.3 %GC	89.6 ± 4.9 %GC	p = 0.002

* indicates significantly different from other two footwear conditions
# indicates significantly different from powered condition

Table 3.4: Mean (± standard deviation) Knee peak kinematic magnitudes and times

	Shoes	Unpowered	Powered (min 4-6)	Powered – Late (min 18-20)	p-value (shoes, unpowered, powered)
Knee Angle Loading Response	27.4 ± 3.9 deg	29.7 ± 4.2 deg	27.7 ± 4.2 deg	28.0 ± 6.3 deg	p = 0.757
Knee Angle Timing Loading Response	14.6 ± 1.3 %GC	13.2 ± 0.9 %GC *	14.6 ± 1.0 %GC	14.4 ± 1.0 %GC	p = 0.033
Knee Angle Terminal Stance	4.1 ± 3.5 deg	5.6 ± 3.3 deg	4.7 ± 2.9 deg	6.3 ± 3.3 deg	p = 0.612
Knee Angle Timing Terminal Stance	40.7 ± 0.9 %GC	41.2 ± 0.8 %GC	43.0 ± 1.2 %GC *	42.6 ± 0.9 %GC	p < 0.001
Knee Angle Pre-Swing	67.7 ± 4.2 deg	69.5 ± 5.5 deg	71.8 ± 4.9 deg	72.2 ± 4.5 deg	p = 0.237
Knee Angle Timing Pre-Swing	73.3 ± 1.0 %GC	73.3 ± 1.2 %GC	75.4 ± 0.9 %GC *	75.6 ± 0.9 %GC	p < 0.001
Knee Angle Terminal Swing	4.5 ± 5.1 deg	8.4 ± 3.2 deg	6.7 ± 6.6 deg	6.2 ± 3.5 deg	p = 0.446
Knee Angle Timing Terminal Swing	98.5 ± 1.4 %GC	98.4 ± 0.6 %GC	98.7 ± 1.0 %GC	99.1 ± 0.8 %GC	p = 0.611

* indicates significantly different from other two footwear conditions

Chapter 4 : Spatiotemporal and Metabolic Parameters of Gait in Persons with Multiple Sclerosis using Passive and Powered Ankle Foot Orthoses

4.1. ABSTRACT[1]

Objective: To determine if a powered orthosis or exoskeleton that provides dorsiflexor and plantarflexor assistance at the ankle can improve the gait of persons with multiple sclerosis (MS).

Design: Cross-sectional repeated-measures

Setting: University research laboratory

Participants: Sixteen participants with neurologist-confirmed diagnosis of MS and daily use of a prescribed custom passive ankle-foot orthosis (AFO). Post hoc analysis identified difference in impairment severity by groups based on Expanded Disability Status Scale (EDSS) score: moderate (EDSS ≤ 5.5, N = 8) and severe (EDSS ≥ 6.0, N = 8).

Interventions: Three 6-minute walk tests (6MW), one per footwear condition: shoes, prescribed passive AFO, and powered portable AFO (PPAFO), with the assistive devices worn on the more impaired limb.

[1] This chapter has been formatted and prepared to send to the Archives of Physical Medicine and Rehabilitation for future publishing.

Main Outcome Measures: Distance walked, spatiotemporal parameters of gait, and metabolic cost of transport were collected during each 6MW and compared between footwear conditions and EDSS groups.

Results: PPAFO use resulted in shorter stride length, slower stride velocity, and a shorter 6MW distance than a passive AFO. Regardless of footwear, the moderate group had longer and faster strides, longer 6MW distances, and reduced metabolic cost of transport compared to the severe group.

Conclusions: Use of the current embodiment of the portable powered AFO did not improve gait performance in a sampling of participants with gait impairment due to MS. Further research is required to determine if expanded training or modified design of powered orthoses can be effective at assisting and improving gait function in persons with gait impairment due to MS.

Keywords: orthosis, multiple sclerosis, gait, rehabilitation, exoskeleton

Abbreviations:
6MW (6-minute walk test)
AFO (ankle-foot orthosis)
EDSS (expanded disability status score)
MS (multiple sclerosis)
COT_m (metabolic cost of transport)
PPAFO (portable powered ankle-foot orthosis)
$\dot{V}O_2$ (rate of oxygen consumption)
$\dot{V}CO_2$(rate of carbon dioxide production)

4.2. **INTRODUCTION**

Traditional ankle-foot orthoses (AFOs) often fail to restore normal ankle function because they lack the ability to actively modulate motion control during gait and cannot produce propulsion torque and power. A test bed of a portable powered ankle-foot orthosis (PPAFO) has been developed to explore the issues and challenges related to creating mobile actively-powered orthotic devices [22]. As the PPAFO is still a research device, not ready for commercialization, it is being use as a platform to address some of these challenges such as issues of control, runtime, and weight. Researching these topics is helping to advance the field of powered orthotics towards commercial use. The PPAFO can provide modest dorsiflexor or plantarflexor torque at the ankle using a portable pneumatic power source (Figure 4.1). The bidirectional assistance is applied at the ankle as needed throughout all phases of the gait cycle including late stance and propulsion, which is unavailable in current commercially-available technologies [14]. The timing of when to apply the bidirectional assistance is determined by a user-specific tuned kinematics-based controller that uses the PPAFO's toe and heel force sensors and ankle angle to estimate the state of the limb during the gait cycle [23]. The PPAFO was built as a modular system with small, medium, and large foot beds and small, medium, and large tibial shells in order to choose the appropriate size for each participant. Padding was used to create a custom fit in the best-sized components for each participant. To further inform the design needs and functional impact of a powered ankle assistance device, a study was designed to assess the effect of added dorsiflexor and plantarflexor torque on a population of persons with gait impairment due to multiple sclerosis.

The major disease process of Multiple Sclerosis (MS) is characterized by demyelination lesions of the white matter of the brain stem, cerebellum, and spinal cord [70] resulting in lower limb weakness leading to gait impairment [71]. This gait impairment presents major personal, social, and economic burdens on those living with MS [73]. Passive AFOs are often used clinically to assist with foot drop due to lower limb weakness in persons with MS in attempts to mitigate the resulting gait impairment [74].

Passive ankle-foot orthoses have shown mixed results when analyzed in a research setting in persons with MS [75, 80, 83]. Ramdharry et al.[80] found that, although the AFO did initially cause destabilization of standing posture, after four weeks of using an AFO, persons with MS reported fewer limitations in their mobility (e.g. walking, running, stair climbing) when assessed using the Multiple Sclerosis Walking Scale-12 [81]. Sheffler et al. [75] saw no significant improvement in persons with MS with the AFO compared to no-device trials on functional gait tasks (timed 25-foot walk, five components of the modified Emory functional ambulation profile [82]). McLoughlin et al. [83] investigated the effect of a dorsiflexion assist orthosis during a modified 6-minute walk test. When using the dorsiflexion assist orthosis, there was no difference in distance walked or perceived fatigue, but there was reduced physiological cost of walking (change in heart rate when walking normalized by gait speed) [83].

Passive ankle-foot orthoses have also been studied in mixed populations of MS and stroke [76-78]. Bregman et al. [76]completed a more in depth analysis of the impact of a passive AFO on gait in persons with MS and stroke. They evaluated the effect of the mechanical properties of the AFO (stiffness and neutral angle as measured by a BRUCE device [79]) on the energy cost of walking (calculated from the net oxygen consumption, and respiratory exchange ratio, then normalized by walking speed and body mass [90]), walking speed, gait kinematics, and kinetics [76]. The researchers concluded that if the mechanical properties of the AFO matched the patient's needs, the patient greatly benefited with improved outcome measures. Overall they found that walking with an AFO decreased cost of walking and greater walking speed; although when the participants were divided into two groups (presence or absence of foot drop during swing), statistical power was diminished. Expected differences in ankle angle (decreased plantarflexion during swing and initial contact in the foot drop group, and no change in the no foot drop group) were observed due to the AFO, but without statistical significance [76]. Bregman et al. continued to study the impact of AFOs on persons with MS and stroke, specifically by using a spring-like carbon-composite AFO [77]. With the AFO in that study, Bregman et al. observed a decrease in energetic cost of walking, decreased range of motion of the ankle, and decreased net work at the ankle. The decreased range of motion was due to reduced plantarflexion, which was mechanically blocked by the AFO, leading to reduced push-off. Even with decreased push-off, the AFOs were able to provide enough assistance in dorsiflexion to lead to an overall reduced metabolic cost. Hwang et al. compared an AFO-shaped band (elastic band attached to the thigh and shank with a loop to hold up the forefoot), a molded plastic AFO, and barefoot conditions in persons with stroke and MS [78]. The AFO-shaped band increased the gait velocity and cadence compared to the barefoot and AFO conditions, with mixed results in stride length differences. Although most of these studies agree in that passive AFOs can assist persons with MS to improve gait from an unassisted condition, the existing research is limited to devices that only provide assistance with dorsiflexion.

The goal of this current investigation was to determine if a powered device that provided active bi-directional (dorsiflexor and plantarflexor) assistance can improve the gait of persons with gait impairment due to MS. This study evaluated both metabolic and spatiotemporal aspects of gait with the PPAFO compared to both a patient-prescribed AFO and shoes-only condition during a six minute walk protocol.

4.3. METHODS

4.3.1. Participants

The study included 16 participants with a neurologist-confirmed diagnosis of MS and daily use of a prescribed custom passive AFO for gait assistance (12 female/4 male, age: 54.6 ± 5.3 years, Expanded Disability Status Score: 5.75 [4, 6] (mean [IQR]). Participants were allowed to use their normal assistive device: 4 used a single point cane, 1 used a two-wheeled walker, 3 used a four-wheeled walker, and 2 used the walls and arms of caregivers on a regular basis. Approval for the study was granted by the University of Illinois at Urbana-Champaign Institutional Review Board and participants gave informed consent.

4.3.2. Experimental Procedure

First, a clinical examination of disability was performed by trained investigators (REK, YL)[2] to establish Expanded Disability Status Score (EDSS) [107]. Participants completed three 6-minute walk (6MW) tests that were performed over-ground in a 9 m x 12 m hallway loop with four 90-degree turns. One 6MW was completed per footwear condition: shoes, AFO, and PPAFO. The PPAFO was worn on the more affected limb (i.e., the same side as the prescribed AFO, or more impaired side if prescribed bilateral AFOs) with the participant's normal walking shoe on the contralateral limb. For this study, the electronics control box of the PPAFO was worn at the chest on an over-the-shoulder harness, and the portable pneumatic air tank was carried by a research assistant following the participant. Participants first wore the PPAFO for a training period (~ 10 minutes of intermittent walking) where their gait pattern was programmed into the PPAFO controller. During the training period, sensor data from the heel, toe, and angle sensors on the PPAFO were used to estimate the phases of gait for each participant. The timings of these phases were then used to better inform the controller of appropriate timings for applying gait assistance. The shoes condition 6MW was always completed first as a baseline. The remaining two conditions (AFO and PPAFO) were randomized between subjects with 10-minute periods of recovery between all walks.

4.3.3. Data analysis

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[2] Rachel E Klaren (REK) and Yvonne Learmonth (YL) from the Exercise Neuroscience Research Lab in the Department of Kinesiology and Community Health at the University of Illinois at Urbana-Champaign performed the EDSS clinical examination.

Multiple parameters of interest were collected during each 6MW. Spatiotemporal parameters of gait were measured as the participant passed over an 8.9 m pressure sensitive walkway (GAITRite, CIR Systems Inc., Sparta, NJ) with each lap (≥ 4 passes). Bilateral values of stride length (cm), stride width (cm), and stride velocity (cm/s) for each participant were determined from the software (GAITRite 4.0). Averaged values were then computed from the data recorded during the six-minute period. The total 6MW distance traveled (m) was recorded with a measuring wheel (RT312, Rolatape, Watseka, IL). Rates of oxygen consumption ($\dot{V}O_2(\mathrm{mL/min})$) and carbon dioxide production ($\dot{V}CO_2(\mathrm{mL/min})$) were measured breath-by-breath using a commercially available portable metabolic unit (K4b2, Cosmed S.r.l., Rome, Italy). $\dot{V}O_2$ and $\dot{V}CO_2$ were measured as 30-second averages for 1 minute before the 6MW (resting) and over the entire 6MW (walking). Steady-state values were computed from average walking values of the final 3 minutes of the 6MW. Net values were computed by subtracting average resting values from the steady-state values (Eqns. 1 and 2).

$$\left[\dot{V}O_2\right]_{\mathrm{net}}(\mathrm{mL/min}) = \left[\dot{V}O_2\right]_{\mathrm{steady-state}} - \left[\dot{V}O_2\right]_{\mathrm{resting}} \tag{1}$$

$$\left[\dot{V}CO_2\right]_{\mathrm{net}}(\mathrm{mL/min}) = \left[\dot{V}CO_2\right]_{\mathrm{steady-state}} - \left[\dot{V}CO_2\right]_{\mathrm{resting}} \tag{2}$$

Metabolic cost of transport (COT_m) was used to quantify the energy expenditure during gait [86]. COT_m was computed from the net $\dot{V}O_2$ and $\dot{V}CO_2$ using a modification to the Brockway equation [84] to calculate the COT_m as suggested by Donelan et al. [86], with updated coefficients by Adamczyk et al. [87] (Eqn. 3).

$$\text{Metabolic Cost of Transport } (\text{COT}_\text{m}) =$$

$$\frac{16.477\ (\text{W} \cdot \text{s/mL}) * \left[\dot{\text{V}}\text{O}_2\right]_{\text{net}}(\text{mL/min})\ + 4.484\ (\text{W} \cdot \text{s/mL}) * \left[\dot{\text{V}}\text{CO}_2\right]_{\text{net}}(\text{mL/min})}{60\ (\text{s/min}) * \text{body weight (N)} * \text{gait speed (m/s)}} \tag{3}$$

For the PPAFO condition, the additional mass of the PPAFO and controller box (2 kg) was added to the participant's mass to be included in the body weight.

4.3.4. Statistical analysis

Gait and metabolic data were analyzed for the effect of footwear and disability status. Post-hoc analysis identified a difference in participants based on severity of impairment. Participants were divided into groups based on EDSS score: moderate (EDSS ≤ 5.5, N = 8) and severe (EDSS ≥ 6.0, N = 8). Multiple spatiotemporal parameters of gait (stride length, stride width, and stride velocity) were analyzed with a two-way repeated-measures multivariate analysis of variance (MANOVA) test to compare parameters across footwear conditions (shoes, AFO, and PPAFO) and between EDSS groups for all 16 participants. Follow-up univariate ANOVAs were examined on significant variables, followed by LSD post-hoc comparisons where appropriate. Due to equipment error, one participant completed the 6MWs but the final distance was not available; a two-way repeated-measures univariate ANOVA was used to compare the 6MW distance across footwear conditions (shoes, AFO, and PPAFO) and between EDSS groups for the remaining 15 participants (11 female/4 male, age: 54.9 ± 5.4 years, EDSS: 5.2 ± 1.1). Another participant completed the 6MWs but there was an equipment error with the portable metabolic unit; a two-way repeated-measures univariate ANOVA was used to compare the COT_m for 15 participants (11 female/4 male, age: 54.3 ± 5.3 years, EDSS: 5.2 ± 1.1). All statistical tests were performed using SPSS (v22, IBM Inc., Armonk, NY) and the level of significance was set at $\alpha = 0.05$.

4.4. **RESULTS**

The MANOVA test on spatiotemporal variables of gait found significant main effects ($p < 0.001$ (due to group), $p = 0.010$ (due to footwear)), but no interaction effect (Figure 4.2). Follow-up univariate ANOVAs with respect to stride length and stride velocity found significant main effects for footwear ($p = 0.016$ and $p = 0.026$, respectively) and for group ($p = 0.002$ and $p < 0.001$, respectively). No significant differences were found for stride width. The univariate ANOVA for COT_m found significant main effects ($p < 0.001$) for group, but no significant difference for footwear condition (Figure 4.2). The univariate ANOVA for 6MW distance found significant main effects for footwear ($p < 0.001$), group ($p < 0.001$), and the footwear by group interaction ($p = 0.004$).

The follow-up univariate ANOVAs, with LSD comparisons between footwear conditions, indicated that PPAFO use resulted in shorter stride length, slower stride velocity, and a shorter 6MW distance than a custom AFO, especially in the moderate group compared to the severe group. Regardless of footwear, the moderate group had longer and faster strides, and longer 6MW distances than the severe group. Metabolic expenditure was greater for the severe group than the moderate group, while footwear did not result in a significant change in COT_m.

4.5. **DISCUSSION**

This study was the first to assess the use of a powered ankle-foot orthosis to provide bidirectional powered gait assistance to persons with MS. Participants wore the powered orthosis (PPAFO), their own passive ankle-foot orthosis (AFO) inside of their shoe, or only their shoes (Shoes) in a 6-minute walk test. Participants were grouped based on EDSS score: EDSS ≤ 5.5 (moderate) and EDSS ≥ 6.0 (severe). Spatiotemporal parameters of gait and metabolic measurements were collected and then compared across footwear conditions (PPAFO, AFO, Shoes) and between disability groups (moderate, severe). Significant differences were found between disability groups and across footwear conditions for the spatiotemporal parameters of gait and metabolic cost of transport.

The significant differences between disability groups in the gait parameters examined here have been previously noted. The group differences of longer, faster strides and longer 6MW distances in the moderate group have been previously observed [108-110]. Although the group differences had been previously observed, it is unique that the group differences in stride velocity and stride length held across different footwear conditions, including a new device to the participants, our PPAFO. From the group differences, even with the PPAFO, the moderate disability group kept a greater stride length and stride velocity than the severe impairment group. It is worth noting though that the group difference seen in 6MW distance was accompanied by a significant interaction between footwear and group. This interaction suggests that even though the moderate group was able to maintain their stride length and stride velocity performance above the severe group with the PPAFO, these faster and longer strides did not necessarily translate to a longer 6MW distance with the PPAFO. The use of the PPAFO in the moderate group significantly limited their 6MW distance compared to the severe group. This difference between groups for the 6MW distance with the PPAFO possibly suggests that while the moderate impairment group was better able to become accustomed to using the PPAFO in their gait, it was possibly at the expense of the total distance covered in the 6MW. Since the severe impairment group did not reduce their 6MW distance as much while using the PPAFO, it is possible that their disability had already limited their mobility so much that a new device could not further reduce their performance.

We also found a difference in metabolic cost of transport between the moderate and severe disability groups, with the severe disability group having a significantly higher COT_m than the moderate disability group. The factors contributing to this increased COT_m for the severely disabled group have been detailed further by using a metabolic parameter of oxygen cost of walking in research by Sandroff et al. [111]. Generally, it is expected that a person with a more advanced impairment would need to work harder to overcome that impairment while ambulating.

The resulting spatiotemporal gait parameter comparisons regarding the use of an AFO compared to shoes seen in this study are also in line with previous research. In agreement with studies by Bregman et al. [76, 77], the AFO footwear condition resulted in faster walking speeds (greater stride velocity and greater 6MW distance) than a shoes-only condition; however, our results did not agree since we did not find a significant reduction in metabolic parameters of walking due to wearing an AFO. The lack of this difference could be due to any number of factors, such as a difference in the participants use of their AFOs or different calculations of metabolic parameters. More studies on AFO use in persons with gait impairment should quantify the amount of daily use that a participant uses their AFO, how many years they have used an AFO, and possibly what type or types of AFOs the person has used. The only inclusion criteria for our study was that a participant had an AFO. Having an AFO, does not necessarily mean that each of our participants used their AFOs in similar ways, or for similar amounts of time. It is expected that the amount of metabolic benefit that one would receive from using an AFO, is related to the amount that the AFO is actually used. Different metabolic calculations to evaluate the energy used during gait are used across various fields of research, making it difficult to make direct comparisons between studies [84, 87, 89, 90]. A study by McLoughlin et al. [83] found a similar decrease as Bregman et al, in the physiological cost of walking when using a dorsiflexion assist orthosis that we did not find in our study. There was not a significant decrease in metabolic measurement in our study (COT_m) with the AFO compared to the shoes condition (Fig. 2).

4.5.1. Study Limitations

The shorter, slower strides seen when wearing the PPAFO compared to the AFO condition could be due to a number of factors. In this study, the participants were given a minimum of 20 minutes to adapt to the device as the controller was programmed to their gait, but even more training and practice might have been ideal to elicit the improvement in spatiotemporal gait parameters that we expected. Some studies have that suggested upwards of 20 minutes or even multiple days of use may be necessary to adapt to a powered ankle orthosis [37, 38, 41, 43]. The practice time with the PPAFO was not limited, although care was taken not to severely fatigue the participant, giving them sufficient rest (at least 10 minutes) between each 6MW, as well as extra rest as needed and requested.

Some of the participant's gait might have also been impacted by the fit and comfort of the PPAFO devices. Even though the devices were sized and fit to the participant, they were not custom fabricated for each participant. Custom therapeutic AFOs range in weight based on the type and the size for example for a medium-sized female: a carbon fiber strut would weigh 0.1 kg, a solid thermoplastic AFO would weigh around 0.4 kg, and a thermoplastic hybrid with metal ankle joints and uprights would be upwards of 0.7 kg or more [112]. The PPAFO added weight to the distal ankle (1.8 kg total PPAFO weight), possibly impacting the participant's gait [104, 105]. The PPAFO was also designed with a heel rocker and forefoot rocker in the sole that were not accounted in the gait analysis completed here [101-103].

4.6. **CONCLUSIONS**

These results indicate that within this study design, the participants did not overcome their gait impairment while using the PPAFO. Yet, the PPAFO did not negatively impact the COT_m used to walk. The hypothesized impact of a powered ankle-foot orthosis with bidirectional ankle torque providing improved gait assistance was not realized in this study and could be due to any number of factors, such as a need for more training and experience walking with the PPAFO, fatigue, or a need for improved device design. The PPAFO has many areas for further development such as more advanced controls and sensors to better track impaired gait. These updated controls could impact the amount of training needed for comfortable use of the PPAFO. A hardware redesign for the PPAFO would be beneficial to even further reduce the weight of the device while increasing the amount of assistive torque that it is able to produce. Lastly, from the experiences with using the PPAFO in this study, an analysis of the structural components (tibial and footbed carbon fiber shells) is warranted to better understand how these components interact with the wearer in regards to comfort, fit, and walking action.

4.7. ACKNOWLEDGEMENTS

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4.8. **FIGURES/TABLES**

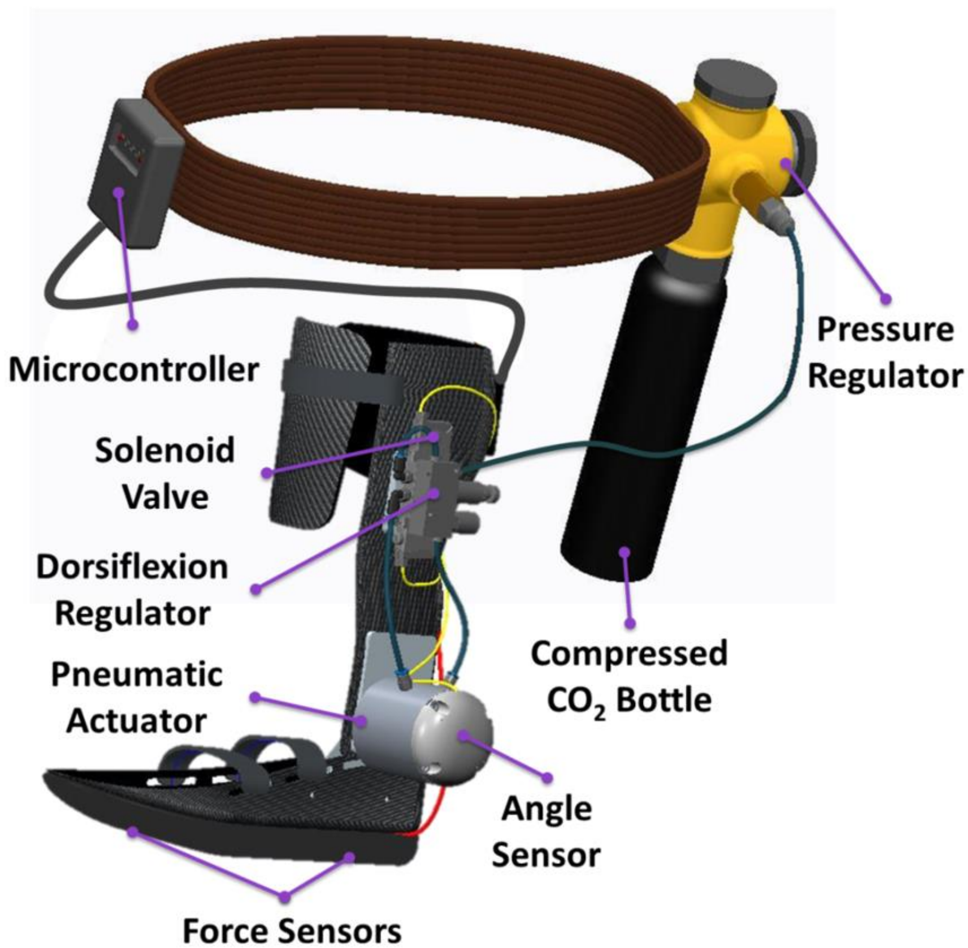

Figure 4.1: **The portable, powered ankle-foot orthosis (PPAFO) test bed.** The PPAFO can give powered assistance in both dorsiflexion and plantarflexion directions.

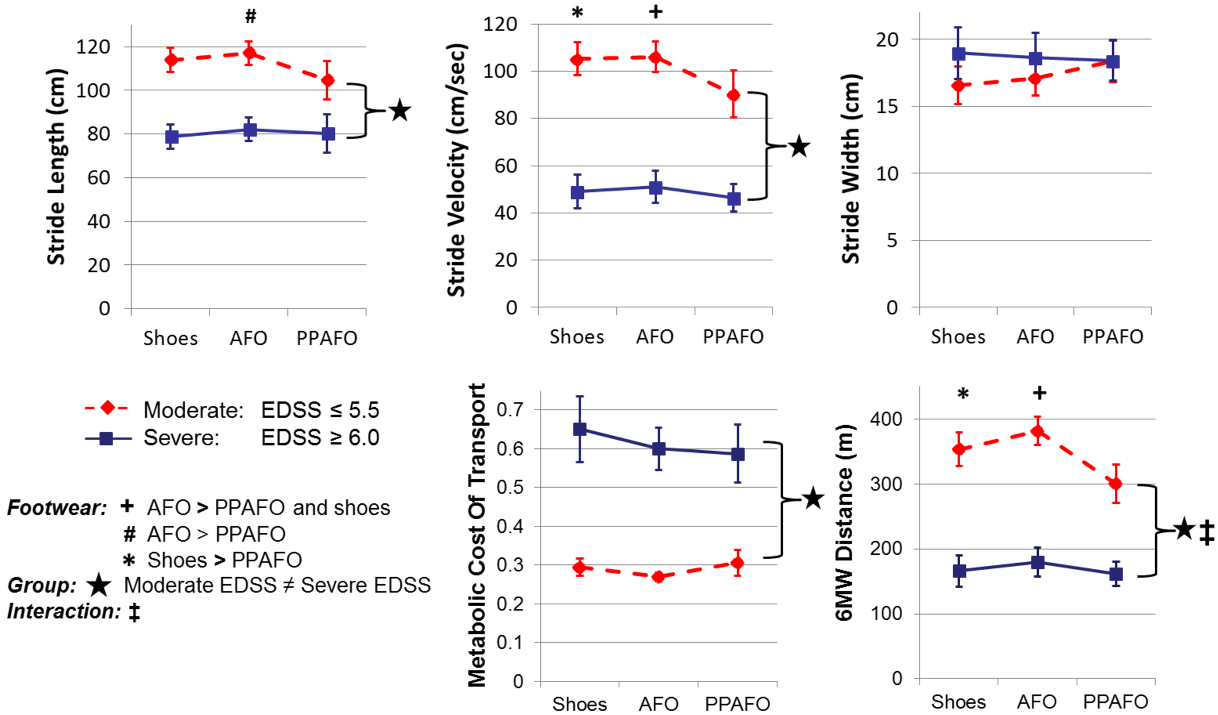

Figure 4.2: **Spatiotemporal gait parameters for 16 persons with MS in three footwear conditions: *Shoes*, *AFO*, or *PPAFO* (mean ± std. err.).** Metabolic cost of transport for 15 persons with MS, and 6MW distance for 15 persons with MS. Symbols represent significant main effect or interaction differences ($p < 0.05$).

Chapter 5 : Kinematic evaluation of a bidirectional powered ankle-foot orthosis in persons with Multiple Sclerosis.

5.1. ABSTRACT

Introduction: Passive ankle-foot orthosis (AFOs) are commonly used by persons with multiple sclerosis (MS) to control foot drop and aid in gait impairment. The aim of this study was to evaluate the use of a portable powered AFO (PPAFO) that provides dorsiflexor and plantarflexor assistance at the ankle in persons with MS by quantifying the change in kinematics of gait between powered and passive AFO, as well as a shoes-only condition in a six-minute walk test (6MW).

Methods: Sixteen persons with MS with daily use of an AFO were recruited for the study. Each participant completed three 6MWs, during which ankle and knee kinematics were recorded over a 3 m x 1.5 m motion capture space. Each 6MW was completed in a different footwear condition, with a shoes-only condition complete first, followed by the participant's prescribe AFO, and then powered AFO condition. Outcome measures of sagittal-plane knee and ankle angles were determined.

Results: Significant differences between peak values for sagittal ankle angle were found such that the PPAFO was able to provide better assistance for foot drop during swing than the AFO or a shoes condition. Significant differences between peak values and timing of peak values for sagittal plane knee angles were found such that the changes are likely due to compensatory reactions to the changes at the ankle induced by the footwear.

Discussion: Continued research needs to be done to optimize the PPAFO and like-devices for assisting those with impaired gait, but this study is a first step towards showing that greater range of motion can be achieved with a powered device.

5.2. INTRODUCTION

Gait impairment is one of the major day to day issues present in persons with Multiple Sclerosis (MS) [68]. The major disease process of MS is characterized by demyelination lesions of the white matter of the brain stem, cerebellum, and spinal cord [70] resulting in lower limb weakness leading to this generalized gait impairment [71]. Much research has been done and is ongoing to better understand and quantify the different aspects of gait impairment in persons with MS. By studying gait, one can possibly better identify and diagnose persons with early onset MS [72], as well as offering interventions towards improving the gait impairment as it has been noted that any gait impairment can place a significant personal and financial burden, as well as a decrease in quality of life for persons with MS [73].

One intervention that has frequently been employed to assist with gait impairment is an ankle-foot orthosis [74]. Mixed results have been reported when evaluating the clinical and biomechanical advantages of physician-prescribed, custom AFOs for individuals with MS [72, 75-77]. Bregman et al. reported that polypropylene custom passive AFOs with low stiffness were able to overcome foot-drop during swing, while not inhibiting the stance phase of gait in persons with central neurological disorders (MS and stroke) [72, 76]. Sheffler et al. reported that 3 months of using a custom passive AFO for dorsiflexion and eversion weakness resulted in no significant improvement in times on clinical gait tests [75]. In another study, Bregman et al. studied the use of a custom carbon composite spring-like passive AFO to aid in the push off ability of the ankle in persons with MS. The custom carbon composite AFO presumably reduced the energy cost of walking by reducing the ankle range of motion and ankle angular velocity.

The reduction of ankle work, contributed to a need for compensatory work at the hip, such that the overall cost of walking was only slightly reduced, but at the expense of altered biomechanics that could be less favorable, causing premature fatigue or even injury [77]. The mixed results within the current field of research involving passive AFOs as assistive devices for foot drop and plantarflexor weakness motivate further device design that can more reliably assist with impairments of gait, specifically seen in persons with MS. It is possible that these traditional passive AFOs often fail to restore normal ankle function because they lack the ability to actively modulate motion control during gait and cannot produce propulsion torque and power.

One reason that the aforementioned passive AFOs fail to restore normal gait is that they lack the functionality of moving with natural gait motion. Passive AFOs are unable to provide variable and bidirectional motion control throughout the phases of gait, especially during late stance where propulsive power is needed at the ankle. New research devices are being developed for powered assistance at the ankle. To better inform the design of such powered devices, we propose to use our portable, powered ankle-foot orthosis (PPAFO) test bed, which can provide bidirectional assistive torque at the ankle during gait [22].

In this study, we used the PPAFO test bed to further understand the functional needs and constraints of an active ankle assist device in a population of persons with gait impairment and Multiple Sclerosis (MS). The study design was based around a six-minute walk test (6MW), commonly used as a clinical and functional measure of gait in persons with MS [109, 113-116].

The aim of this study was to evaluate the use of powered AFOs to passive AFOs in persons with MS by observing ankle and knee joint kinematics during a 6MW test.

5.3. METHODS

5.3.1. Portable Powered Ankle Foot Orthosis

The PPAFO can provide modest dorsiflexor or plantarflexor torque at the ankle via an untethered pneumatic power source (Figure 5.1) [22]. Torque is delivered to the ankle through a pneumatic circuit. A portable CO_2 tank (20oz, Catalina Cylinders, Hampton, VA) with a bottle top regulator (JacPac J-6901-91, Supplierpipeline, Inc.; Waterloo, Ontario, Canada) is the pneumatic fuel source which is connected to two directional solenoid valves (VUVG 5V; Festo Corp-US, Hauppauge, NY) to direct the pneumatic power into a either side of a rotary actuator (PRN30D-90-45, Parker Hannifin, Cleveland, OH). One valve directs the compressed gas to the side of the rotary actuator for plantarflexion assistance, where the other valve directs the compressed gas to the side of the rotary actuator for dorsiflexion assistance. The valves are controlled by a waist worn microcontroller and powered by a rechargeable battery pack.

The applied torque of the PPAFO can provide directional assistance at the ankle as needed throughout all phases of the gait cycle (Figure 5.3). The PPAFO microcontroller is programmed with a state estimation controller (SE) [24], where the current state of the gait cycle is estimated by sensor input from heel and toe sensors on the sole of the PPAFO, as well as the ankle angle from the rotary actuator position. The SE controller provides actuation during four functional gait tasks: (1) initial contact, (2) loading response, (3) forward propulsion, and (4) limb advancement.

5.3.2. Participants and protocol

The study included 16 participants with a neurologist-confirmed diagnosis of MS and daily use of a prescribed custom passive AFO for gait assistance (12 female/4 male, age: 54.6 ± 5.3 years, Expanded Disability Status Score: 5.1 ± 1.1).

First, a clinical examination of disability was performed to establish EDSS score [107]. Participants first wore the PPAFO for a training period where their gait pattern was programmed into the PPAFO controller. The training period took approximately 20 minutes; during that time, the participant walked sporadically with the PPAFO for a cumulative walking time of ~10 minutes. To start the training period, the PPAFO controller was set to baseline assistance timing based on normative able-bodied gait timing values [25]. The participant was asked to walk with the powered PPAFO for more than 10 continuous test steps during which heel, toe, and angle sensor data from the PPAFO were collected. The PPAFO sensor data were then evaluated to adjust for the participant's gait pattern. The PPAFO controller was then updated with the new actuation timings, and the participant was asked to repeat the test steps such that the process would be repeated. The controller adjustment process was repeated at least three times for each participant. Verbal feedback from each participant was also used to confirm comfortable actuation timings. After the controller timings were set, each participant finished the training period by walking until he or she felt comfortable with the actuation of the

PPAFO. At a minimum each participant completed one loop around the 9 m x 12 m hallway while acclimating to walking with the PPAFO.

After completion of the training period and a minimum of 10 minutes of rest, participants completed the three 6-minute walk tests using standardized instructions and delivery [113]. The 6-minute walk (6MW) tests were performed over-ground in a 9 m x 12 m hallway with four 90-degree turns (Figure 5.2). One 6MW was completed per footwear condition: shoes, AFO, and PPAFO. The PPAFO was worn on the affected limb (i.e., the same side as the prescribed AFO, or more impaired side if prescribed bilateral AFOs) with a normal walking shoe on the contralateral limb. The shoes condition 6MW was always completed first as a baseline. The remaining two conditions (custom AFO and PPAFO) were counter-balanced between subjects. Participants were given a minimum of 10 minutes to rest and recover between each 6MW. Participants' lower body movements were recorded using a motion capture system for 3 m of each loop around the 6MW path (Oqus Motion Capture Camera System, Qualisys, Highland Park, IL). A lower body 23 static marker set was used with markers on the ASIS, greater trochanter, mid-thigh, lateral knee epicondyle, medial knee epicondyle, tibia, lateral malleolus, medial malleolus, heel, toe 1, and toe 5 for both the right and left limbs, as well as a sacral marker. In the trials where the PPAFO was worn, an extra marker was placed on the lateral point of the actuator in line with the plantarflexion/dorsiflexion axis of the ankle (Figure 5.4).

5.3.3. Data Processing

The sagittal plane joint angles for the impaired limb of each participant during walking were determined from the motion capture data (QTM v2.7, Qualisys, Highland Park, IL; Visual3D v5, C-Motion, Germantown, MD). The marker data were filtered (4^{th} order Butterworth filter, 12Hz cutoff frequency), and then using inverse kinematics subject-specific models calibrated to a static standing pose, the joint angles were computed. Heel strike and toe off events were determined for each gait cycle by visual inspection. Each participant completed at least one gait cycle per pass through the motion capture space, and each participant completed at least 3 passes per 6MW. To compare joint angles between footwear conditions, for each gait cycle, the local minimum or maximum peak value and peak timing were found during four phases of gait: loading response, terminal stance, pre-swing (near toe-off), and terminal swing.

5.3.4. Statistical Analysis

Outcome measures of four peak magnitudes and four peak timings for each joint (knee, ankle) were compared. One participant was excluded from analysis due to poor marker visibility and unsatisfactory camera calibration, leaving 15 participants for statistical analysis (11 female/4 male, age: 54.6 ± 5.5 years, Expanded Disability Status Score: 5.1 ± 1.1). Further, due to marker occlusion problems with some participants, only ankle and knee data were consistently available. Since four peaks magnitudes and timings per joint angle were examined, a two-way repeated-measures MANOVA test was run for each joint (ankle, knee) to compared across footwear conditions (shoes, AFO, and PPAFO) and between EDSS groups (moderate: ≤ 5.5, N = 8; severe: ≥ 6.0, N = 7)(α = 0.05; SPSS Inc., Chicago, IL). . Follow-up univariate ANOVAs were examined on significant variables, followed by LSD post-hoc comparisons where appropriate.

5.4. **RESULTS**

The ankle MANOVA indicated significant differences for the main effect of footwear ($p = 0.006$), but no significant EDSS group differences or interaction effect (Figure 5.5, Table 5.1). The knee MANOVA indicated significant differences for the main effects of footwear and EDSS group, as well as a significant interaction effect ($p < 0.001$, Figure 5.5, Figure 5.6, Table 5.2). Follow-up analyses found significant differences between peak values for ankle angle during swing. Significant differences between peak values and timing of peak values for sagittal plane knee angles were found during stance and swing.

Subsequent univariate ANOVA analyses found that there were differences in ankle kinematics due to footwear during loading response, pre-swing, and terminal swing (Figure 5.5, Table 5.1). Specifically, the peak plantarflexion angle (minimum ankle angle) during the loading response was significantly greater in the PPAFO condition than the shoes condition; the timing of this maximum plantarflexion during the loading response was significantly later in the PPAFO condition than either the shoes or AFO conditions. The peak plantarflexion during pre-swing (toe-off) was significantly less in the shoes condition compared to both the AFO and PPAFO conditions. The peak dorsiflexion during terminal swing was significantly greater in the PPAFO condition than both the AFO and shoes condition, and the AFO condition showed a significantly greater peak angle than the shoes condition. There were no significant differences between EDSS groups for ankle angle (Figure 5.6).

Subsequent univariate ANOVA analyses for the knee parameters found differences during the loading response, terminal stance, and terminal swing subphases of gait due to footwear and differences during the loading response, terminal stance, and pre-swing subphases of gait due to EDSS group (Table 5.2). Specifically, the peak knee flexion during the loading response was significantly greater in the PPAFO than the shoes condition, and the timing of this peak was significantly earlier in the shoes condition than the AFO condition. The peak knee extension angle during terminal stance was significantly greater and later in the PPAFO condition than both the shoes and AFO trials. The peak knee extension angle during terminal swing was significantly greater in the PPAFO condition than the shoes and AFO conditions although the timing did not vary (Figure 5.5). For knee angle, there were significant differences between the moderate and high impairment EDSS groups ($p<0.001$), regardless of footwear for the peak values at the loading response, terminal stance, and pre-swing phases of gait (Figure 5.6). A footwear by EDSS group interaction was also significant in the MANOVA for the knee kinematic parameters ($p = 0.007$), although this interaction was not specifically seen to be significant in any individual peak or timing parameter.

5.5. **DISCUSSION**

One of the primary reasons of prescribing an AFO is to compensate for the foot drop and foot slap that occur as a result of weakness of lower limb muscles. Foot drop occurs during the swing phase of gait when the foot is in the air, and foot slap occurs during the loading response after the heel contacts the ground. During both of these phases of gait, the PPAFO, which provides active powered dorsiflexion and plantarflexion assistance, was found to significantly increased dorsiflexion angle compared to the shoes condition (Figure 5.5). During swing, the PPAFO also increased dorsiflexion angle compared to the AFO condition (Figure 5.5). These results are similar to those seen by van der Liden et al. in a study of persons with MS using a functional electrical stimulation (FES) device; where by using the FES device, participants experienced more dorsiflexion both at initial contact and during swing [117]. Bregman et al. also found that while wearing an AFO, the overall ankle range of motion was decreased by 12%, especially by reducing plantarflexion at pre-swing and reducing dorsiflexion during terminal stance [77]. Our study found a similar significant reduction in plantarflexion angle during pre-swing, although we did not observe the reduction in dorsiflexion in terminal stance. Although not directly calculated, an overall reduction in range of motion was seen in our study both with the AFO and PPAFO conditions (Figure 5.5).

Few other studies have evaluated changes in kinematics due to AFOs in persons with MS, although some have looked at the kinematics in general in persons with MS and compared them to gait kinematics of healthy controls. Huisinga et al. found decreased peak plantarflexion angle during loading response and at toe-off in their MS population compared to healthy controls [118]. Kelleher et al. also found differences in ankle kinematics between two MS groups based on impairment level and a healthy control group, observing that both MS groups had a decreased overall range of motion at the ankle, mostly due to reduced plantarflexion at toe-off [119].

Using the PPAFO also caused the plantarflexion during loading response to occur later in the gait cycle than in the AFO or shoes condition, which is likely due to the design and control of the PPAFO, as opposed to an inherent effect of assistive powered plantarflexion (Figure 5.5). The PPAFO was designed to give powered dorsiflexion assistance during the loading response phase to mitigate foot slap, which it successfully applied. It is unclear if the delayed timing of the plantarflexion peak had any functionally relevant changes to the participant's gait. If the timing of the plantarflexion peak needed to be changed to better align the assistive plantarflexion with the participant's natural plantarflexion, the timing of the PPAFO controller could easily be adjusted.

A significant difference was also seen in ankle angle during pre-swing (toe-off). The AFO and PPAFO conditions had significantly smaller plantarflexion peaks than the shoes condition. A reduction in peak plantarflexion magnitude was expected in the AFO condition, as passive AFOs typically are designed to hold the foot in dorsiflexion, requiring the wearer to work against the AFO to plantarflex. This difference was unexpected in the PPAFO condition as the PPAFO is designed as a powered exoskeleton with powered plantarflexion and dorsiflexion assistance and the PPAFO was programmed to provide plantarflexion assistance at this point in the gait cycle. The lack of expected plantarflexion assistance at pre-swing (toe-off) could be due to a number of factors including needing a better timed and calibrated controller, or better physical attributes of the PPAFO. For example, the rocker bottom sole that was intended to assist with rollover during late stance; however it may have contributed to minimizing plantarflexion during pre-swing (Figure 5.1). There was no significant difference between the PPAFO and AFO conditions during pre-swing, suggesting that the powered plantarflexion and dorsiflexion assistance did not interfere with natural plantarflexion any more than a passive AFO.

Significant differences were also found in knee angle peaks and timings between footwear conditions (Figure 5.5). During the loading response, using an AFO or an exoskeleton with powered plantarflexion and dorsiflexion allowed for greater knee flexion with a later peak than the shoes condition. The increased and later flexion during the loading response was likely a result of the more controlled, more positive plantarflexion during the same phase. At mid-to-terminal stance, the knee extension angle was larger and occurred later with the powered plantarflexion and dorsiflexion than the AFO or shoes condition. This change during mid-to-terminal stance was likely due to properties of the PPAFO which provided the powered plantarflexion and dorsiflexion, such as the controller timing and the rocker-bottom rollover sole. The decrease in knee extension seen in terminal swing in the PPAFO compared to the shoes and AFO conditions may also be due to physical attributes of the PPAFO. The PPAFO has an increased weight compared to a normal shoes or AFO, which could require more energy and muscle strength to swing through a full stride, possibly preventing full extension at the knee during terminal swing [104, 105]. Other studies have observed a larger peak knee flexion during swing with a FES in persons with MS compared to a no FES condition [117]. A larger knee flexion peak in swing was, however, not seen in using an AFO or PPAFO in this study. Previous studies of passive AFOs that studied gait kinematics found no significant changes at the knee due to the use of an AFO [77].

There were no differences seen between EDSS groups in ankle angle which was unexpected, but likely due to the large variability within groups, especially in the severe group (Figure 5.6). Even though there were no significant differences, there was a trend that found decreased dorsiflexion during swing in the severely impairment group, as would be expected due to increased tibial anterior muscle weakness. Previous kinematic studies of persons with MS have found differences in ankle angle between MS groups and healthy controls, with MS groups having consistently reduced plantarflexion at loading response and toe off, and reduced dorsiflexion during swing [118, 120]. Huisinga et al. also found that the peak plantarflexion at toe off significantly correlated with EDSS score in the MS group, although our work did not directly support this relationship [118].

Differences were found in peak knee angles between groups, in that the moderately impaired group had greater knee flexion during the loading response, less knee extension during mid-to-terminal stance, and greater knee flexion during pre-swing (toe-off). These differences are likely due to the moderate EDSS group's increased mobility over the severe EDSS group. Previous studies have also observed changes in knee angles in persons with MS [118-120]. Specifically decreased knee extension at heel strike was found in two studies [119, 120]. A generally reduced knee range of motion was also found for persons with MS compared to healthy controls [119]. Confirming the differences in knee angle that were seen in this study between EDSS groups, Huisinga et al. had previously reported that peak knee flexion angle and knee range of motion were significantly correlated with EDSS score in MS patients [118].

5.6. CONCLUSIONS

These results indicate that within this study design, the use of an ankle exoskeleton with powered assistive plantarflexion and dorsiflexion did significantly change the ankle and knee kinematics of gait in persons with MS when compared to use of a prescribed passive AFO or no assistive device (shoes condition). The powered exoskeleton provided greater dorsiflexion assistance during swing, while allowing plantarflexion at pre-swing (toe-off) similar to that as the passive AFOs. EDSS group differences suggest that persons with a higher EDSS have much more impaired gait kinematics at the knee than the moderately impaired group; whereas at the ankle, the groups had similar joint kinematic patterns. The powered exoskeleton was able to provide some additional gait assistance, especially during swing and the loading response, to overcome the classic foot drop and foot slap seen with lower leg weakness.

5.7. **TABLES**

Table 5.1: Mean (standard deviation) for peak ankle magnitude and time during four phases of gait

Parameter	Phase of Gait	Footwear condition				EDSS group	
		Shoes (A)	AFO (B)	PPAFO (C)	p-value†	Moderate (D)	Severe (E)
Peak plantarflexion angle (deg)	Loading Response	-11.9 (4.6)	-8.6 (3.9)	-7.3^{A} (4.9)	0.025	-8.9 (3.1)	-9.6 (2.9)
Peak plantarflexion time (%GC)	Loading Response	4.7 (4.0)	5.9 (3.7)	$8.4^{A,B}$ (4.1)	0.018	7.3 (2.5)	5.2 (3.2)
Peak dorsiflexion angle (deg)	Terminal Stance	10.0 (5.6)	10.2 (3.9)	7.4 4.2)	0.116	10.2 (4.3)	8.1 (2.3)
Peak dorsiflexion time (%GC)	Terminal Stance	49.2 (4.8)	50.8 (3.0)	51.1 (3.5)	0.226	51.0 (2.5)	49.6 (3.3)
Peak plantarflexion angle (deg)	Pre-swing	$-16.6^{B,C}$ (13.0)	-7.6 (6.2)	-7.1 (7.0)	0.026	-11.1 (6.4)	-9.7 (4.9)
Peak plantarflexion time (%GC)	Pre-swing	70.5 (4.7)	68.9 (4.2)	69.0 (4.2)	0.420	67.7 (1.4)	71.6 (3.8)
Peak dorsiflexion angle (deg)	Terminal Swing	-5.0 (6.1)	-1.4^{A} (4.8)	$2.5^{A,B}$ (6.7)	<0.001	0.0 (4.0)	-2.8 (5.6)
Peak dorsiflexion time (%GC)	Terminal Swing	84.9 (21.3)	91.3 (4.3)	89.3 (3.5)	0.289	91.7 (2.3)	84.7 (9.1)

† p-values from follow-up univariate ANOVAs are listed since the main effect was found to be significant in the overall MANOVA.

Superscript ($^{A,\ B,\ or\ C}$) signify significant difference from condition

Table 5.2: Mean (standard deviation) for peak knee magnitude and time during four phases of gait

Parameter	Phase of Gait	Footwear condition				EDSS group		
		Shoes (A)	AFO (B)	PPAFO (C)	p-value†	Moderate (D)	Severe (E)	p-value†
Peak knee flexion angle (deg)	Loading Response	14.2 (12.8)	15.9 (13.2)	18.9^{A} (11.4)	0.026	24.54 (8.9)	7.02^{D} (7.1)	0.001
Peak knee flexion time (%GC)	Loading Response	8.8^{B} (5.3)	14.6 (5.9)	10.4 (1.1)	0.009	12.5 (2.9)	10.0 (3.5)	0.152
Peak knee extension angle (deg)	Terminal Stance	1.5 (9.2)	2.9 (8.9)	$11.0^{A,B}$ (14.6)	0.030	9.4 (7.9)	0.21^{D} (7.8)	0.041
Peak knee extension time (%GC)	Terminal Stance	40.1 (5.2)	38.9 (4.4)	$47.3^{A,B}$ (8.8)	0.004	43.6 (4.2)	40.4 (3.7)	0.142
Peak knee flexion angle (deg)	Pre-swing	41.0 (21.8)	44.3 (20.1)	43.4 (19.4)	0.516	55.8 (11.6)	28.2^{D} (15.3)	0.002
Peak knee flexion time (%GC)	Pre-swing	74.9 (7.0)	74.5 (7.0)	76.0 (6.9)	0.456	75.5 (5.9)	74.7 (7.2)	0.802
Peak knee extension angle (deg)	Terminal Swing	8.5 (11.0)	9.9 (9.4)	$12.6^{A,B}$ (9.8)	0.026	12.8 (9.9)	7.44 (9.2)	0.297
Peak knee extension time (%GC)	Terminal Swing	95.9 (4.0)	94.9 (6.6)	95.8 (3.5)	0.650	95.9 (2.6)	95.0 (4.2)	0.594

† p-values from follow-up univariate ANOVAs are listed since the main effect was found to be significant in the overall MANOVA.
Superscript ($^{A,\ B,\ C,\ D,\ or\ E}$) signify significant difference from condition

5.8. FIGURES

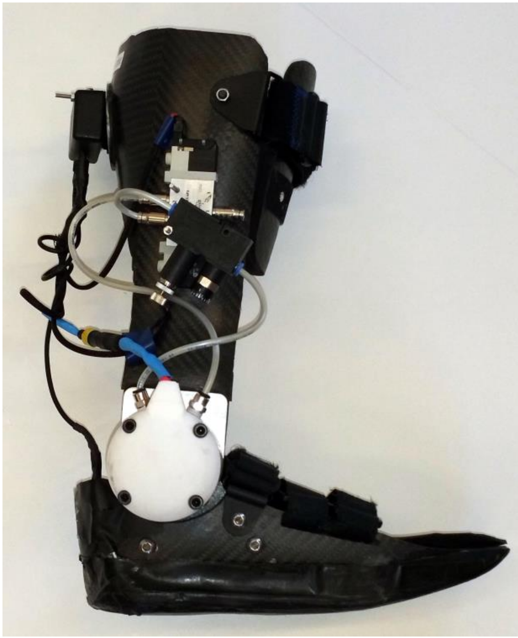

Figure 5.1: **Portable Powered Ankle-Foot Orthosis**

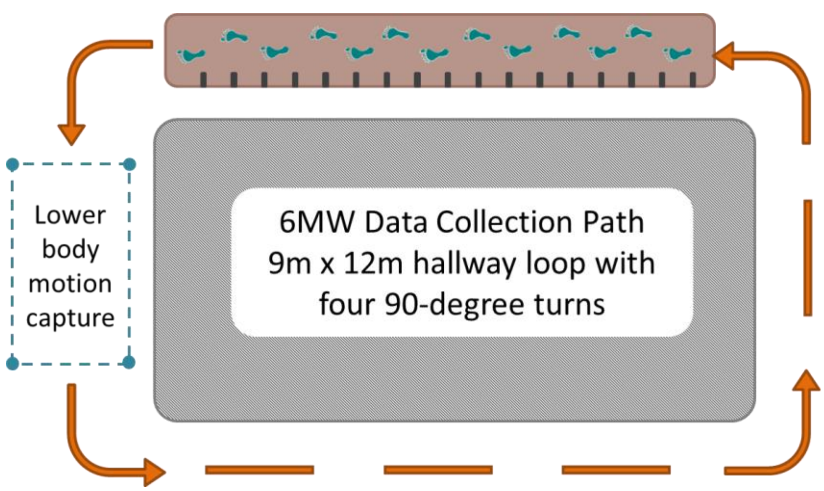

Figure 5.2: **6-minute walk test path with motion capture space in the loop**

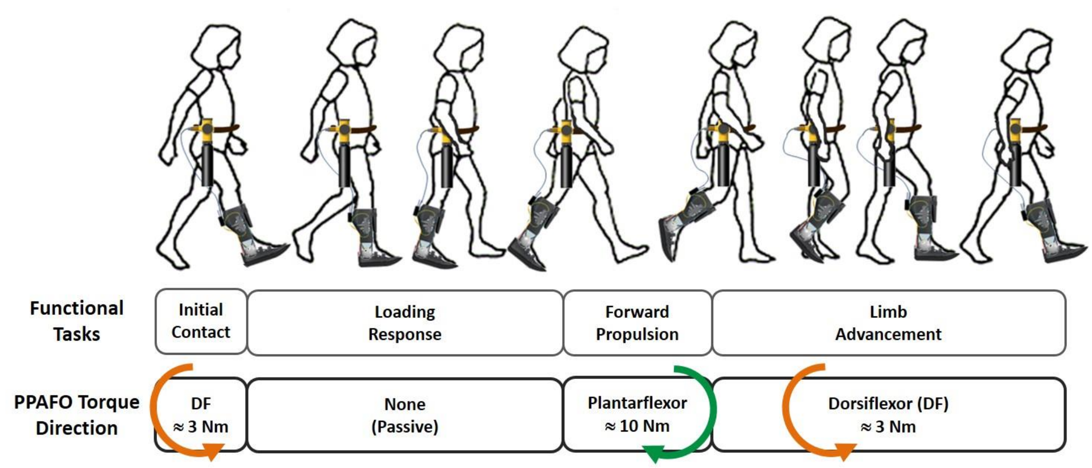

Figure 5.3: **Assistance during a gait cycle provided by the PPAFO.** Torque values are based on 100 psig during plantarflexor assistance and 30 psig during dorsiflexor (DF) assistance.
Figure adapted from Kirtley (2006)[26]

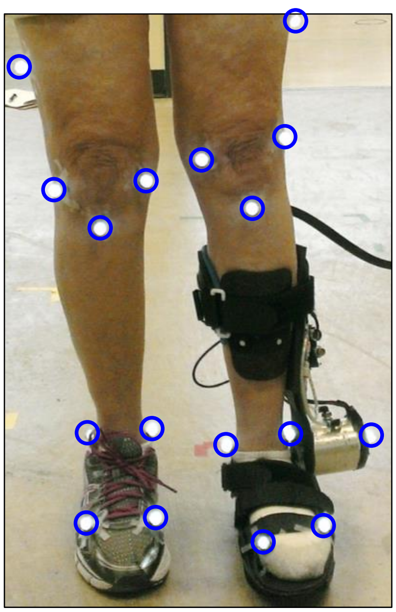

Figure 5.4: **Example of motion capture marker placement used with PPAFO**; Not shown: heel, ASIS, greater trochanter, and sacral markers.

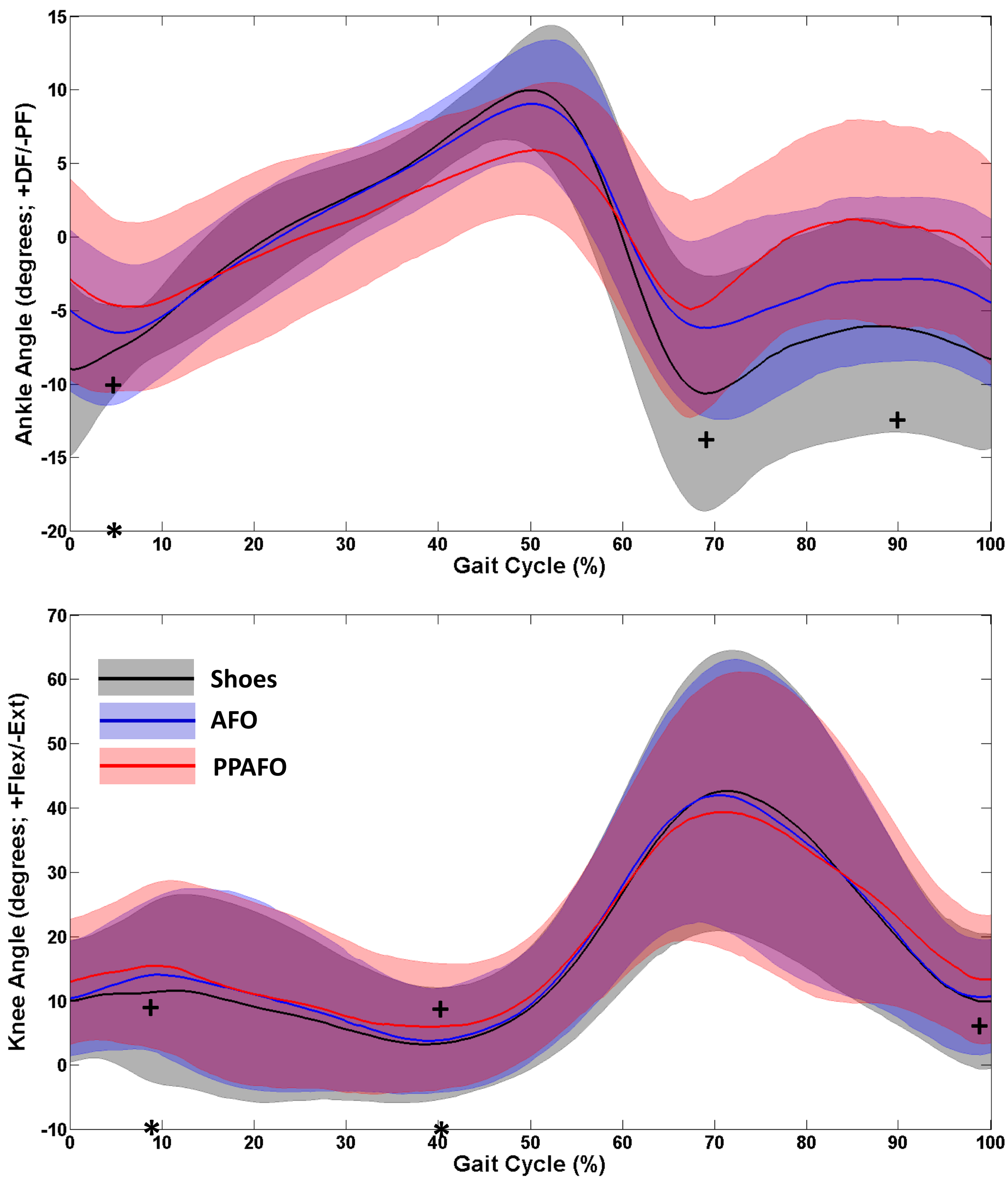

Figure 5.5: **Joint angles during gait with each footwear condition.** Line shows mean value with shaded area ± 1 standard deviation. Ankle (top), Knee (bottom). Positive/negative values indicate dorsiflexion/plantarflexion.

+ indicates significant differences in peak angle values near the angle value
* indicates significant differences in peak timing values on the x-axis.
significance was set at $p < 0.05$.

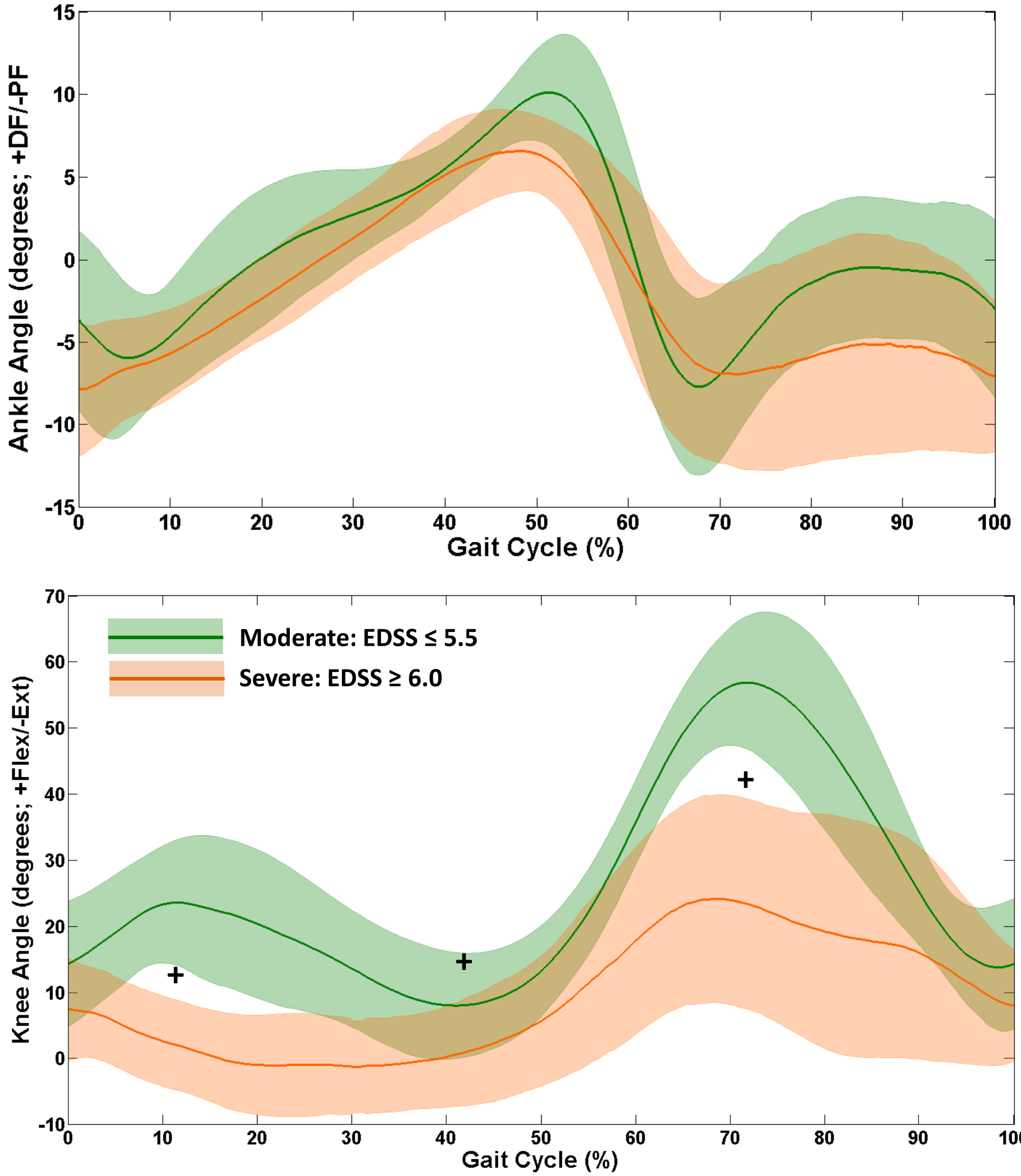

Figure 5.6: **Ankle and knee joint angles, by EDSS group, averaged over all footwear groups.** . Line shows mean value with shaded area ± standard deviation. Ankle (top), Knee (bottom). Positive/negative values indicate flexion/extension.

+ indicates significant differences in peak angle values near the angle value significance was set at $p < 0.05$.

Chapter 6 Conclusions

The work in this dissertation addressed some of the many issues in research regarding portable powered orthoses or exoskeleton robotics. Our lab has developed a pneumatically powered ankle-foot orthosis (PPAFO) to address some of the issues in portable robotic exoskeleton development [22]. Although many of the research areas span various disciplines, such as control, design, fuel and efficiency, user-interfaces, neuroscience, and kinesiology, only a few are addressed here [8, 14]. Chapter 2 provides an assessment of the available portable compressed gas tanks that could be used to power pneumatic robots, especially in the expanding field of soft-robotics [57]. Chapters 3, 4, and 5 evaluate the use of a our PPAFO during walking.

Chapter 1 provides a literature review of the current state of research regarding portable ankle exoskeletons. Specific foci included available portable power sources for soft robotics, as well as the use of robotic ankle exoskeletons in able-bodied persons and persons with gait impairment, focusing on persons with Multiple Sclerosis (MS). One issue that that persists throughout the field of research is determining consistent ways to compare the impact of powered exoskeleton devices on human movement and energy use. One common goal in exoskeleton research, regardless of target population, is to reduce the amount of energy used in walking related tasks. It is difficult to compare the effects of different exoskeletons on the energy expenditure as, there are many methods used to quantify the energy expenditure used during walking.

In chapter 2, we investigated the use of portable compressed gas tanks as a fuel source for portable pneumatically powered robotics. With the growing field of soft robotics [52, 121, 122], more pneumatic fuel sources are being designed and needed to power these robotics [57]. Although they are an old technology, portable compressed gas tanks provide a quantifiable source of fuel that have potential for powering soft robotics. Small, wearable, compressed gas tanks filled with CO_2 and N_2, which are currently commonly available, were evaluated for use in portable robots. Issues of cooling while the tank empties during use are fundamental thermodynamic issues [64]; but with N_2, the fuel tank does not cool near as quickly or to as cold of temperatures as with CO_2. The differences in cooling are due to the basic properties of the different gases when stored under high pressure at room temperature. Carbon dioxide when stored in a pressure vessel at high pressures (~760 psig) is in a two-phase state of liquid and gas, whereas N_2 when compressed stays a gas at room temperature in the portable tanks used here. Some of the benefits of CO_2 are the commercial availability of refilling the tanks and the relatively low cost of CO_2 tanks compared to high pressure air (HPA) tanks needed for compressed N_2. Overall, N_2 was preferred over CO_2 due to its warmer minimum temperature and slower rate of cooling, with similar normalized run times.

In chapter 3, we used the PPAFO to evaluate the use of bilateral bi-directionally powered ankle-foot orthoses in young, able-bodied persons while walking over-ground. The PPAFOs were programmed to provide assistance in the appropriate direction (plantarflexion or dorsiflexion) during all phases of gait. With the powered PPAFOs, participants were able to match the metabolic cost of transport needed for walking compared to the shoes condition. In previous research [2, 3, 16, 49], the goal has often been to try to reduce the metabolic needs for walking with a powered device to below that of normal walking in shoes. While completing this research, we have often wondered if reducing the metabolic cost of walking below that in shoes is an appropriate goal for these powered orthoses and exoskeletons, especially for able-bodied persons. Often the human body is already operating in the lowest energy state necessary to achieve its goal, as it is a quite sophisticated machine. An issue with trying to reduce the metabolic cost of walking with powered devices, may also be in the training that is provided on how to use the devices. Every researcher uses a different training protocol for their studies, usually based on the needs of their device controllers and participant comfort. Future work could be done in providing more intentional and specific instructions to participants towards the goal of reducing the amount of work that they need to walk. Aside from the goal of reducing metabolic cost of an already efficient human body, perhaps researches should be content with matching the low metabolic cost of normal walking, while using powered devices to add functionality to normal human gait.

Some kinematic changes were seen while using the PPAFOs, primarily at the ankle where an unexpected reduction in plantarflexion during toe-off was observed. Previous studies with prior versions of our PPAFO showed increased plantarflexion compared to shoes conditions [27, 123], so this reduction in plantarflexion was especially unexpected. It is difficult to know exactly what caused this decreased plantarflexion with the PPAFO, although a number of factors could have been at play. This was the first over-ground walking study completed with the PPAFO, and inherently, lab floor that is swept and cleaned each night is a much different surface than that of a treadmill. It is possible that participants walked much more cautiously, in a more marching gait fashion as to not slip on the slicker lab floor. It is also possible that small changes to the footwear structure of the PPAFO, primarily the rocker-bottom sole of the PPAFO caused changes in kinematics that we previously did not see. Differences in PPAFO functionality also could have caused this unexpected change in kinematics, specifically the controller timing and design. Much work has been done on other devices [38, 47]and our PPAFO [24, 91, 96{Li, 2013 #84] to determine different control strategies for the powered devices. Along with determining which sensors and signals can or should be used to capture the current state of the device and the wearer, researching the proper timing of plantarflexion assistance has also been a focus. Although the discussion and research of different controllers was out of the scope of the research in this dissertation, the impact that a well-designed and well-trained controller was not underestimated. As more research continues on the human aspect of powered orthotics and exoskeletons, a reflection back to the influence of the controller design and operation on the human behavior is essential.

A preliminary adaptation study to wearing the PPAFOs was also completed to determine the time needed to metabolically and kinematically adapt to a low-power kinematically-controlled ankle exoskeleton. There were no statistically significant differences observed between minutes 3-6 and minutes 17-20 of continuous over-ground walking with the powered devices in any spatiotemporal, kinematic, or metabolic parameters. The lack of measured differences suggests that the participants were able to adapt quickly to wearing the device, after the device controller had been trained to their gait pattern. Although this was faster adaptation than previous powered devices have reported, the PPAFO is lower powered, and controlled by kinematic input signals (compared to EMG). Another factor that may have contributed to the faster adaptation time was pre-training the controller to each participant's gait pattern. During the training time, participants walked casually in intermittent sets of 5-10 steps so that the pattern of kinematic input signals for each person could be input into the controller code. The input signal patterns then dictated the timing of the actuation of the plantarflexion and dorsiflexion assistance. It is possible that some human adaptation happened during this training time, although due to the limited nature of the walking, we believe any real adaptation to be highly unlikely. In previous research, most studies have had some loosely defined period of device training or device fitting that happens before the adaptation study. A thorough investigation into human adaptation to powered devices might need to take these training and fitting periods into account.

In chapters 4 and 5 we addressed the use of a powered bidirectional ankle-foot orthosis in persons with gait impairment due to multiple sclerosis (MS). A traditional gait assistance device in persons with MS is a passive ankle-foot orthosis (AFO) that holds the foot in a neutral position to act against foot drop, especially during swing. Participants for this study were selected based on use of a passive AFO in their daily lives.

We evaluated the use of our PPAFO in this population of persons with MS who have AFOs. The powered AFO was tested on their more impaired limb against their own AFOs, and a shoes-only no AFO condition with a series of 6 minute walk tests. The spatiotemporal and metabolic gait parameters (analyzed and presented in Chapter 4) indicated that within the study design, the participants did not overcome their gait impairment while using the PPAFO. Yet, the PPAFO did not negatively impact the COT_m used to walk. Ankle and knee sagittal plane joint kinematics were analyzed and presented in Chapter 5. The differences seen in the kinematic parameters of peak ankle and knee angle magnitudes and timings suggest that the powered exoskeleton provided greater dorsiflexion assistance during swing, while allowing plantarflexion at pre-swing (toe-off) similar to that as the passive AFOs.

Together these two analyses (Chapters 4 and 5) provide a very broad pictures of the changes of gain in an impaired population while using the PPAFO. Although the PPAFO induced kinematic changes, that seemed to possibly benefit the participant (increased dorsiflexion during swing), these changes did not translate to improved spatiotemporal outcomes, or a longer 6MW distance with the PPAFO. The powered exoskeleton was able to provide some additional gait assistance, especially during swing and the loading response, to overcome the classic foot drop and foot slap seen with lower leg weakness.

Many lessons and future ideas came out of working with the participants from our population of persons with MS compared to able-bodied participants. Persons with gait impairments have much more irregular gait patterns than able-bodied persons, and early in our research, our controller was not prepared to expect these differences. Future work on powered orthoses and exoskeletons should be carefully evaluated when transitioning from an able-bodied population to an impaired population, both in terms of device design and experiment design. The experimental design was important to evaluate as impaired populations (especially those with MS) are more likely to be fatigued sooner than an able-bodied population and cannot withstand as much testing in a single session as able-bodied persons.

Another aspect of PPAFO operation that should be considered is the noise made during the pneumatic actuation. In the generation of the PPAFO test bed used in these studies, there are no mechanical noises made by the actuator which makes for a very quiet operation, but there is pneumatic exhaust noise made every actuation cycle. Some of this noise was lessened by the use of silencers, but the exhaust noise was not totally dissipated. The noise itself does not pose an issue in research settings, although it could confound any results regarding neuromuscular function, comfortable speed choice, and adaptation to controller timings by acting as an auditory cue to the participant. Although this issue could be specific to the pneumatic noise based on the operation of the PPAFO, other researchers in the field should be aware of unintentional physical, auditory, and behavioral cures that their devices are creating, especially when drawing conclusions from human behaviors and adaptation.

Also we received much qualitative feedback about the perception of our device. In our able-bodied population, most participants were university students who were interested in engineering and were excited about testing out the PPAFO. For our population of persons with MS, their background and familiarity with engineering and technology varied, as did their reaction to the PPAFO. It could have been useful record our participant’s initial opinion of the

PPAFO to see if a negative or positive initial opinion influenced their physical outcome measures during gait testing.

As mentioned above, some of the unexpected results from these studies motivate us to further research the mechanical design of the device so that users can better match their natural gait pattern in regards to spatiotemporal and kinematic parameters. In the next generation of PPAFO design, intentional decisions and research should consider the utility of the rocker-bottom sole, the weight distribution at the ankle, and the fit and comfort of the device. In order to fundamentally understand the differences made by the powered bidirectional torque during gait, these other possible confounding factors need to be minimized.

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