The Effect of Walkway Incline Angle on Balance Control in Healthy Older Adults and Ankle Foot Orthosis Wearers with Lower Motor Neuron Pathologies

by

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Abstract

Navigating a slope carries additional demands compared to level ground, that may be amplified in older adults or individuals with mobility impairments. This study intends to understand the effect of walkway incline angle on balance control in healthy older adults and ankle foot orthosis (AFO) wearers. Twelve healthy older adults ($\geq$60 years of age) and four AFO wearers ($\geq$60 years of age) ascended and descended slopes ranging from 0° to 8°. AFO wearers completed the protocol with and without their AFOs. Older adults and AFO wearers had a minimum margin of stability that was larger during descent and less during ascent. AFO wearers walked more cautiously, with a wider stance and larger margin of stability in the anterior direction. The results may provide clinicians with information on how orthosis use affects dynamic balance during sloped gait and provides a basis for future work informing environmental design.

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Chapter 1
Introduction, Objectives, and Hypotheses

1 Introduction

1.1 Motivation

Fall risk is influenced by a number of factors including age, the environment, and the presence of a neurological pathology [2, 3, 4]. Navigating environmental features such as slopes carries additional challenges including increased friction demands, and increased range of motion at the hip, knee and ankle, [5, 6]. Walking down a slope is particularly risky because there are increased shear forces near heel strike and during early stance, and a decreased ability to recover if a slip should occur [6]. In a study of postal workers, falls occurred 30% of the time walking down a sloped drive compared to 2% when walking up [3].

Given the elevated fall risk associated with sloped walking, it is important to understand the biomechanical task demands, including the effects on balance control. However, to date there is limited research investigating the impact of sloped gait on balance measures in older adults. One study has reported margin of stability in older adults, although this particular study was limited to only one incline angle using a treadmill [7]. In another study, a greater range of angles were investigated, reporting the COP-COM inclination angle, although results were limited to ascending slopes only. In the natural environment, slopes exist in various angles, requiring both ascent and descent; investigating the impact of several walkway angles would provide a more complete picture of how walking on slopes affects dynamic balance.

In addition to the challenge of walking on slopes, individuals with neurological pathologies are at an increased risk of falls due to several factors such as an underlying gait disorder, sensory deficit, and/or loss of consciousness due to a seizure or syncope [4]. Many individuals with neurological pathologies wear ankle foot orthoses (AFOs) to correct or prevent deformity and improve function [8]. Information on AFO usage is scarce due to a lack of easily accessible records and the complex nature of orthotic treatment [9]. However, available statistics state that in the year 2000, 3,514,000 individuals in the United States wore orthoses, with 68% of orthoses classified as “leg orthoses” [10]. It has been predicted that the number of people needing orthotic treatment will rise by 31% between 1995 and 2020 [10]. AFO wearers will generally perform

activities of daily living while wearing their devices, including walking on slopes [11]. Given the demands of sloped gait, it is unsurprising that past research has shown walking on slopes affects spatiotemporal aspects of gait in AFO wearers [12]. However, currently it is unknown how walking on a slope affects dynamic balance in AFO wearers.

1.2 Objectives and Scope

This study aims to understand the effect of walkway incline angle on balance control in healthy older adults and ankle foot orthoses wearers. The specific objectives of this study are to: (1) investigate the effect of walkway incline angle during slope ascent and descent on the margin of stability in healthy older adults, and (2) to compare the margin of stability between AFO wearers with lower motor neuron pathologies and healthy older adults, across ascent and descent slope angles.

1.3 Hypotheses

In the sagittal plane, healthy older adults will walk more cautiously to deal with the risk of slope descent. Whereas for slope ascent, healthy older adults will adopt a more unstable posture to help preserve forward momentum.

Specifically:

- As slope angle increases during descent, minimum anterior margin of stability will increase.
- As slope angle increases during ascent, minimum anterior margin of stability will decrease
- As slope angle increases during descent, minimum lateral margin of stability will decrease
- As slope angle increases during ascent minimum lateral margin of stability will increase

AFO wearers will be more unstable compared to healthy older adults.

Specifically:

- AFO wearers will have smaller minimum anterior and lateral margin of stability compared to healthy older adults, and these differences will increase with increasing slope angle.

Chapter 2
Review of the Literature

2 Review of the Literature

2.1 Balance

The ability to maintain balance is critical for several activities including safe ambulation and the completion of activities of daily living [13]. To remain upright, the individual must control the position and velocity in relation of their centre of mass (COM) to their base of support (BOS) [14]. During standing tasks, factors such as increasing age [15], proprioceptive decline [15], history of falls [16], and the presence of neuropathy [17] is associated with increased COM sway area [15]. Considering the latter, while not all individuals with neuropathy are shown to be equally unstable during quiet stance, certain neurological conditions result in increased sway compared to healthy participants [17]. When the COM exceeds the BOS during standing, a reaction must occur for balance to be maintained – this can take the form of muscle generation [18], a reaching reaction [19], or a stepping reaction [19] . During more dynamic tasks, such as walking, the demands on the body's COM differ from standing tasks. During walking, the COM moves outside of the base of support during single limb support [20]. However, the COM velocity plays a role in dynamic balance control [14]. Hof et al. introduced the concept of the extrapolated center of mass, which takes into account both position and velocity of the center of mass [21], which helps to further understand how balance is controlled during dynamic tasks. The margin of stability is the difference between the border of the base of support [21] and the extrapolated COM. This concept will be discussed further in Section 2.3.2.

2.2 Balance and Gait

2.2.1 Normal Bipedal Gait

Normal human gait is a process involving the coordination of several muscles and joints to move the body along a desired line of progression [22]. A gait cycle consists of one heel strike to the next ipsilateral heel strike. During level ground walking, the following eight phases have been described (see Figure 2.1), with details specific to the general gait of healthy adults [22]:

- *Initial contact* is the instant when the heel makes contact with the ground *Loading response* is the first period of double limb support, when weight is transferred quickly onto the outstretched

limb and foot rapidly plantarflexes to ~five degrees; an action is known as the “heel rocker”. The pretibial muscles contract to control the plantarflexion torque, pulling the tibia forward, creating knee flexion. The knee flexes to ~fifteen degrees, maintaining limb stability and absorbing shock, while the hip remains stable. Late in the loading response, the gastrocnemius and soleus contract to control the advancement of the tibia.

- During *mid stance*, a period of single limb support, the body progresses over a stable limb as the soleus and gastrocnemius control tibial advancement. This action is called the “ankle rocker”.
- In *terminal stance*, weight is transferred onto the forefoot, an action known as the “forefoot rocker”, and the body moves ahead of the stance limb. This is a period of peak calf muscle activity, which controls dorsiflexion and allows the heel to rise, increasing contralateral step length.
- *Pre-swing* initiates the second period of double limb support and involves weight quickly being transferred from the ipsilateral limb to the contralateral limb. The knee flexes to ~forty degrees with minimal muscle activity.

  During *initial swing*, the knee flexes, and the foot comes off the floor as the thigh advances forward. *Mid swing* involves the knee extension necessary for maximizing step length, and the ankle returns to neutral as the foot clears the ground. *Terminal swing* prepares the ipsilateral limb for initial contact. The ankle remains at neutral, the hip falls to ~twenty degrees of flexion, and the knee extends to neutral.

Alternatively, the gait cycle can be divided into three functional tasks: weight acceptance, single leg support, and swing advancement [22]:

- Weight acceptance includes initial contact and loading response. This is a period of double limb support focused on shock absorption and maintaining stability while continuing forward progression.
- Single limb support includes mid stance and terminal stance, and involves moving the body over one stable limb.
- Swing limb advancement includes pre-swing, initial swing, mid swing, and terminal swing. Swing limb advancement involves unloading of the ipsilateral limb, and moving it to the front of the body in preparation of taking the next step.

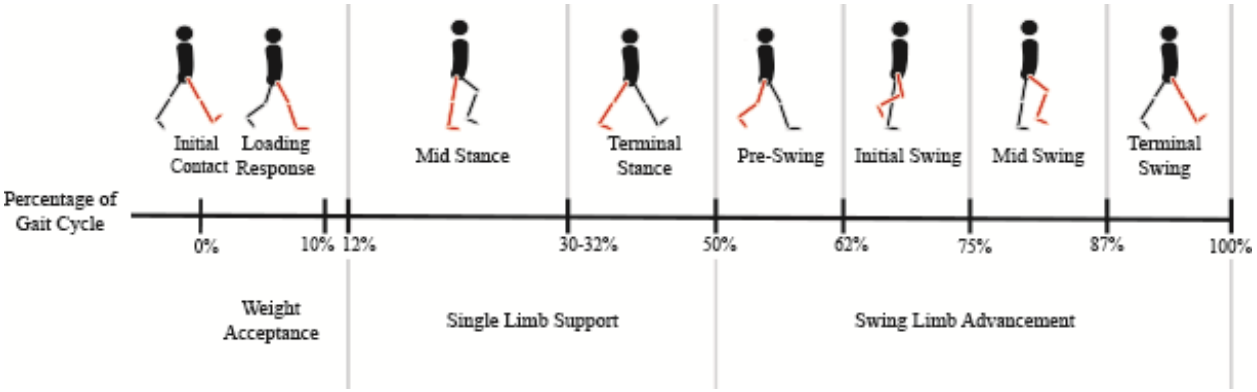

Figure 2.1 The Gait Cycle (adapted from [22]**)**

2.2.2 Dynamic Balance During Gait

Compared to standing balance, balance during gait carries a different set of requirements. During quiet standing, postural stability is defined as the ability to keep the center of mass within the base of support [23]. However, walking forward involves accelerating the body's center of gravity (COG) ahead of the base of support [24]. Therefore, safe foot placement is vital for balance maintenance during gait [24]. In addition to safe foot placement, the body must maintain trunk alignment over a shifting base of support, while moving along the desired line of progression [25].

2.2.3 Aging and Balance During Gait

Compared to younger adults, older adults adopt a more cautious gait pattern that aims to increase stability [26], which is reflected in differences in temporal spatial parameters. For example, older adults age 85+ spend more time in double limb support compared to 70-84 year old adults [27], who also spend greater time in double limb support than younger adults [26, 27]. Older adults also walk with a slower gait speed [26], [28], and wider steps [29]. Begg and Sparrow found that normalized step length did not show age affects in a population of healthy older adults (2006). However, other studies found that stride length [28], and step length [30] decreased in older adults compared to healthy younger adults. These differences in temporal spatial metrics have been shown to be associated with fear of falling and increased fall risk [31].

Centre of mass (COM) control also changes with age [32]. Mediolateral centre of mass displacement and velocity increase with increasing age during gait, suggesting decreased COM

control [32]. Thus it is not surprising that in general older adults walk with a more cautious or conservative gait, as shown by a greater margin of stability in the anterior-posterior direction [33], or by a smaller separation between the COM and the BOS during single limb support [20]. Dynamic stability during level ground walking also does not appear to decrease proportionately by age in older adults [34]. For example, a study of margin of stability, where higher numbers meant that participants were more stable against backwards balance loss, found little change among adults aged 65-85, with significant decline at 85+years [34]. These differences highlight the increased fall risk with older age.

2.3 Balance, Gait, and Slopes

2.3.1 Physical Demands of Sloped Gait

Compared to level ground, sloped gait carries a distinctive set of requirements to ensure safe forward progression. In early and late stance, there are periods of high horizontal force, resulting in high friction demands [5]. Additionally, sloped gait involves different work requirements, demands on foot clearance, the controlled raising or lowering of the body's center of mass (COM), and postural adjustments [5, 35].

During ascent, the lower limbs must work to drive the body upward and forward. The trunk leans forward to aid with the generation of forward momentum [35]. A greater flexor-extensor moment about the hip from 0-30% of the gait cycle has been shown, which is associated with assisting the forward and upward motion of the body [5]. The second ankle power phase, associated with push off, is larger for inclined gait compared to level ground or downhill walking [5]. This large muscular power demand has been supported by increased contributions by soleus for propulsion on an incline compared to level ground [36]. Walking up an incline also carries additional toe clearance demands [5]. To meet the larger required toe clearance, an increase in knee and hip flexion has been shown [5]; increased dorsiflexion during the first 50% of the gait cycle has also been reported when compared to level ground walking [5].

During descent, the body works to lower the COM in a controlled fashion. About the hip, there is quick hip flexion at heel strike, where the hip contracts eccentrically [5]. This power absorption phase aids in controlling the downward motion of the body during early stance. At the knee, the rectus femoris, vastus lateralis, medialis, intermedis absorb power from the ipsilateral leg

providing more braking during descent, compared to the muscular demands associated with level ground walking [36]. These differences in power absorption during descent have been shown to be associated with longer duration of activity by these muscles, delivering more power to the trunk. At the ankle, soleus also absorbs more power during descent [36]. During declined gait, the ankle is more dorsiflexed at about 50% of the gait cycle, and less plantarflexed at toe off compared to level walking. There is also an ankle power absorption phase during early stance that only occurs during downhill conditions [5]. This phase slows the downward motion of the body and controls the anterior rotation of the tibia [5].

2.3.2 Balance and Sloped Gait

In addition to the distinct physical demands of sloped gait, both inclined and declined gait present with additional challenges to balance. Downhill walking in particular can be risky because of the increased shear forces, and the decreased ability to recover should a slip occur [6]. In one study of walking on level ground and descending slopes with oily contaminants in a lab it was shown that during level ground walking four out of sixteen slips resulted in a fall, whereas, for a 5 degree slope seven participants slipped and recovered and nine participants lost balance and fell [37]. For a 10 degree ramp with oil contaminant, all trials resulted in a fall [37]. In another study that followed falls in postal workers, 30% of falls occurred during a slip while descending a sloped drive compared to 2% walking up a sloped drive [3]. To accommodate the challenge and improve balance control, individuals appear to take smaller steps during downhill walking [38, 7]. It has been shown that smaller step lengths can affect angular momentum through changing the external moment arm and the ground reaction force (GRF) [38]. By taking smaller steps during slope descent, individuals can more tightly control the range of angular momentum during descent in the transverse and sagittal planes compared to level ground walking [38]. Step length has also been shown to affect margin of stability during treadmill gait [39] where an increase in the step length results in lower anterior margin of stability during heel strike, when compared to “normal steps”, but a higher margin of stability during the first half of single limb support [40]. Lateral margin of stability was influenced by stride width, with narrower steps resulting in lower margin of stability during heel strike [40].

Margin of stability is the distance between the extrapolated COM (XCOM) and the border of the base of support (BOS) [21]. During quiet standing, stability is achieved when the COM is within

the BOS [23]. However, in dynamic situations, COM velocity can also affect whether someone can achieve balance [14]. The XCOM takes into account both the position and the velocity of the COM through the addition of the term $\frac{\mathrm{vCOM}}{\sqrt{\mathrm{gl}^{-1}}}$ [21]. The XCOM is defined as:

$$\mathrm{XCOM} = \mathrm{pCOM} + \frac{\mathrm{vCOM}}{\sqrt{\mathrm{gl}^{-1}}}$$

Where pCOM is the vertical projection of the COM to the ground, vCOM is the velocity of the COM, g is the acceleration due to gravity, and l is the distance between the COM and the ankle joint.

Margin of stability has been used previously in level ground walking studies [40, 39], identifying risks in stair dimensions [41], sit to walk [42], and cell phone use during gait [43]. Although margin of stability has been previously used in studies of inclined and declined walking [44] [7], there are limitations of using margin of stability as an outcome measure of dynamic balance control on slopes [45]. For example, the body behaves less like an inverted pendulum during sloped gait, and XCOM does not incorporate angular momentum, which may result in an underestimation of stability in the sagittal plane [45].

To date, there is limited research investigating the effect of sloped walking on measures of dynamic balance. One study has compared level ground walking and walking up and down multiple slope angles (3.43, 4.57, 5.71 degrees) on a treadmill, which showed changes in both frontal and sagittal plane margin of stability in healthy young adults [44]. During descent, the XCOM exceeded the anterior border of the BOS by more during heel strike compared to level ground and inclined gait for all angles [44]. The authors suggested that the changes in anterior MOS were due to participants taking smaller steps during descent [44]. In the frontal plane, the XCOM during descent was also closer to the lateral border of the foot during foot contact compared to ascent [44]. The authors suggested that the reduced margins of stability during slope descent were due to increased movement of the pelvis, which leads to larger and faster pelvic oscillations, influencing the margin of stability calculations [44]. Building on this work, the same research group then included older adults in a follow up, investigating one slope angle only (4.57 degrees) [7]. Older adults showed a similar trend for both frontal and sagittal plane margin of

stability compared to young adults during downhill walking, with no significant age effects between older and younger adults found for the slope angle tested [7]

In contrast to downhill walking, falls resulting from slips are less likely to occur during slope ascent since peak shear forces occur during push-off, when slips are less likely to occur [46]. However, uphill gait carries additional demands not seen during descent [5]. For example, older adults demonstrate significantly larger variability in minimum toe clearance for uphill walking compared to downhill walking [47]. Considering more detailed measures of dynamic balance, walking uphill on a treadmill at angles of 3.43, 4.57, and 5.71 degrees has previously resulted in larger margins of stability in the frontal plane compared to slope descent [44], as the XCOM is positioned further from the lateral border of the foot. In the sagittal plane, inclined walking has also shown the XCOM to exceed the anterior BOS during heel contact by less compared to level ground walking and descent [44]. Similar to treadmill slope walking, overground uphill walking (on 5, 10, and 15 degree slopes) has also resulted in differences in temporal-spatial measures, and measures of balance (such as the COM-centre of pressure (COP) inclination angles (COM-COP inclination angle)) when compared to level ground gait [48]. The COP-COM inclination angle is the orientation of the line that connects the COM and the COP and has been used to describe dynamic control of the body. When walking uphill, both older and younger adults demonstrate generally decreased cadence, walking speed, and duration of single limb support, while also increasing the time spent in double limb support, compared to level ground walking [48]. Compared to younger adults, older adults also increase step width and time spent in double limb support, but decrease cadence, walking speed, step length, and duration of single limb support for all slope conditions [48]. In this particular study, although no significant differences were found between slope conditions in both younger and older adults in the sagittal and frontal planes for the COM-COP inclination angle [48], the temporal-spatial modifications during slope ascent were found to influence the rate of change of the inclination angle at heel strike, which decreased linearly with increasing slope angles during ascent in both the frontal and sagittal planes [48]. It is likely that through spending more time in double limb support and decreasing the rate of change of the inclination angle, participants were able to maintain the same COM-COP inclination angles as level ground walking to improve stability during the more challenging task of walking uphill [48].

As summarized above, previous literature has begun to investigate the impact of sloped gait on balance control in older adults. However, questions still remain. Existing literature on margin of stability was performed on a treadmill, which differs from over-ground walking in other measures of balance control. For example, on level ground individuals had a greater range of COM-COP inclination angles in the frontal plane and greater peak rate of change of the inclination angle on a treadmill compared to level ground [49]. This may indicate that medio-lateral control is more difficult on a moving surface [49]. The movement of the treadmill belt has also resulted in differences in both COM-COP inclination angles and the rate of change of the inclination angle in the sagittal plane [49], where overground walking results in a larger posterior rate of change at heel strike because the COP moves more anteriorly as the heel rises at the end of single limb support [49]. Conversely, treadmill walking results in an anteriorly-directed rate of change of the inclination angle because the posterior velocity of the belt drives the COP posteriorly [49]. Despite these differences between testing paradigms with several measures of balance control, other studies during level ground overground and level ground treadmill gait showed similar values for measures of stability involving the XCOM and margin of stability [50, 51]. However, it is possible that walking on an incline or decline would yield different results when considering the margin of stability. Existing studies on margin of stability in older adults during sloped walking have also studied only one slope angle [7]. Given that slopes exist in a variety of angles throughout naturally occurring and built environments, investigation including a variety of slopes would provide a more thorough understanding of the effect of slope on balance.

2.4 Ankle Foot Orthoses

In addition to the need to understand balance control during sloped gait as it relates to healthy aging, it has been also been reported that individuals with conditions affecting mobility have difficulty navigating slopes [52, 53] , For example, in a study of limb salvage patients, AFOs, and uphill gait, it was found that not enough participants were able to complete the protocol without their AFOs to provide a comparison between an AFO and no AFO condition; in this study measures of balance were not investigated [53]. Additionally, a study of older adults who were transtibial amputees found that amputees had difficulty climbing slopes due to reduced ankle range of motion and power, and the stiffness of the heel region of the prosthetic foot could increase the time between heel contact and foot flat, putting more stability demands on the

residual limb [52]. In order to more fully understand incline and decline walking, investigation of sloped gait and balance should also be expanded to include a broader range of pathologies - for example persons who use ankle foot orthoses, where sloped gait is a reported issue although the impact of the device use on balance is currently unknown. AFO use and gait is described more thoroughly in the following sections.

2.4.1 Introduction to Ankle Foot Orthoses

An orthosis is a device used to improve the function of a joint, or to support, align, prevent, or correct a deformity [8]. An AFO covers some portion of the leg and foot and controls movement and alignment at the foot and ankle [8]. AFOs are prescribed to aid with various pathologies including stroke, cerebral palsy, and various lower motor neuron pathologies [54, 55, 56]. Orthoses prescription and design depends on a number of factors including physical capability, patient and clinician goals, and comorbidities [8].

There are several common types of AFOs, including solid, single axis, energy storage and return (ESR), posterior leaf spring (PLS), and spiral. See Appendix A for figures and descriptions of various AFO types.

2.4.2 Biomechanics of Ankle Foot Orthoses

Orthoses operate on a series of forces and counter forces called pressure systems. AFOs usually act on a three-point pressure system, which involves one principal force and two counter forces; one proximally, the other distally, acting in opposition to the principal force [8]. These three-point pressure systems help to ensure proper alignment or prevent unwanted motion [8]. Controlling motion at the ankle can affect the relationship between the ground reaction force and the body, which can affect the direction and magnitude of joint moments at the ankle and other more proximal joints [55].

2.5 Ankle Foot Orthoses and Function

2.5.1 AFO Use and Level Ground Gait

Past literature has demonstrated the important impact of AFO use during level ground walking. In particular, AFO use, and design characteristics of the AFO, may affect spatiotemporal, kinematic, and kinetic aspects of walking. These are described below.

2.5.1.1 Spatiotemporal

AFO use has been shown to improve several spatiotemporal gait parameters in various clinical populations. In a study of individuals with Charcot-Marie-Tooth, carbon fiber AFOs were shown to improve gait velocity and cadence in six of eight of the participants, and stride and step lengths in seven of eight of participants [57]. Additionally, in a study of individuals with hemiparesis following a stroke, improved step length, stride length, and mean comfortable speed was noted with AFO use [58]. However, not all studies reported improvements in spatiotemporal aspects of gait. In studies involving individuals with multiple sclerosis, there were either no significant differences in terms of ambulation velocity [59], or slower gait speed [60] while wearing AFOs during a timed walk test. Participants were slowest while wearing the solid AFOs compared to wearing single axis AFOs [60]

2.5.1.2 Kinematics

AFO use has been previously shown to affect lower limb kinematics during level ground walking. In a study of individuals with stroke and multiple sclerosis wearing carbon composite AFOs during level ground gait, there was a significant decrease in ankle range of motion compared to without the AFO, stemming from less dorsiflexion in the middle of stance phase, and no plantarflexion at the end of stance phase [61]. During hemiplegic gait, AFOs decreased plantarflexion during swing phase for improved toe clearance [62]. There were no significant differences in knee or hip kinematics with AFO use [61]. When specifically investigating various AFO types, or design characteristics of the AFO, results have also shown kinematic differences during level ground gait. For example, although hip kinematics did not change [63], a slight decrease in peak knee flexion during early single leg support was noted when the AFO provided had a more compliant strut condition [63]. Strut stiffness also affected ankle kinematics at certain parts of the gait cycle [63], where ankle plantarflexion increased from the compliant to the nominal condition during heel strike and early stance; peak dorsiflexion significantly decreased for stiff compared to nominal condition, during second double support [63]. The location of the strut's bending axis, comparatively, did not affect plantarflexion angles during level ground gait to a clinically meaningful degree [64] Ankle joint type and angle affected more proximal kinematics. In participants with a peroneal nerve block, the knee and hip were more flexed with the nerve block [65]. None of the ankle joint conditions brought the knee and hip flexion back to

normal, but a 5 degree dorsiflexion stop and spring assist AFO brought the hip flexion closer to normal levels, and also reduced maximum foot clearance to normal levels [65].

2.5.1.3 Kinetics

Similar to effects on kinematic measures, AFO use has been shown to influence kinetic outcomes to varying degrees. During level ground walking, significantly lower plantarflexion moments at the ankle and abduction moments at the knee were reported for healthy participants wearing solid AFOs compared to non-AFO use gait [66]. Conversely, in a study of participants with Charcot Marie tooth wearing carbon fiber AFOs, ankle joint power was significantly greater during the maximum support moment compared to a non-AFO use condition and was significantly greater under the braced condition for most participants [57]. Foot plate length also affected knee kinetics in hemiplegic stroke patients [67]. A full-length foot plate increased the peak knee extensor moment in participants with knee hyperextension, however, a three-quarter length foot plate did not [67]. Finally, in healthy individuals that underwent peroneal nerve blocks, when the ankle was held in place with a pin-stop, plantarflexion moments were restored to normal levels during gait; whereas the spring assisted AFO did not result in similar "normalized" joint moment profiles [65].

2.6 Ankle Foot Orthoses and Balance

AFO design, the population wearing the AFO, and the task being performed have all been shown to influence balance. AFOs are designed to restrict motion in one or more planes [68]. This could aid balance if the AFO blocks an inappropriate motion either due to a compromised joint or weak muscle [68], although could also interrupt balance mechanisms if the stiffness of the material interferes with an individual's ability to respond to a perturbation, or the material is not stiff enough to provide sufficient structural support [69].

2.6.1 Static Balance and AFO Use

Several studies have reported improvements in static balance with AFO use. For example, Wang et al. (2005) found that individuals with hemiparesis for less than six months had less body sway on a foam surface while wearing an off-the-shelf plastic AFO than compared to when they were not wearing an AFO [54]. However, in the same study, there was no significant difference between AFO and non-AFO conditions in individuals with hemiparesis of twelve or more

months [54]. In another study of individuals with multiple sclerosis, both solid and single axis AFOs improved standing balance [60]. It has been suggested that AFOs may improve static balance by providing sensory cues to the lower limbs [70, 71]. However, these studies had limitations including uncertainty that improved balance did not come from the additional mechanical support of the AFO.

2.6.2 Dynamic Balance and AFO Use

Although the investigation of AFO use and consideration of dynamic balance is limited, there are a few studies which have investigated AFOs and balance in response to perturbations and during gait. Generally, it has been shown that responses to perturbations improved for AFO wearers while wearing the appropriate AFO [72]. However, during “normal” treadmill gait, wearing a solid AFO resulted in higher range of angular momentum in both the frontal and sagittal planes for healthy adults [66]. Since regulation of angular momentum has been shown to play an important role in fall prevention [73], these results highlight the balance-associated challenges faced by AFO wearers. Similarly, it was found that participants wearing ”standard short AFOs” (see Appendix A) had decreased tibialis anterior and soleus activity and increased distance between the COM and COP during step execution compared to without the AFOs [74], further emphasizing the challenging nature of AFO gait.

2.7 Ankle Foot Orthoses and Sloped Gait

The literature on AFO use and sloped gait is very scarce. To date, only one study has directly compared walking on slopes to walking on level ground in AFO wearers. Lewallen et al. (2010) found that for individuals with stroke who wore AFOs, walking on slopes affected certain spatiotemporal aspects of gait, with slope descent resulting in slower walking speeds and shorter step lengths compared to level walking or slope ascent [12]. AFO type and design has also been investigated during sloped gait. Lewallen et al (2010) found that compared to the posterior leaf spring (PLS) AFO, single axis AFO, and shoes only, wearing a solid AFO resulted in the slowest speed and shortest affected and unaffected step length for individuals with stroke. Another study investigated strut stiffness for AFOs during uphill walking only [53]. Among strut conditions, there were limited kinematic and kinetic differences during uphill walking and asymmetries still existed between the affected and unaffected sides [53]. There were also no spatiotemporal differences between the carbon fiber energy storage and return AFO wearers and the able-bodied

controls [53]. Finally, when investigating AFO design and sloped gait while ascending and descending slopes for three different types of AFOs with various trim lines and hinges [75], in individuals with osteoarthritis, differences in forefoot and hindfoot range of motion were found, which suggest design specific characteristics of the AFO can provide varying levels of forefoot and hindfoot movement restriction during sloped gait.

Based on the past literature, it is clear that AFO use impacts several parameters of gait during sloped walking. However, to our knowledge there are no studies that investigate the effect of sloped gait on dynamic balance in AFO users. Such information will provide clinicians with information on how orthotic intervention affects their patients while walking on slopes and guide safe walkway design.

Chapter 3
Methods

3 Methods

3.1 Sample Population and Inclusion/Exclusion Criteria

Four AFO wearers (18+ years) and 12 healthy older adults (60+ years) were recruited for this study. AFO wearers were recruited using posted advertisements in orthotics clinics in the Greater Toronto Area. Healthy older adults were recruited through posted advertisements in community centres.

To be included in the study, all participants were within the specified age range. Participants in the AFO group were diagnosed with a lower motor neuron pathology (self-report), and wore either an energy storage and return, or flexible, AFO (unilateral or bilateral wearer) (Figure 3.1). All participants were able to comprehend instructions given in English. Exclusion criteria included known vestibular disease, plantarflexion tone (>1 on the Modified Ashworth scale), and self-reported osteoporosis or a recent injury that would increase the risk of injury if an individual fell into a safety harness.

Figure 3.1 A flexible AFO [1] **(left) and an Energy Storage and Return (ESR) AFO (right)**

3.2 Sample Size Calculation

To date, no studies have evaluated measures of balance control in AFO users during sloped walking. Therefore, no formal sample size calculation was conducted for this initial work. However, based on previous AFO and walking studies, 10 AFO users and 10 age-matched healthy control participants were estimated to be included in this study, with the current results serving to establish baseline data and to inform future sample size calculations.

3.3 Experimental Environment

The study took place in the RampLab in Toronto Rehabilitation Institute's Challenging Environment Assessment Laboratory. The lab consists of a 5x5m platform. In the center of the lab, a 4.8 m long, 1.8 m wide walkway was positioned, consisting of four force plates (AMTI, Watertown MA, USA) and four wooden blanks (Figure 3.2). A fifteen camera, 3D motion capture system (Motion Analysis, Santa Rosa, CA USA) recorded kinematic data, sampled at a rate of 250 Hz. Two video cameras (FLIR Integrated Imaging solutions Inc., Richmond BC, Canada) were also positioned within the lab space, permitting the recording of video data throughout the data collection sessions. For safety, all participants were harnessed into an overhead harness system with fall-arresting capabilities and the walkway was equipped with a handrail.

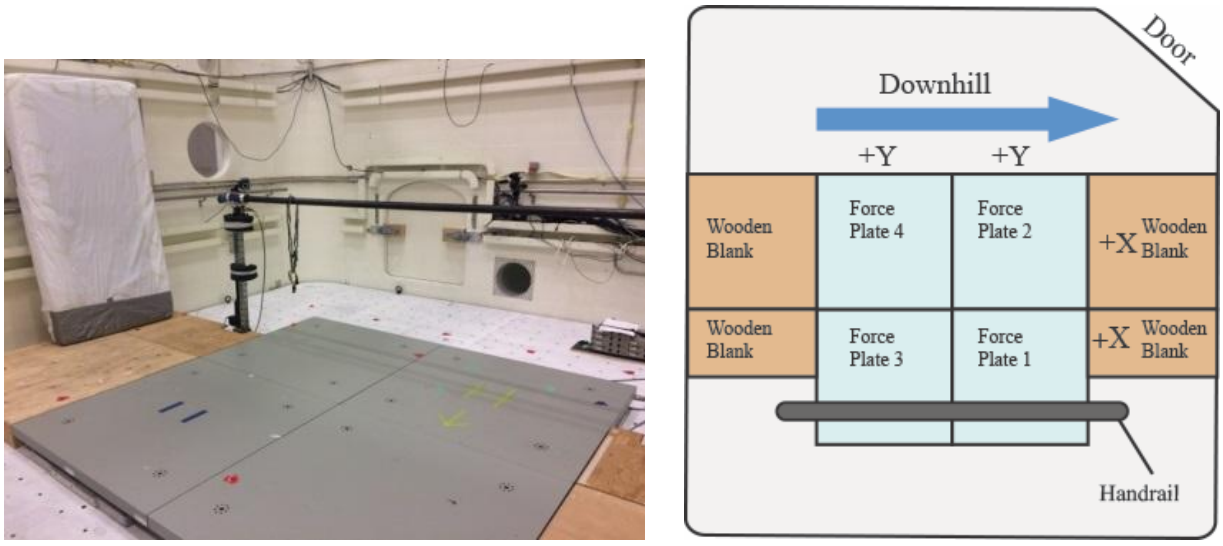

Figure 3.2. A photograph of the interior of the lab (left) and a labelled diagram of the lab's interior (right)

3.4 Participant Instrumentation

Participants were instrumented with small reflective markers to track the body's motion in three-dimensional space (tracking markers) (Figure 3.3). These reflective markers were individually secured on the participants' shoes, upper and lower limbs. Rigid clusters of three to four non-collinear markers were also attached bilaterally to the participant's shanks, thighs, and to the pelvis, and trunk. Additional markers were used to identify segment endpoints for the determination of the biomechanical model (described in detail in section 3.6 below). These removable markers were present for a single, static pose only and were removed during the walking trials.

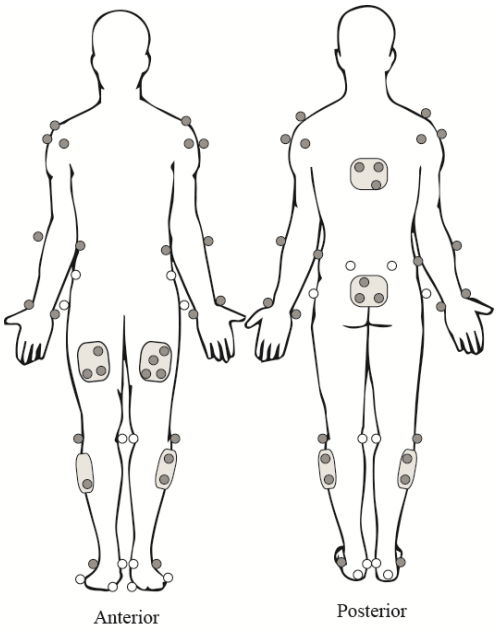

Figure 3.3 Participant setup, including locations of the tracking markers (grey) and removable markers (white). The removable markers were used only for the purposes of the determination of segment endpoints during a single static pose

3.5 Experimental Protocol

All participants underwent pre-screening via a telephone call to ensure that they fit the inclusion/exclusion criteria. Once it was determined the participant was eligible for the study, they were invited to visit the laboratory for a single visit lasting approximately 2.5 hours. Following informed consent, participants completed a questionnaire including information related to participant demographics, AFO type, history of use, and common slope usage. Clinical tests were then conducted to determine clinical measures of muscle tone and function:

- Modified Ashworth Scale: Gastrocnemius muscle tone was measured bilaterally for both groups using the Modified Ashworth scale [76]. Participants were excluded from the study if muscle tone exceeded 1 on this scale.
- The Timed Up and GO (TUG) test: The TUG test involves the participant rising from a chair, walking 3 meters, turning around and walking back, and sitting down. It is often used to test functional mobility in older adults [77], and has been shown to be a valid measure for risk of falls in community dwelling older adults [78].

Following the conduct of the clinical tests and instrumentation, participants were instructed to stand in the center of the lab space in an anatomical position to collect a single static trial. This static trial was used to determine the average location of the markers and define the segment coordinate systems. The additional static-only markers were then removed from the participant set-up, and walking trials commenced. For the walking trials, participants were asked to walk back and forth along the length of the platform three times for each of five slope angles. Walkway angles included 0, 2, 4, 6, and 8 degrees and were block randomized to reduce effects of learning or fatigue.. Participants were also asked about the difficulty of the task and perceived exertion after each slope condition using a visual analog scale (Figure 3.4).

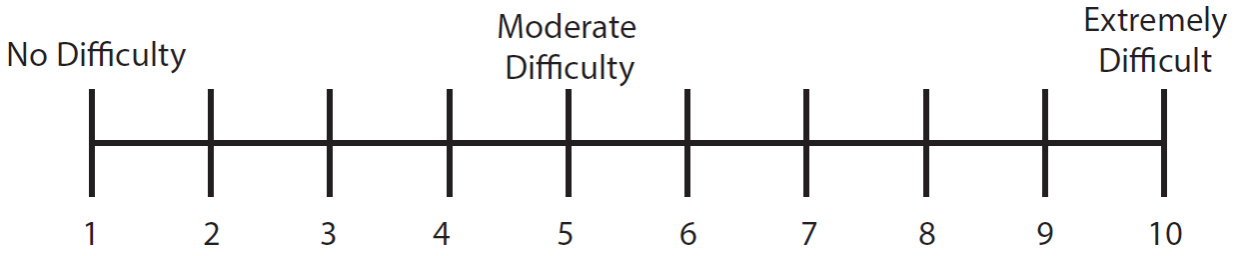

Figure 3.4 Visual Analog Scale

3.6 Data Analysis

Individual marker data was imported into Visual 3D (C-Motion, Germantown, MD, USA) to create an eight segment, link-segment model (see Table 3.1). Raw marker data was interpolated (maximum gap 10 frames) and filtered at 6 Hz with a low-pass Butterworth filter.

For all walking trials, the gait events defining the stance phase of gait (heel contact, toe off) were identified using visual inspection for steady state gait only. Measures of interest (described below) were calculated using Visual 3D (C-Motion, Inc.) for the steady state step which occurred in the middle of the platform.

Table 3.1 Segments Included in the Full Body COM Model, Segment Endpoints and Tracking Markers

Segment	Distal Endpoint	Proximal Endpoint	Tracking Markers
Trunk	Right Iliac Crest, Left Iliac Crest	Right Acromion, Left Acromion	Cluster of three markers attached to upper back
Pelvis	Right PSIS, Left PSIS	Right ASIS, Left ASIS	Cluster of three markers attached to pelvis
Right Thigh	Right Lateral Knee, Right Medial Knee		Cluster of three markers attached to right thigh
Left Thigh	Left Lateral Knee, Left Medial Knee		Cluster of four markers attached to left thigh
Right Shank	Right Lateral Ankle, Right Medial Ankle	Right Lateral Knee, Right Medial Knee	Cluster of three markers attached to right shank
Left Shank	Left Lateral Ankle, Left Medial Ankle	Left Lateral Knee, Left Medial Knee	Cluster of four markers attached to left shank
Right Foot	Right Lateral Toe, Right Medial Toe	Right Lateral Ankle, Right Medial Ankle	Cluster of three markers attached to right foot
Left Foot	Right Lateral Toe, Right Medial Toe	Left Lateral Ankle, Left Medial Ankle	Cluster of four markers attached to left foot

3.7 Outcome Measures

Temporal-spatial measures included: gait speed, stride width, and step length. To characterize balance, margin of stability was considered the primary measures of interest.

Margin of stability (MOS) refers to the distance between the boundary of the base of support and the extrapolated center of mass (XCOM) [40]. During standing, human balance can be modelled using an inverted pendulum model [80, 24], The XCOM builds on the concept of the inverted pendulum and, through adding the term $\frac{\mathrm{vCOM}}{\sqrt{\mathrm{gl}^{-1}}}$ [81] (see Equation 3.1), allows for quantification of stability in dynamic situations [21]. A positive MOS indicates postural stability, since the XCOM is within the base of support [82]. A negative MOS indicates postural instability, as the XCOM has exceeded the base of support [82].

Margin of stability was chosen for multiple reasons. Margin of stability takes into account the velocity of the COM allowing for evaluation of balance control in dynamic situations [21]. It has been used previously to evaluate stability during sloped gait [7] [44], allowing for comparison to previous studies.

The XCOM calculation is defined below (Equation 3.1).

$$\mathrm{XCOM} = \mathrm{pCOM} + \frac{\mathrm{vCOM}}{\sqrt{\mathrm{gl}^{-1}}}$$

Equation 1 Where, XCOM = the extrapolated center of mass, pCOM = vertical projection of the COM to the ground**, g = acceleration due to gravity, l =the distance between the COM and the ankle joint** [82]

The velocity of the COM was calculated as the first derivative of the position of the COM in the sagittal and frontal planes. L was calculated as the distance between the COM and the ankle joint in accordance with the inverted pendulum model. This approach was used by Bosse et al.

For this study, the minimum margin of stability in the sagittal and frontal plane during weight acceptance and single limb support was determined. Minimum margin of stability was chosen to

indicate when people were the least stable. Weight acceptance and single limb support were separately considered, as both have distinct challenges. During weight acceptance, at heel contact, the body loads the stance limb in an insecure posture [25], while there is an increased risk of falls due to slips during descent specifically given the increased shear forces [6]. Single limb support involves the body progressing over a single limb [22]. The associated challenge of single limb support is stability as the body progresses over a single limb [25, 22]. Of note, since the base of support changes throughout the gait cycle, the markers defining the base of support also varied (See Figure 3.5). The anterior border of the BOS was defined by the first metatarsal head marker of the leading foot, and the first metatarsal head marker of the stance limb during single limb support. The lateral border of the BOS was defined by the fifth metatarsal head of the trailing leg during double limb support, and the fifth metatarsal head of the stance limb during single limb support.

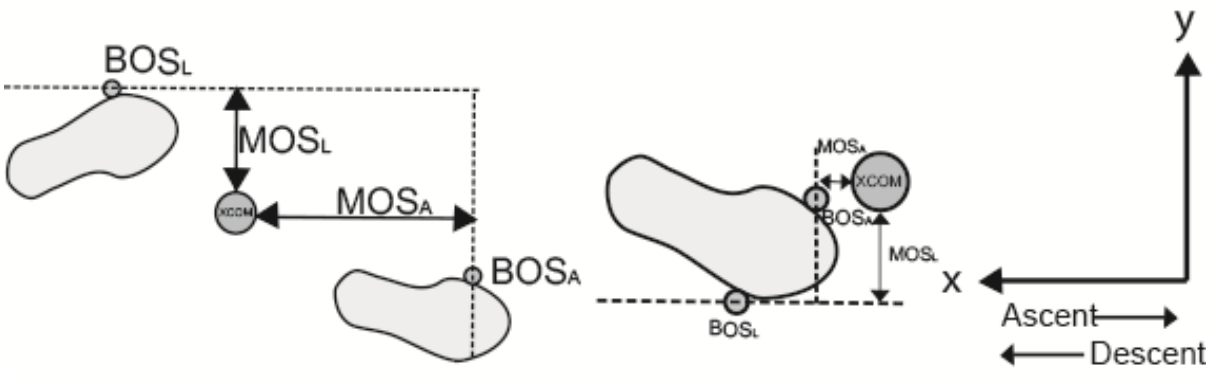

Figure 3.5 Boundaries of the base of support during weight acceptance (left) and single limb support (right). The grey circles intersecting with the dotted line represent the markers used for the base of support boundary. The larger grey circle in the figure represents the extrapolated centre of mass. Margin of stability was calculated as $MOS_A = -(BOS_{Ax}-XCOM_x)$ for ascent and $MOS_A = (BOS_{Ax}-XCOM_x)$ for descent. Lateral margin of stability was caluclated as $MOS_L = (BOS_{Lx}-XCOM_x)$ for ascent and $MOS_A = -(BOS_{Lx}-XCOM_x)$ for descent. Negative values indicate that the XCOM has exceeded the BOS. Adapted from [40]

3.8 Statistical Analyses

Data was extracted from individual trials then averaged, with inclined and declined walking separately analyzed. For all variables of interest, descriptive statistics, including means and standard deviations were determined.

Since only four AFO users participated in this study, statistical analyses were limited to the healthy older adult group only; the AFO group were descriptively compared to the healthy group as appropriate to address Objective 2. To address Objective 1, for the healthy older adults only, repeated measures ANOVAs with one within subject factor (walkway angle) was performed for all outcome measures. When a significant main effect was determined, post hoc analyses were performed, where pairwise comparisons with a Bonferroni correction were performed. Significance was set at $p<.05$ for all analyses.

Chapter 4
Results

4 Results

4.1 Participant Demographics

Fourteen healthy older adults (60+ years old) and four AFO wearers (18+ years old) participated. Two healthy older adults were excluded due to equipment malfunction. One participant in the AFO group did not feel comfortable completing the protocol without AFOs and only participated for the AFO condition. Given the small sample size for the AFO group, only qualitative comparisons of the AFO users and healthy older adults were performed.

Participant characteristics are presented in Tables 4.1 to 4.3. All participants, including the AFO group, were over the age of 60. In general, AFO group demonstrated slower Timed Up and Go (TUG) scores when qualitatively compared to the healthy older adults (Table 4.1). Considering the AFO group specifically, all were bilateral AFO wearers; three wore AFOs due to Charcot Marie Tooth, and one due to drop foot. All wore energy storage and return (ESR) AFOs (Table 4.2.

Table 4.1 Mean participant demographics and standard deviations () for healthy older adults and AFO wearers.

Demographics	Older Adults (n=12)	AFO Group (n=4)
Age (years)	65.5 (4.6)	68.0 (7.9)
Sex (M/F)	4/8	3/1
Height	169 (8.9)	168 (4.8)
Weight (kg)	69.7 (11.9)	71.9 (15.6)
TUG score (seconds)	9.25 (1.89)	11.89 (3.14)

*Note the TUG test scores were obtained for AFO users while wearing their AFO

Table 4.2 Detailed participant characteristics for the AFO group only

Participant Number	Unilateral or Bilateral AFO use	AFO Type	Condition that lead to AFO use	Hours/Day of AFO use	Years of AFO use	Time (months) after the development of condition AFO received
A01	Bilateral	ESR	Charcot Marie Tooth	>12	>20	>12
A02	Bilateral	ESR	Charcot Marie Tooth	8-10	10-20	>12
A03	Bilateral	ESR	Drop foot	>5	5-10	>12
A04	Bilateral	ESR	Charcot Marie Tooth	8-10	5-10	>12

When asked about common ramp walking preference and perceived difficulty (Table 4.3), the AFO wearers generally reported "sometimes" having difficulty walking both up and down, avoiding slopes, or falling on slopes, whereas the healthy older adult group reported "rarely" or "never" when responding to the same questions.

Table 4.3 Total responses for healthy older adults and AFO wearers related to ramp walking perceived difficulty and preferences

Survey Question	Healthy Older Adults	AFO Wearers
Difficulty walking down ramps/slopes?	Always = 0 Most of the time = 0 Sometimes = 0 Rarely = 6 Never = 6	Always = 0 Most of the time = 0 Sometimes = 3 Rarely = 0 Never = 1
Difficulty walking up ramps/slopes?	Always = 0 Most of the time = 0 Sometimes = 0 Rarely = 5 Never = 7	Always = 0 Most of the time = 0 Sometimes = 3 Rarely = 0 Never = 1
Preference?	Ascent = 2 Descent = 0 Both, No Preference = 10 Neither, Avoids = 0	Ascent = 0 Descent = 2 Both, No Preference = 1 Neither, Avoids = 1
Fallen While Walking on Slopes? (Yes/No)	(0/12)	(1/3)

4.2 Spatiotemporal Features of Gait

4.2.1 Gait Speed

As illustrated in Figure 4.1 for healthy older adults, no significant main effect of slope angle was found across all walkway angles during descent (p=.356). During ascent a significant main effect was found (p=.005). The AFO group (both with and without AFO use) generally ascended and descended all slopes with a slower walking speed when qualitatively compared to the healthy older adults. Across all slope angles, participants in the AFO group walked approximately 31% faster with their AFOs compared to without, with the largest differences between with AFO and no AFO condition being during level ground walking (58%). Comparing groups, the largest differences between AFO wearers (with AFO) and healthy older adults were at the steepest angle (approximately 17% for descent and 22% for ascent).

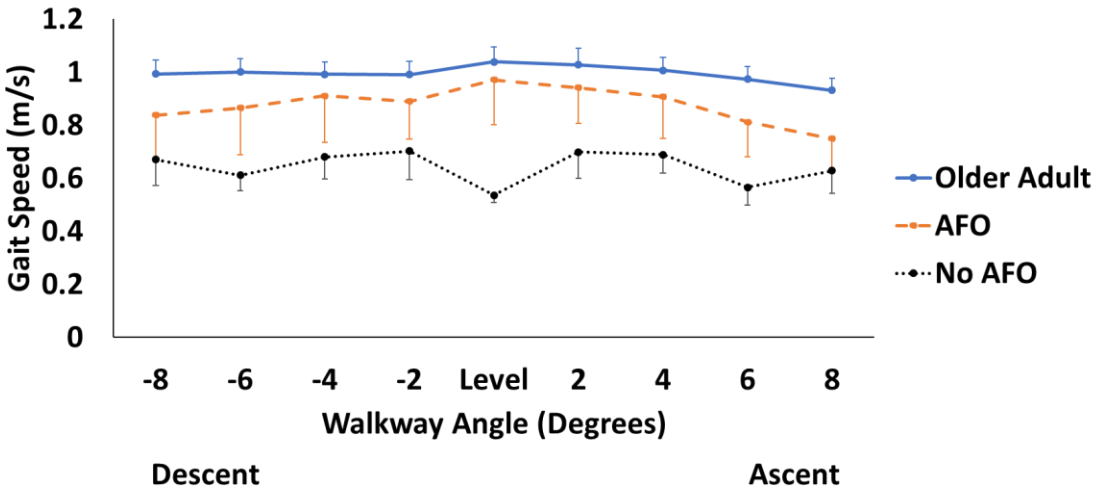

Figure 4.1 Gait speed (m/s) for healthy older adults (solid) and AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. Error bars represent standard error.

4.2.2 Step Width

As illustrated in Figure 4.2, step width did not differ across slope angles during descent (p=.113). During ascent a significant main effect was found (p=.002) where during ascent the older adults took significantly narrower steps at the steepest angle (8 degrees) compared to level ground (p=.008). In general, the AFO wearers took wider steps (approximately 38% wider when wearing

their AFO, and approximately 40% wider without the AFO) when qualitatively compared to healthy older adults across all slopes. During ascent, the largest differences between both the AFO condition and no AFO condition compared to the healthy older adults were at the steepest angle.

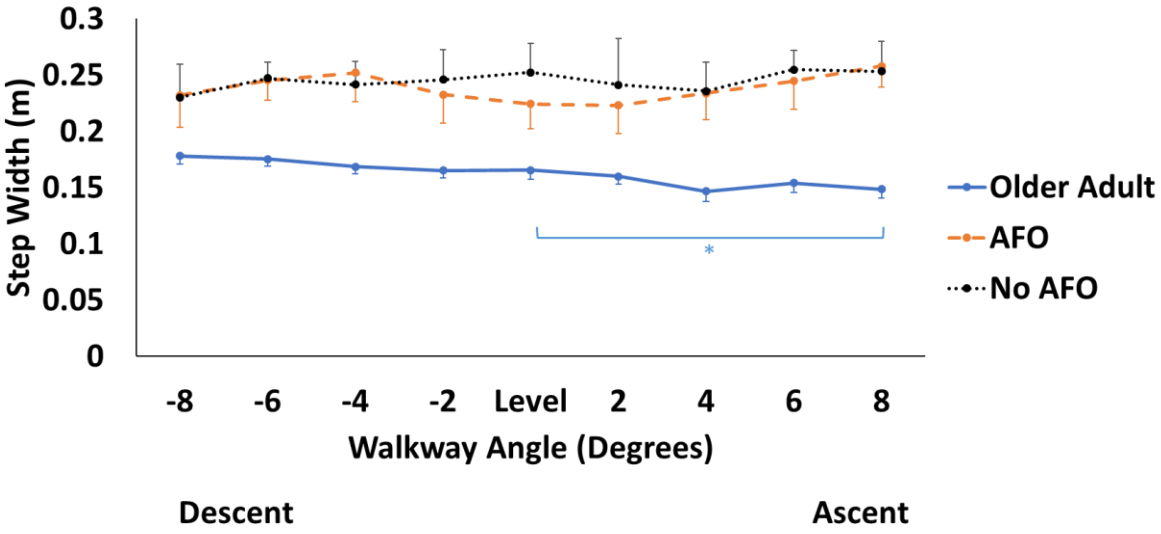

Figure 4.2 Step width (m) for healthy older adults (solid) and AFO wearers with (dashed) and without (dotted) their AFOs while walking up and down slopes. *Indicates significant effect of slope angle for the older adult group only. Error bars represent standard error.

4.2.3 Step Length

The mean step length across slopes angles is presented in Figure 4.3. A significant main effect of slope angle was found during descent ($p<.001$), but not during ascent ($p=.114$) Healthy older adults took significantly shorter steps when descending 4, 6, and 8 degrees, compared to 0 degrees ($p<.035$); shorter steps were also detected when descending 6 and 8 degree slopes compared to 2 degree slopes ($p<.035$). Overall, AFO wearers had shorter steps when qualitatively compared to healthy older adults. Although they performed similarly during descent, level ground, and 2 degrees ascending, while ascending larger slope angles (4 to 8 degrees), participants in the AFO group had slightly longer step lengths without AFOs compared to when they were wearing their AFOs.

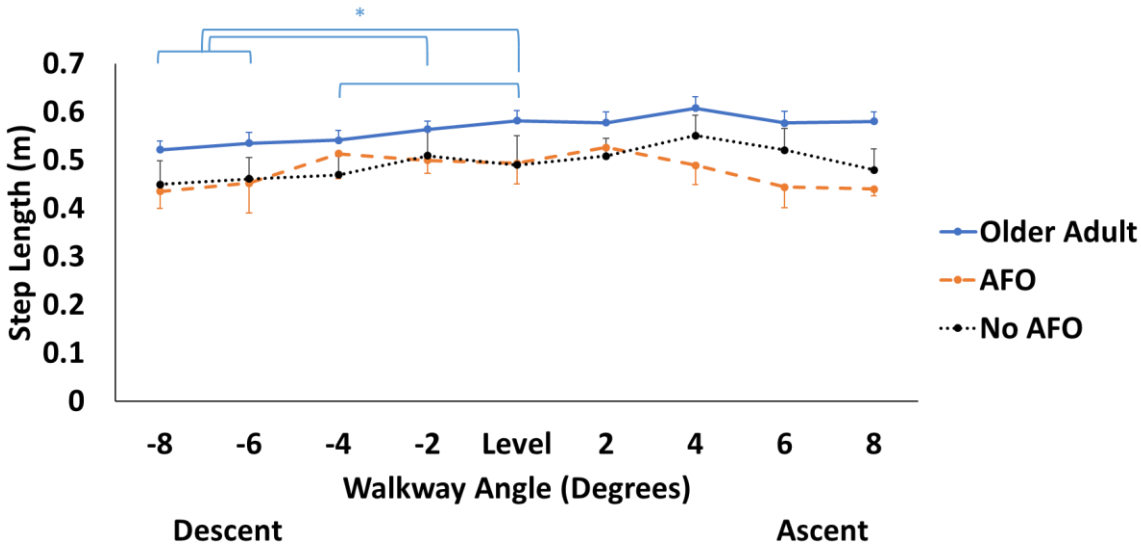

Figure 4.3 Step length (m) for healthy older adults (solid) and AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. *Indicates significant effect of slope angle for the older adult group only. Error bars represent standard error.

4.3 Measures of Balance Control

4.3.1 Frontal Plane Margin of Stability During Weight Acceptance and Single Limb Support

As shown in Figure 4.4, no significant differences in lateral MOS across angles was found for healthy older adults during either weight acceptance (descent: p=.058, ascent: p=.826) (Figure 4.4A) or single limb support (descent: p=.795, ascent: p=.647)(Figure 4.4B). When qualitatively compared to healthy older adults, the AFO wearers generally presented with larger lateral margins of stability during both weight acceptance and single limb support. Although AFO wearers had faster lateral COM speed compared to healthy older adults during weight acceptance (see Appendix C), the observed difference was likely due to the AFO wearers maintaining their COM further from the lateral border of the BOS (see Appendix B) resulting in the larger lateral MOS.

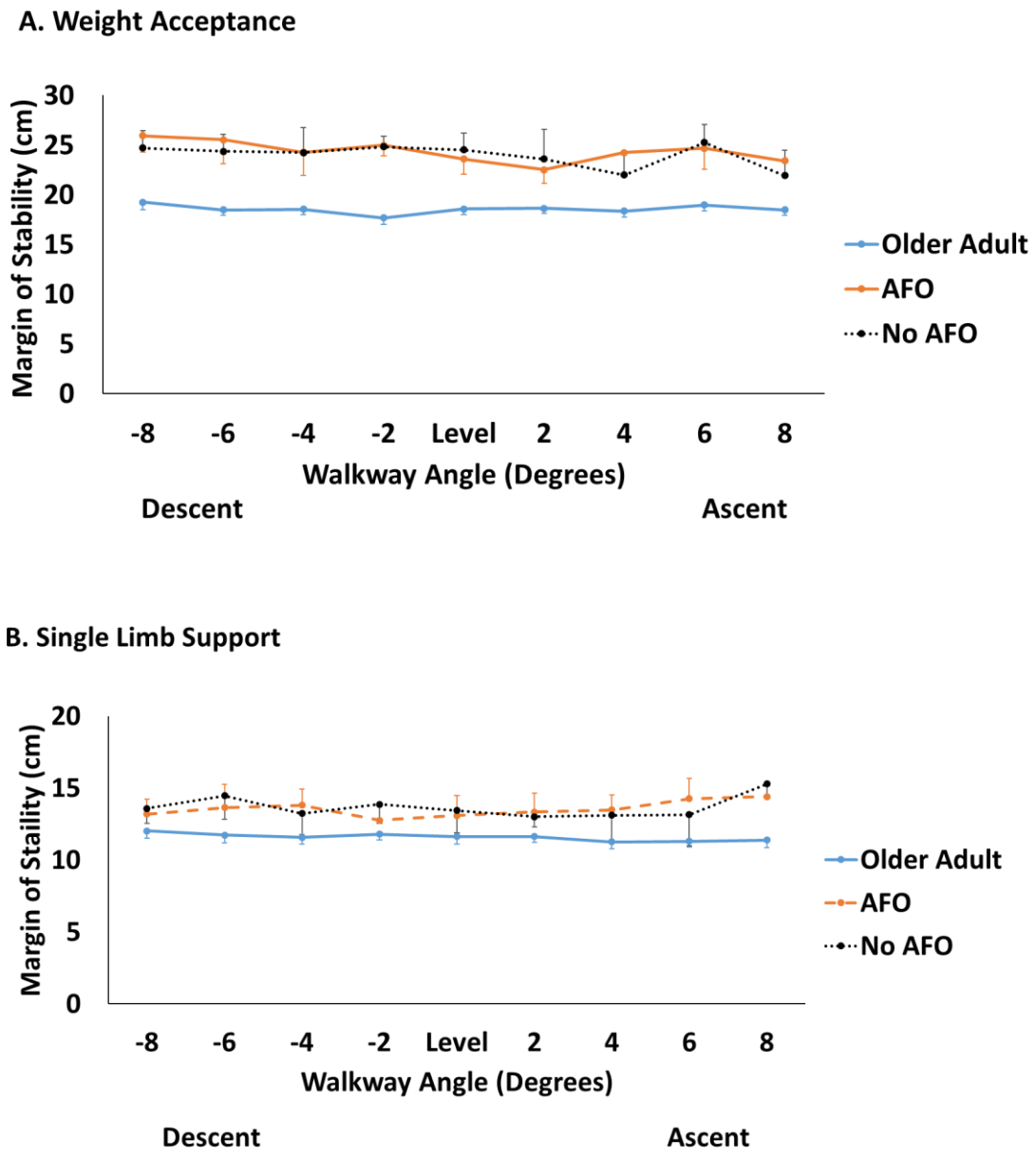

Figure 4.4 Minimum lateral margin of stability (cm) during weight acceptance (A, TOP) and single limb support (B, BOTTOM) for healthy older adults (solid) and AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. Error bars represent standard error.

4.3.2 Sagittal Plane Margin of Stability During Weight Acceptance

The sagittal plane margin of stability during weight acceptance is presented in Figure 4.5. For all groups, the minimum anterior margin of stability decreased with decreasing angles during descent ($p<0.001$); the anterior MOS decreased further as participants ascended the increasing walkway angles resulting in the XCOM exceeding the BOS during sloped walking uphill ($p<0.001$). When qualitatively compared to the healthy older adults AFO wearers had a larger anterior MOS during descent and level ground walking, but performed similarly to the older adults during ascent – this was seen because despite that fact that AFO wearers had slower COM velocity, they also maintained less space between their COM and the anterior border of the BOS during ascent (see Appendix B), resulting in similar peak MOS compared to the healthy older adults. In contrast, when not wearing their AFO's, the anterior MOS was generally greater across all angles compared to the other groups, likely due to a slower COM velocity (see Appendix C).

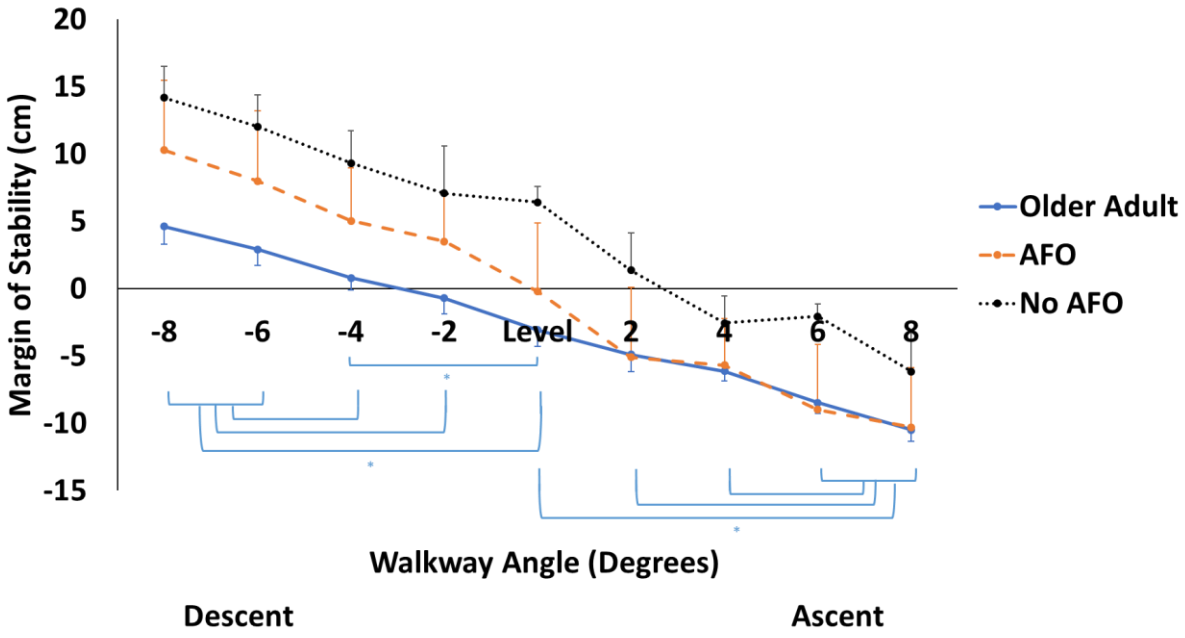

Figure 4.5 Minimum anterior margin of stability (cm) during weight acceptance for healthy older adults (solid) and AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. *Indicates significant effect of slope angle for the older adult group only. Error bars represent standard error.

4.3.3 Sagittal Plane Margin of Stability During Single Limb Support

As illustrated in Figure 4.6, during single leg support the XCOM exceeded the BOS for all angles resulting in a negative margin of stability, which became more negative with increasing slope angle (descent: $p<0.001$, ascent: $p=0.001$). During descent, the MOS was significantly more negative at 4, 6, and 8 degrees compared to level ground ($p\leq.005$) , and as well as between 8 and 2 degrees ($p\leq.028$). During ascent, there was a significant difference between level ground and 8 degrees only ($p=.007$). The negative MOS during single limb support was the result of the COM exceeding the anterior border of the BOS for all angles for healthy older adults, with greater difference between the COM and BOS with decreasing angles during descent and increasing angles during ascent (See Appendix B).

AFO wearers followed a similar trend during single limb support, but with less negative margin of stability both with and without their devices. Overall, these observed differences were seen as the AFO group adopted a position of their COM which exceeded their BOS by less compared to healthy older adults. The anterior COM velocity was also reduced for the AFO group when qualitatively compared to the healthy older adults resulting in the generally less negative MOS (See Appendix C).

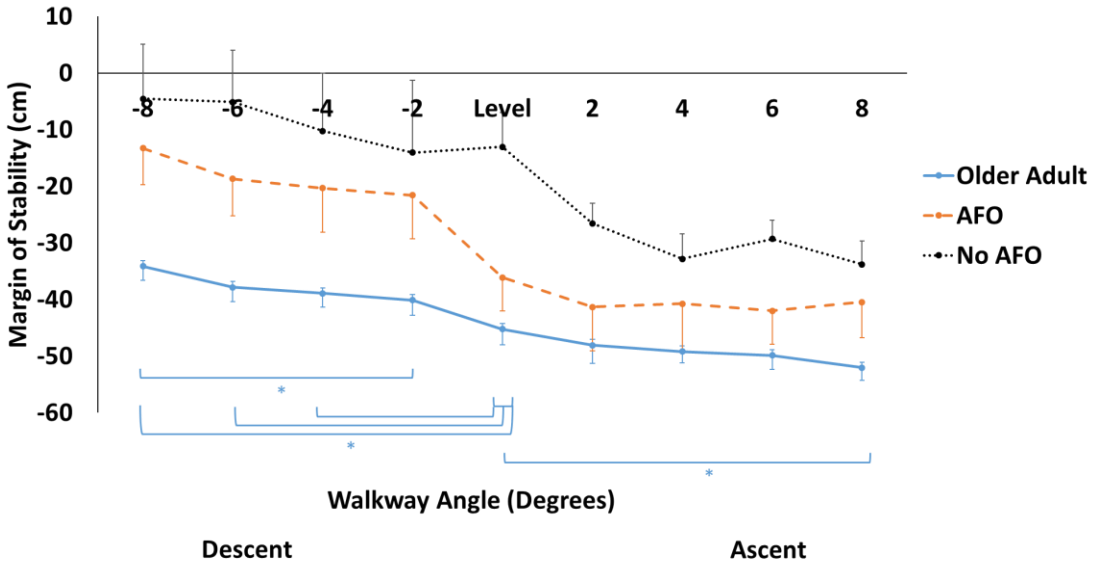

Figure 4.6 Minimum anterior margin of stability (cm) during single limb support for healthy older adults (solid), AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. *Indicates effect of slope angle for the older adult group only. Error bars represent standard error.

Chapter 5
Discussion

5 Discussion

Our built environment is designed such that inclined and declined surfaces must be negotiated – whether these sloped surfaces are man-made to address accessibility issues or occur naturally in the environment. As individual's age with or without impairment, the ability to navigate such sloped surfaces may be challenged [5, 36], although to date little information is available regarding how sloped walking affects balance control. This is the first study to investigate margin of stability in healthy older adults over a variety of angles during overground walking. This is also the first study to investigate slope ascent and descent in AFO wearers. The minimum anterior and lateral margin of stability during both weight acceptance and single limb support was determined. The main findings showed that older adults shifted their COM more anteriorly with steeper angles during ascent and more posteriorly with steeper angles during descent resulting in more instability at greater angles. While a more anterior COM during ascent could aid with forward momentum [5], it could increase the risk of a fall if a trip should occur [83]. During descent, moving the COM closer to the posterior BOS boundaries could result in a greater risk of backwards balance loss due to slipping [84]. AFO wearers had a more conservative gait pattern with a wider stance, also demonstrating less instability in the anterior direction both with and without their devices.

5.1 The Impact of Walkway Incline Angle on Margin of Stability in Older Adults

As a measure of dynamic balance control, the margin of stability was determined for both inclined and declined slope walking. In agreement with our hypothesis, healthy older adults navigated the slopes with a minimum anterior margin of stability that was greater during descent and less during ascent. This meant that individuals were maintaining a more upright posture (resulting in a positive margin of stability at weight acceptance, and a "lesser" negative margin of stability through single limb support) during descent. As the slope angle increased from the largest negative to level ground, and then to positive slope angles, the anterior margin of stability generally decreased linearly – where the most "unstable" posture was seen with the largest angle during ascent. The differences we saw in margin of stability was due to changes in COM

positioning. During descent, external forces such as gravity and the mechanical effect of the slope move the body downward and forward, and as a result the trunk moving backward may be related to decreasing forward momentum [35]. However, during descent this adjustment could mean a greater risk of slipping as the COM is positioned closer to the posterior base of support boundaries [84]. Since the risk of falls due to slips is more likely to occur during descent [6] it was noted that the healthy older adults took significantly shorter steps during descent, in agreement with previous studies on sloped gait [38] [44]. Taking smaller steps has also been shown as a compensatory mechanism to aid in tighter control of angular momentum [38], an important component of fall prevention [38], [73]. In a similar manner, during ascent this more forward positioning of the COM could increase the risk of a fall should a trip occur [83]Positioning the body's COM more forward during ascent could also be related to preserving forward momentum as has been reported in recent literature [5, 35], although the cost may be a generally less stable posture as indicated by the margin of stability. Of importance, based on the findings from this study, it is clear that the slope angle of the built environment has important implications of fall risk for healthy older adults.

Very recently, one other study has investigated margin of stability during sloped walking with an older adult population, finding backward margin of stability greater for descent compared to level ground and ascent [7], contrary to our current results. There are several possible explanations for the differences between our results and those of previous studies. In contrast to the methodology we used to calculate MOS, Viera et al. (2017) defined backward margin of stability as the minimum position of the XCOM in relation to the heel marker over 150 steps, which would result in different values because the XCOM is being measured in relation to two different anatomical landmarks. ,Also, for our results, a larger anterior margin of stability would indicate greater stability in the anterior direction, whereas a larger value in the study by Viera et al. (2017) means that the XCOM has exceeded the heel marker by a greater amount, indicating greater instability in the anterior direction. Additionally, our study took place during overground walking rather than during treadmill walking as was the case for the past study [44]. Previous literature has shown that the position of the COM in relation to the BOS and velocity differs during treadmill gait compared to over-ground walking [87]. For example, Yang and King (2016) found that an individual's COM was more anterior and the COM velocity was slower during level ground treadmill gait compared to over-ground walking. While this resulted in

similar values for MOS for level ground walking, how these differences translate to balance during sloped gait is unknown. The posterior movement of the treadmill has also been shown to affect other measures of balance including rate of change of the COP-COM inclination angle (RCIA). In particular, during heel strike, the posterior motion of the treadmill resulted in a positive RCIA compared to a negative RCIA during overground walking [88], as well as greater ranges of the COP-COM inclination angle (IA) and increased average anterior RCIA during treadmill walking [88]. These differences in findings between our studies and the past literature highlight the importance of considering the task parameters (such as treadmill vs. over-ground walking) when interpreting the implications of the results.

In contrast to the results shown in the sagittal plane (and contrary to our hypothesis), in the frontal plane, the walkway angle did not affect minimum lateral MOS for both ascent and descent compared to level ground. This was consistent with previous literature on sloped gait and margin of stability on a treadmill [44]. It is possible that the challenge of the task was not enough and we may see more significant differences if steeper angles were included in the study. Although we did not directly compare ascent and descent, a previous study on margin of stability on slopes on a treadmill found that lateral margin of stability was less for descent compared to ascent [44]. The differences were attributed them to changes in foot height and faster pelvic oscillations [44].

5.2 AFO Wearers Exhibit More Conservative Gait During Sloped Walking Compared to Healthy Older Adults

Differences in spatiotemporal features have been shown to affect margin of stability during level ground walking in other populations [39], indicative of a more conservative gait strategy. During sloped walking, the current results demonstrated that AFO wearers had a wider stance and larger lateral margin of stability compared to healthy older adults. As shown in Appendix D (Figures D-1 and D-2), these compensations were likely adopted to accommodate the larger lateral COM range of displacement of the AFO group, and assist with controlling lateral stability. However, of importance, past research has shown step width as predictive of falling while walking, suggesting that wider may not necessarily translate to safer gait, but instead is independently associated with a fear of falling and predictive of future falls [31].

AFO users demonstrated less instability according to the margin of stability measure in the anterior directions across slope angles when compared to their healthy counterparts, also suggesting AFO wearers adopt a conservative gait strategy to adjust to task demands. However, contrary to our hypothesis, these group differences did not result in the AFO group demonstrating greater instability than older adults with the increasing slope angles. During descent, in fact, the AFO group appeared to adopt safer anterior margins of stability compared to the healthy group at the largest angles. Although we were limited in our ability to conduct statistical analyses of the AFO data due to the small sample size, it does appear that at the largest slope angles (during both ascent and descent) the AFO group walked with a larger difference in the spatiotemporal parameters compared to the healthy older adults and may provide an explanation for the lack of interactions that are qualitatively noted in the margin of stability results.

During weight acceptance, it was interesting to note that the AFO group (while wearing their AFOs) appeared to adopt a COM position relative to their BOS that was similar to the healthy older adults during descent, while during ascent, the COM was more forward compared to the healthy older adult group. It is possible that the orthosis is limiting ankle ROM in the sagittal plane, and increasing the forward trunk lean could be a compensatory action to aid with forward momentum when the ankle is unable to do so. It is not surprising that compensations would be made to increase forward momentum given the physical challenges associated specifically with inclined walking, including the greater concentric action of the muscles required to lift the body against gravity [36] and greater ankle range of motion at heel strike and the first 50% of the gait cycle [5]. This is also of particular interest and deserves further investigation given that the participants included in this study wore energy storage and return (ESR) AFO's, which permitted some ankle range that might not be permitted in other AFO designs. With other designs, for example a solid AFO which is designed to prevent deformation during loading [89], walking on slopes may be even more difficult as the range at the ankle is likely even more limited. If AFO wearers are forced to lean forward to a greater degree to compensate for decreased ankle range of motion, they could be at a greater risk of falling should a trip occur [83].

5.3 AFO Use and the General Challenge of Sloped Walking

In general, the challenge of the slope walking task included in this study did not appear to substantially limit mobility for the AFO group. All but one of the participants were able to complete the testing protocol without incidence, although one AFO wearer did not complete the protocol without their AFOs as they did not feel comfortable walking in shoes without AFOs. Interestingly, the three participants that completed the entire protocol also reported no previous falls on slopes, whereas the participant who was unable to complete the protocol without the AFOs reported previously falling on slopes 3-4 times for both ascent and descent on the survey. For this particular participant with fall history, it is interesting to note that when investigating in detail their performance compared to the other three participants, the participant was similar to the other AFO wearers in terms of margin of stability. The participant did rate more angles to be more difficult and had higher perceived exertion for more angles however compared to the other AFO wearing participants, suggesting fear of falling and/or physical capacity as concerns to independent mobility on slopes.

As indicated by our biomechanical findings, and by the subjective ratings of task difficulty, the slope walking task included in this study was rated as more challenging, and with higher perceived exertion, without AFOs use compared to with the AFO (See Appendix E). In the latter case, most AFO wearers had only little to moderate difficulty with the slope walking task. Although AFOs are generally designed with level ground in mind, energy storage and return AFOs appear to be appropriate for sloped gait. However, it is possible that inclusion of different AFO user types, or slope angles could result in different findings. For example, we only included one AFO type – an energy storage and return AFO. Previous literature on AFOs and slopes showed that AFO type, specifically solid AFOs significantly affected spatiotemporal features of gait such as affected and unaffected step length and gait speed [12], which can affect margin of stability [39]. Solid AFOs deliver a high level of control to the ankle and foot, restricting dorsiflexion during midstance and plantarflexion during terminal stance [8, 89]. A single axis AFO has hinges, allowing for ankle movement in the sagittal plane [8], and may not result in the same compensations. Given the range of motion and muscular demands of sloped gait at the ankle [5, 36], such restrictions due to AFO type may result in adjusted performance and greater difficulty navigating slopes. It is therefore is important to consider AFO design when considering the challenges of sloped gait. Additionally, given that AFO design depends on many careful

considerations including patient and clinician goals, physical factors and comorbidities [8], it is also important to consider factors such as patient population. Although AFO use negatively impacted dynamic balance in healthy populations [74] [66], people that wear AFOs may wear them for reasons such as gait disorder [90], and spasticity [89], both of which can affect balance [91, 90, 92]. It is possible an AFO could aid balance in these populations through preventing an unwanted motion [68].

It is also interesting to note that when AFO wearers were asked to complete the task without their AFO's, no substantial differences in lateral margins of stability were noted between the two conditions. However, in the sagittal plane, the AFO wearers had a larger minimum anterior margin of stability when not wearing their AFOs compared to the "with-AFO" condition. As indicated by the results, the more "stable" margin of stability detected with the no-AFO condition was due to both a slowing of the COM velocity through weight acceptance and single limb support paired with less anterior positioning of their COM in relation to the BOS during single limb support . Despite these modifications to become more dynamically stable, this may not necessarily indicate less risk. In a study of individuals with multiple sclerosis higher anterior-posterior margin of stability at heel strike was positively correlated with number of falls and Expanded Disability Severity Scale [93] – the same could apply here. It is possible that although AFO wearers are more cautious without their devices, perhaps due to lack of comfort completing the task under the more novel conditions, the concomitant translation to risk is not necessarily present. Further research is required to understand the link between dynamic stability for AFO users during tasks of daily living and falls, as has been done with other populations [93].

5.4 Limitations

There are several limitations related to this present study that should be acknowledged:

- Four AFO participants were included in the current study. As a result of this low sample size, statistical analyses with AFO users were not possible limiting our ability to compare between groups and across slope angles.
- Only one type of AFO was tested. Previous literature on AFOs and slopes have shown AFO type significantly impacts functional balance tasks[60], and spatiotemporal features of gait [12], which in turn can affect the margin of stability [94]. Given the past literature highlighting the differences in functional performance between AFO types, it is possible

that different AFO types, particularly a solid AFO, could affect balance control differently during sloped walking.

- This study included only participants with lower motor neuron pathologies who were bilateral AFO wearers. Other neurological conditions such as upper motor neuron pathologies can involve spasticity [95], which can add additional challenges to balance control [96]. Further, while there is no literature comparing unilateral and bilateral AFO wearers during sloped gait, one study comparing unilateral, bilateral, and no AFO use in healthy children found that the compensations due to AFO use during stair ascent and descent increased with bilateral AFO use compared to unilateral AFO use [97], suggesting that inclusion of only bilateral use may limit generalizability of the findings
- A recent study has compared margin of stability during sloped gait to other inverted pendulum balance metrics. Compared to other models, such as Foot Placement Estimate (FPE), Margin of stability does not take angular momentum into account and as a result, it may underestimate stability in the sagittal plane [45].

5.5 Future Directions

The following future directions should be considered:

- As this present work was limited by issues with participant recruitment, future work should include a larger sample size to allow for statistical comparisons between groups and across slope angles.
- The present study only investigated a very specific group of AFO wearers with lower motor neuron pathologies. In reality, an individual may require an AFO for a number of a different reasons, such as stroke, cerebral palsy, and Charcot Marie tooth [54, 55, 57]. Since population and AFO type both affect measures of balance [98, 99, 100, 60], greater variety of pathologies, including both unilateral and bilateral wearers and AFO types, should be investigated.
- The current results highlight differences between AFO users and healthy older adults, where AFO users generally adopt more cautious gait to accommodate to the task demands. However, the relationship of these differences in biomechanical balance measures with actual fall risk remains unknown and should be investigated in the future.

5.6 Conclusion

To date, there is limited research on the effects of inclined walking on dynamic balance control, especially considering the use of assistive devices such as AFOs. This current study addressed a knowledge gap concerning the effect of walkway incline angle during overground walking. Results show that healthy older adults and AFO wearers both shifted their COM more posteriorly with steeper angles during descent and more anteriorly with steeper angles during ascent resulting in changes to anterior margin of stability. AFO wearers walked more cautiously, both with and without their devices, compared to their healthy counterparts to ensure adequate performance.

The results from this study provide important information on how ramp angle affects dynamic stability in both healthy older adults and AFO wearers with lower motor neuron pathology. The results may provide orthotists and other clinicians with information on how slopes in general, and orthosis use specifically, affects dynamic balance during sloped gait. The knowledge generated from this work is another step towards helping orthotists make more informed decisions when prescribing or designing an AFO. More generally, the results of this research also provide a basis for future work informing interventions and built environment design to support healthy and active aging.

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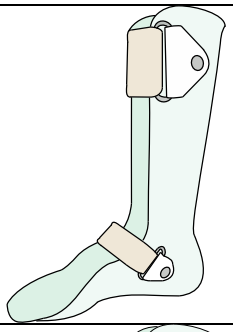
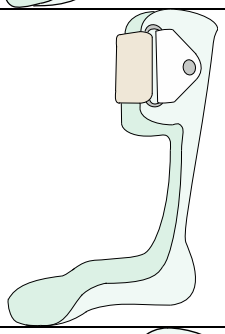
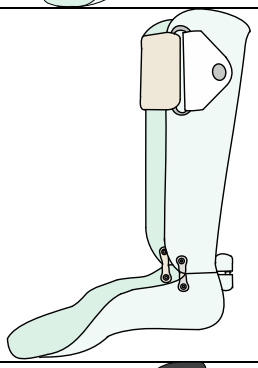
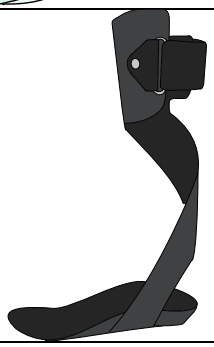
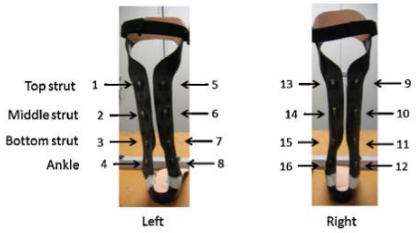
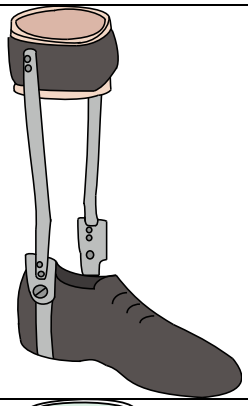
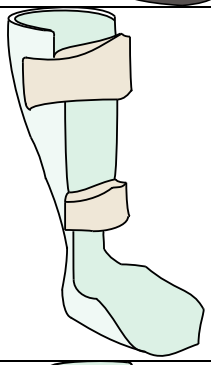
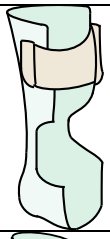
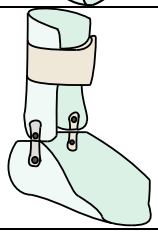
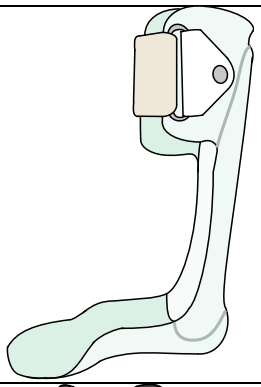
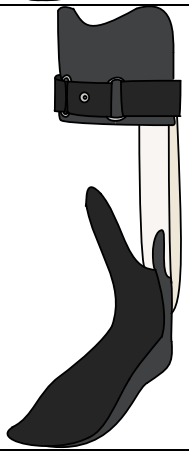
Appendices

Appendix A: AFO Types and Descriptions

Name	Figure	Description
Solid AFO		Covers the posterior portion of the leg with trim lines anterior to the malleoli and delivers a high level of control to the ankle-foot complex [8]
Posterior Leaf Spring (PLS) AFO		Covers posterior portion of leg with trim lines posterior to the malleoli which allow it to bend and recoil throughout the gait cycle [8]
Single Axis AFO		Covers the posterior portion of the leg, with hinges at the ankle to allow motion in the sagittal plane. Can be fabricated with a plantarflexion stop to block plantarflexion. [8]
Spiral AFO		Consists of a foot plate and tibial band connected by a spiral which encircles the leg. [8]The design allows the AFO to uncurl during weight acceptance, compress through mid stance and then uncoil as it is offloaded [8]

Carbon Fiber AFO [57]	 [57]	Custom made AFO fabricated from bi-directional carbon-Kelvar, bidirectional carbon, vacuum formed with epoxy resin [57]. Main structure of the brace is posterior to the leg and wraps anteriorly with a posterior strap [57].
Klenzak (Lehmann et al.)		Blucher shoe with rigid stirrup attached to the sole [65]. A calf band connected to two metal uprights attached to two adjustable Becker joints [65]. The joints were set at 5 degrees of dorsiflexion, 5 degrees of plantarflexion, and with a dorsiflexion spring assist [65].
AFO (Huang et al.)		Custom-made AFO with a proximal trim line distal to popliteal fossa, and distal trim line proximal to the metatarsal heads [75]
HFO-R (Huang et al.)		Custom-made AFO with a proximal trim line inferior to the muscle belly of the gastrocnemius, and the distal trim line at the distal end of the fat pad of the heel [75]
HFO-A (Huang et al.)		Custom-made AFO with a proximal trim line distal to the muscle belly of the gastrocnemius and the distal trim line proximal to the metatarsal heads [75]

Energy Storage and Return (ESR) AFO		The ESR AFO covers the posterior portion of the leg with trimlines posterior to the malleoli [101]. The ESR AFO consists of a copolymer base material and two overlapping polyproplylene reinforcements on the posterior and plantar surfaces [101]. The ESR AFO is named for its ability to store and return energy and is used for a number of pathologies [101].
Intrepid dynamic exoskeletal orthosis (IDEO)		Consists of a carbon fiber tibial cuff, deformable carbon fiber posterior strut, and carbon fiber supramalleolar foot plate with a foam wedge beneath the heel. [53] This AFO has energy storage and return properties, strut stiffness can be modified [53].
Standard Short AFOs	 [74]	Rigid orthosis that can block plantar and dorsiflexion [74]

Appendix B: Position of Center of Mass in Relation to Base of Support

The position of the COM relative to the BOS was determined at the instant of minimum margin of stability during both weight acceptance and single limb support. Given that COM position is one of the components of the margin of stability equation, COM position relative to the BOS was calculated to provide insight into differences found in margin of stability. A significant main effect was found during descent ($p<.014$), but not during ascent ($p=.652$). As illustrated in Figure B-1, during descent the COM was significantly further from the lateral border of the base of support (BOS) compared to 2 degrees ($p=.019$). Otherwise, there were no significant differences for the older adult group.

In general, when qualitatively compared to the older adult group, the AFO group's COMs were further from the lateral border of the BOS during weight acceptance (Figure B-1). As shown by Figure B-1, AFO use did not affect positioning of the COM relative to the lateral border of the BOS during weight acceptance.

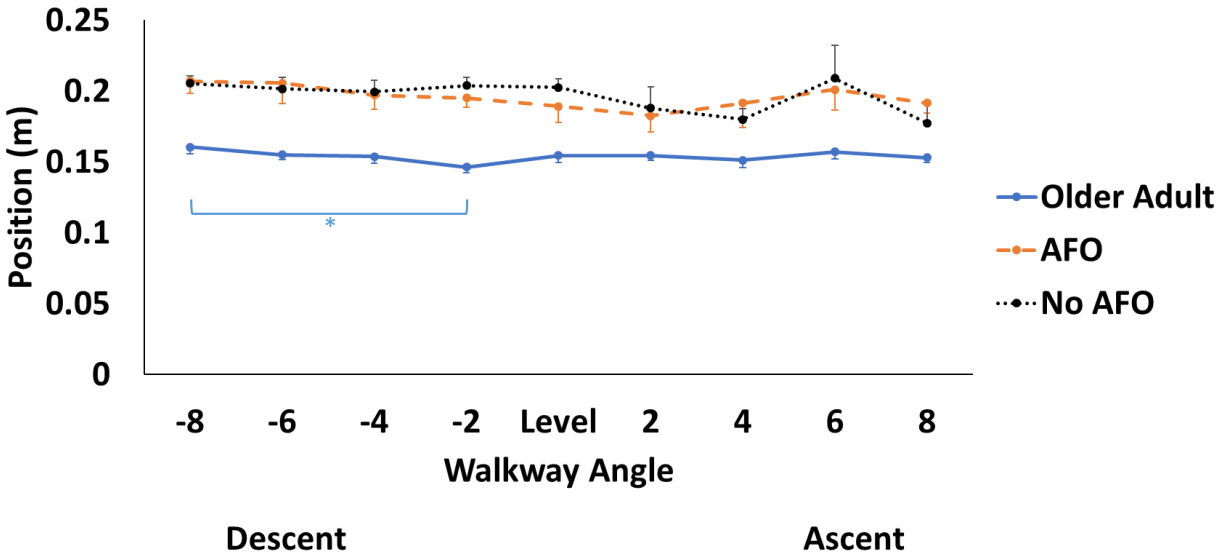


Figure B-1 Position of COM relative to lateral border of the base of support (BOS) during weight acceptance for healthy older adults (solid), AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. *Indicates effect of slope angle for the older adult group only. Error bars represent standard error.

During single limb support, there were no significant differences among any of the angles tested for COM position in relation to the BOS (Figure B-2). As illustrated by Figure B-2, similar to weight acceptance, the AFO group's COM was further from the lateral border when qualitatively compared to the older adult group, and AFO use did not affect positioning of the COM relative to the lateral border of the BOS.

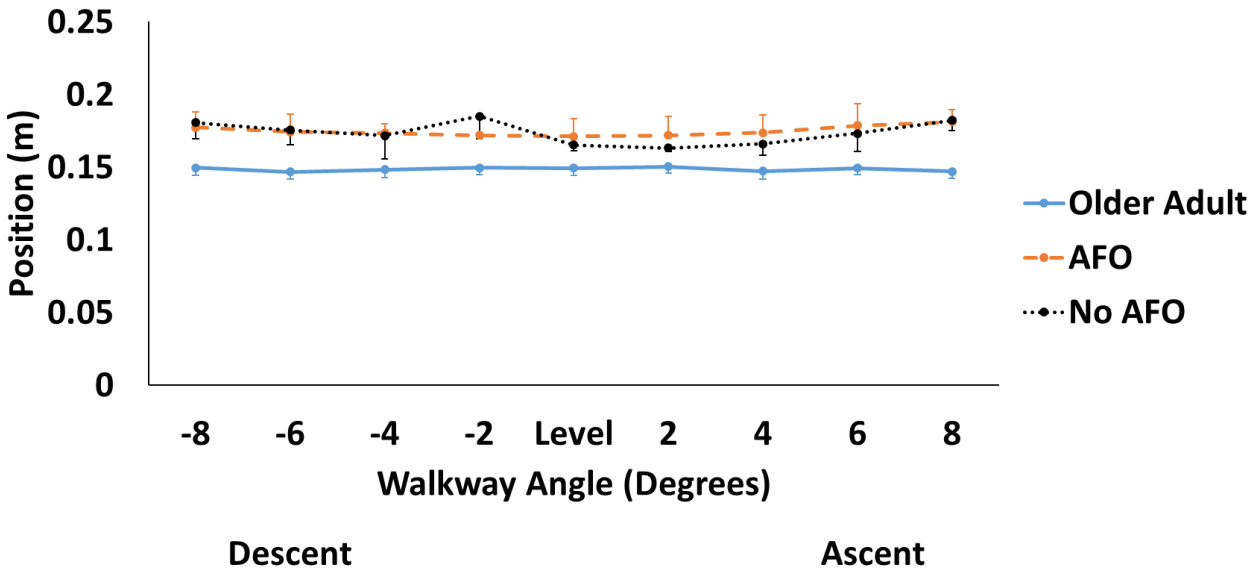


Figure B-2 Position of COM relative to lateral border of the base of support (BOS) during single limb support for healthy older adults (solid), AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. Error bars represent standard error.

As illustrated in Figure B-3, in the sagittal plane, during weight acceptance, the distance between the older adults COM and the anterior border of the BOS decreased in a fairly linear fashion with decreasing angle during descent ($p<0.001$), and further decreased further as angles increased during ascent ($p<0.001$).

The AFO group positioned their COM further from the anterior border of the BOS for all angles when walking without their AFO. With their AFOs the AFO group performed similarly to the healthy older adult group during descent. During ascent, they had less space between the COM and the anterior border of their BOS. Results are shown in Figure B-3.

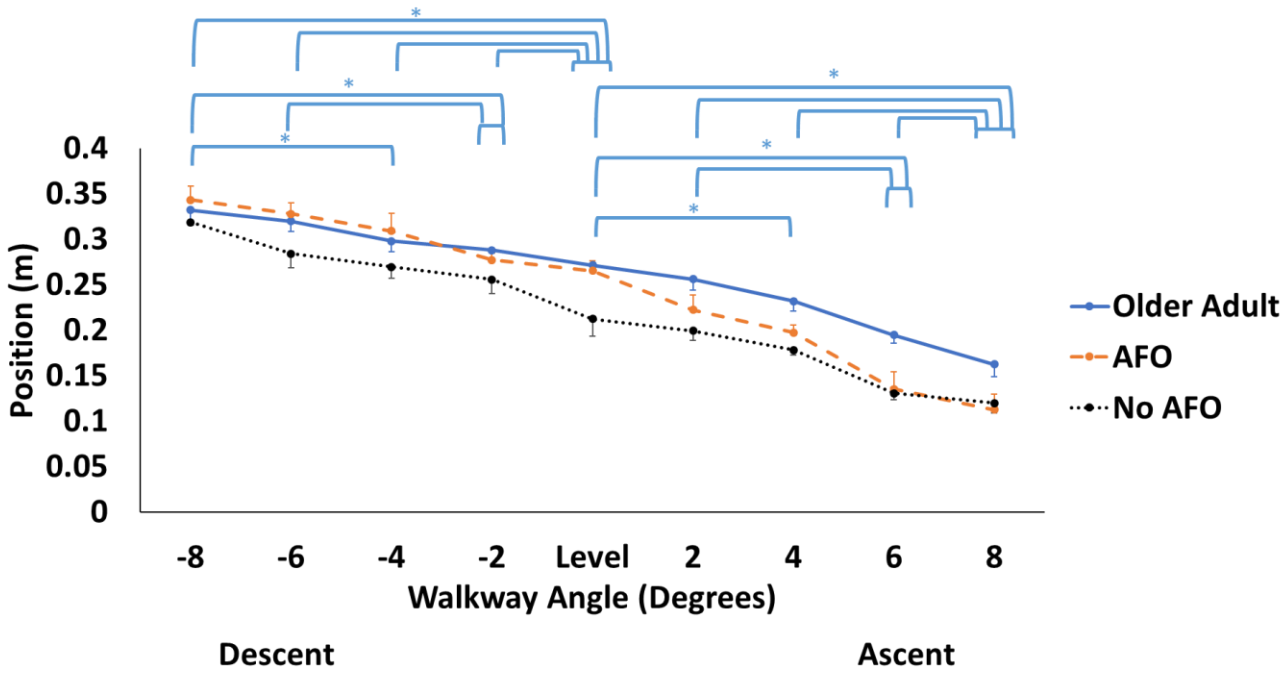


Figure B-3 Position of COM relative to anterior border of the base of support (BOS) during weight acceptance for healthy older adults (solid), AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. *Indicates effect of slope angle for the older adult group only. Error bars represent standard error.

As illustrated in Figure B-4, during single limb support, the COM exceeded the BOS for all angles for healthy older adults. The distance that the COM exceeded the BOS by increased with decreasing angle during descent ($p<0.001$), and increased with increasing angle during ascent ($p<0.001$).

AFO wearers with their AFOs exceeded their BOS by less when qualitatively compared to healthy older adults and AFO wearers without their devices exceeded the BOS by the least. Results are presented in Figure B-4.

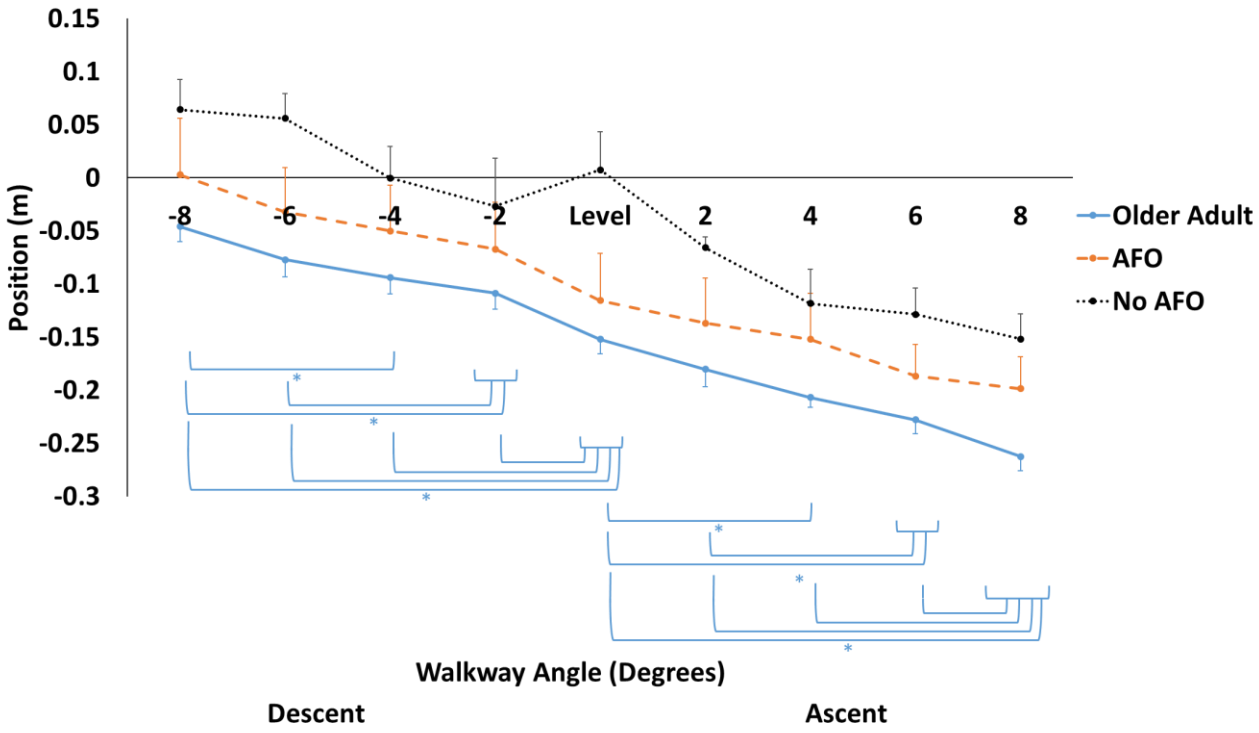


Figure B-4 Position of COM relative to anterior border of the base of support (BOS) during single limb support for healthy older adults (solid), AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. *Indicates effect of slope angle for the older adult group only Error bars represent standard error.

Appendix C: Center of Mass Velocity at Minimum Margin of Stability

COM velocity is a component of the margin of stability equation and was determined to further provide understanding of the differences found in margin of stability. The COM velocity was calculated as the first derivative of the COM position. As illustrated in Figure C-1, there were no significant differences across any of the angles tested for the healthy older adults (descent: p=.827, ascent: p=.967). AFO wearers had faster lateral velocity when qualitatively compared to healthy older adults (Figure C-1).

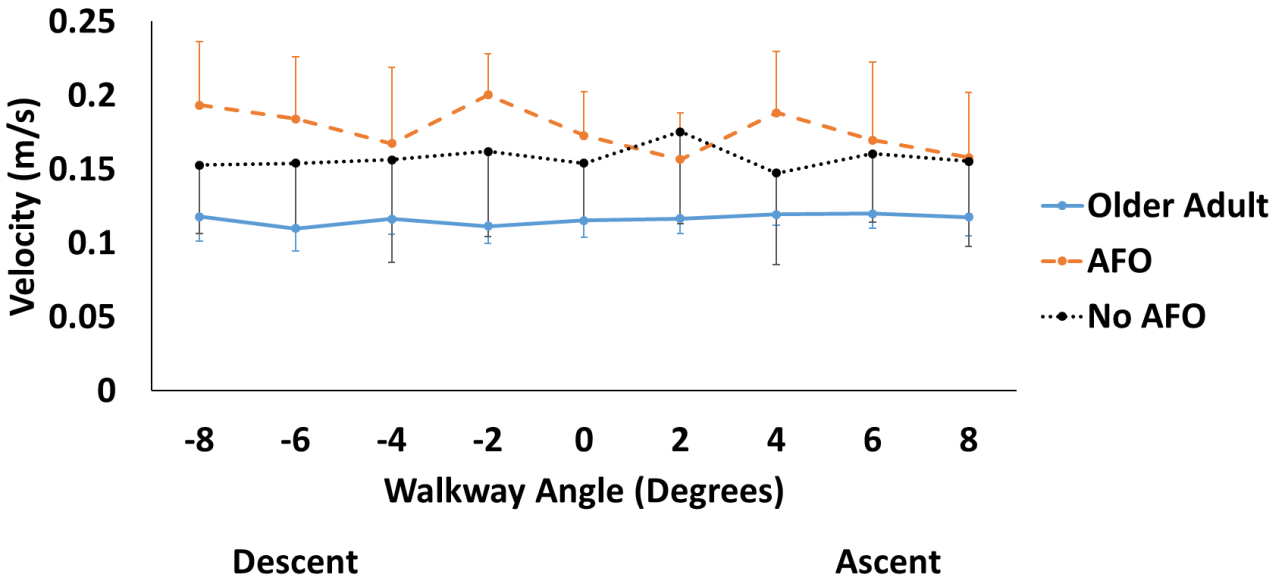


Figure C-1 COM lateral velocity during minimum margin of stability during weight acceptance for healthy older adults (solid), AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. Error bars represent standard error.

As shown by Figure C-2, there were no significant differences across any of the angles tested for the healthy older adults (descent: p=.391, ascent: p=.256). When qualitatively compared to the healthy older adults, the AFO condition had faster lateral velocity with their AFOs over most conditions, with the biggest differences compared to the older adult group at 8 degrees descending (approximately 57% faster) and the smallest at 2 ascending (approximately 23% faster). The AFO group had similar lateral velocity to the older adult group without their AFOs.

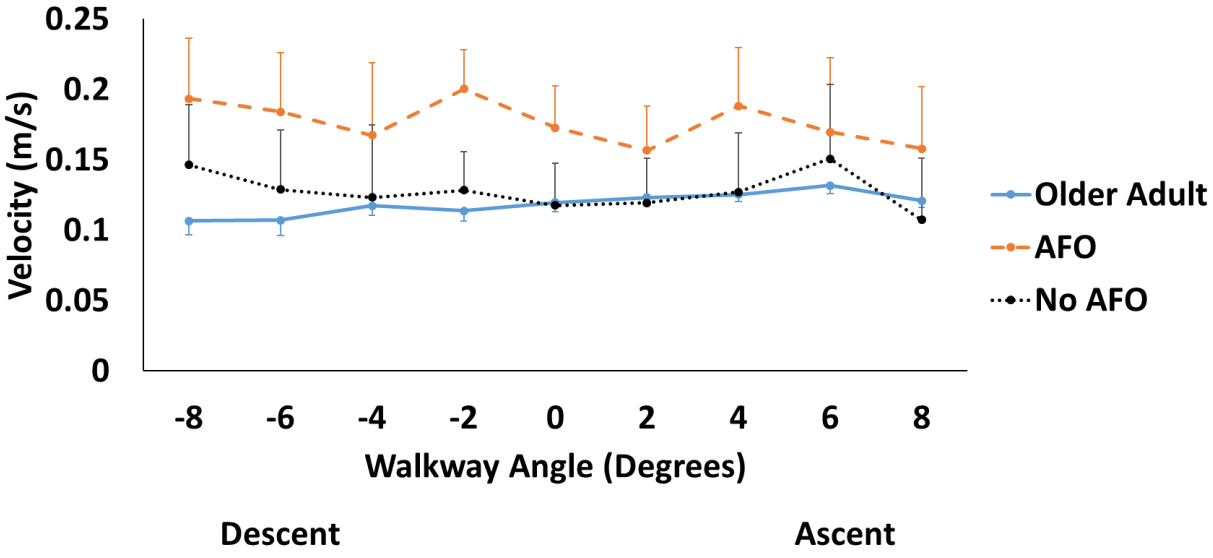


Figure C-2 Lateral COM velocity during minimum margin of stability during single limb support for healthy older adults (solid), AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. Error bars represent standard error.

The anterior velocity of the COM was calculated at the instant that margin of stability was at a minimum during weight acceptance and single limb support. A significant main effect of walkway angle was found during ascent (p=.003), but not during descent (p=.488). Compared to level ground walking, during weight acceptance, older adults' anterior COM velocity was significantly slower during ascent at six and eight degrees ($p \leq .041$). Results are presented in Figure C-3. AFO wearers were qualitatively slower than the healthy older adults, with the slowest condition being the AFO group without their devices (Figure C-3).

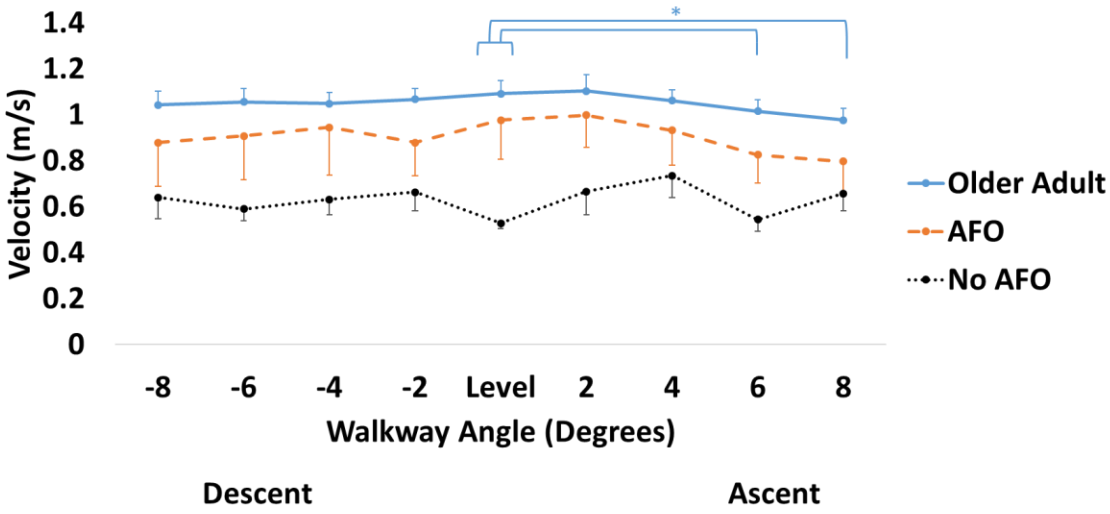


Figure C-3 Anterior COM velocity during minimum margin of stability during weight acceptance for healthy older adults (solid), AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. *Indicates effect of slope angle for the older adult group only. Error bars represent standard error.

A significant main effect of walkway angle was found for ascent ($p<0.001$), but not descent ($p=.724$). As illustrated in Figure C-4, during single limb support, 8 degrees was significantly slower than level ground and 2 degrees ($p\leq.040$), and 6 degrees was significantly slower than level ground ($p=.044$). As shown in figure C-4, similar to weight acceptance, AFO wearers were qualitatively slower than the healthy older adults with the No AFO condition being the slowest.

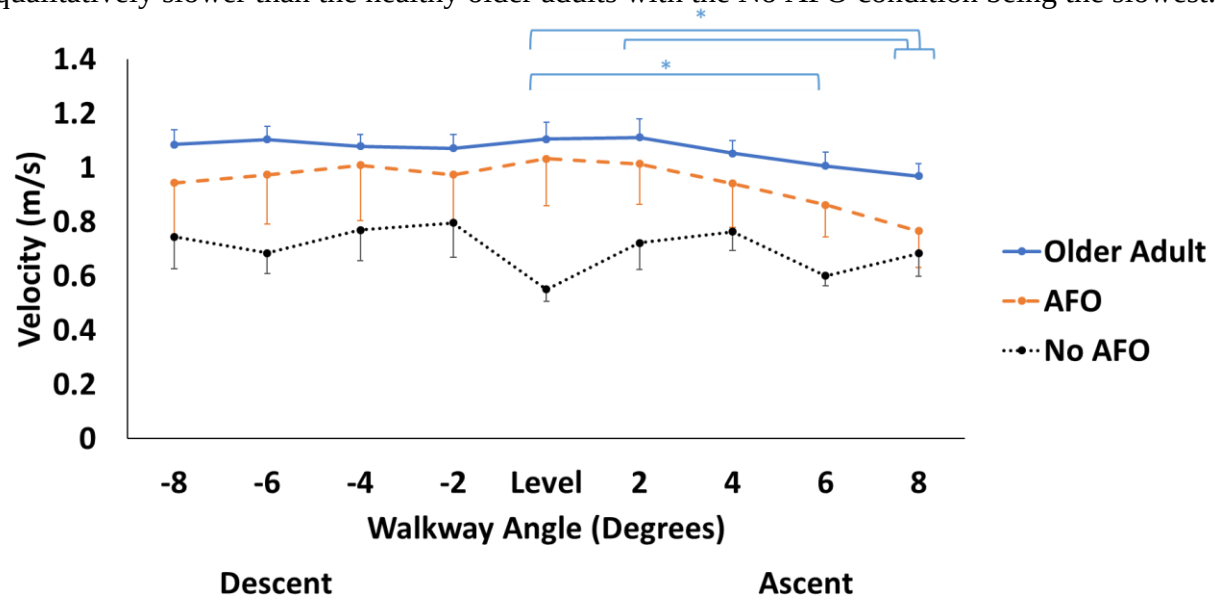


Figure C-4 Anterior COM velocity during minimum margin of stability during single limb support for healthy older adults (solid), AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. *Indicates effect of slope angle for the older adult group only. Error bars represent standard error.

Appendix D: Center of Mass Range of Displacement

Center of mass range of displacement was calculated in the frontal plane as the difference between the minimum and maximum values during weight acceptance and single limb support. During weight acceptance, a significant main effect of slope angle was found during descent (p=.029), but not during ascent (p=.803). During single limb support no significant differences were found during descent (p=.247). However during ascent, a significant main effect was found (p=0.002). Older adults took COM range of displacement was larger at 8 degrees compared to 2 degrees (p=.019) During weight acceptance and single limb support, AFO wearers had a greater COM range of displacement across all walkway angles when qualitatively compared to the healthy older adults (See Figures D-1 and D-2). Older adults had a significantly greater COM range of displacement at 8 degrees ascending compared to 2 degrees ascending (p=.006) during single limb support (See Figure D-2).

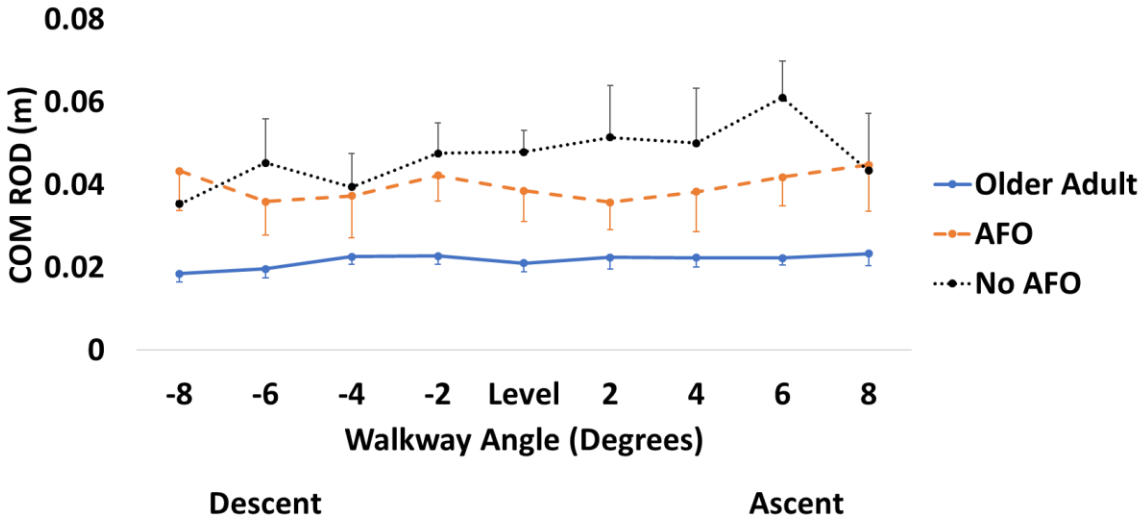


Figure D-1 COM frontal plane range of displacement (ROD) during weight acceptance for healthy older adults (solid), AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. Error bars represent standard error.

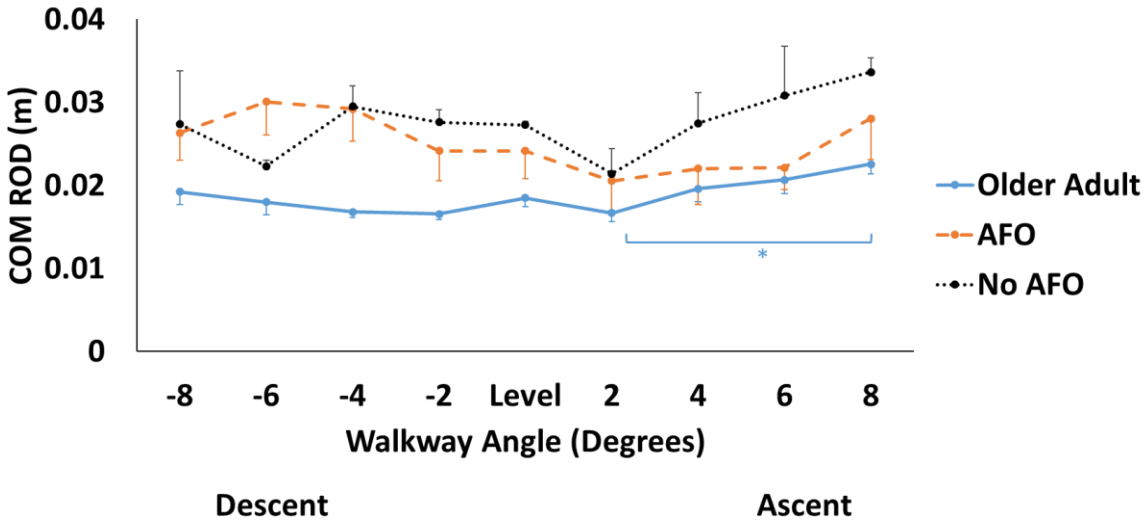


Figure D-2 COM frontal plane range of displacement (ROD) during single limb support for healthy older adults (solid), AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. *Indicates effect of slope angle for the older adult group only Error bars represent standard error.

Appendix E: Perceived Exertion and Difficulty

Participants were asked about the perceived difficulty of the task and perceived exertion using a visual analog scale (See Figure 3.4) after each slope condition.

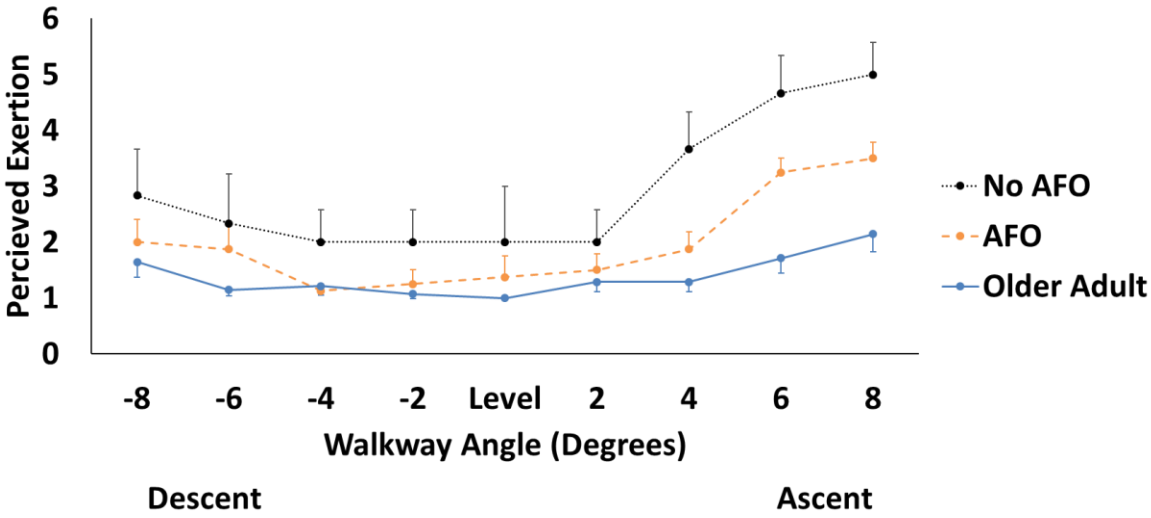


Figure D-1 Perceived exertion for healthy older adults (solid), AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. Error bars represent standard error.

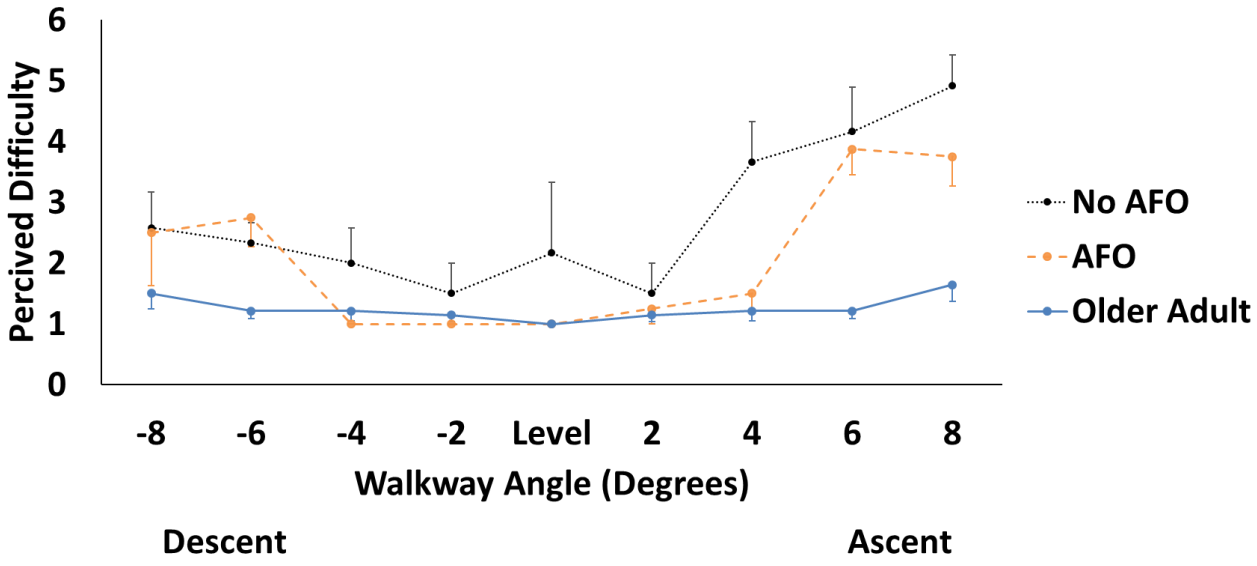


Figure D-2 Perceived difficulty for healthy older adults (solid), AFO wearers with (dashed) and without (dotted) their AFOs while walking down and up slopes. Error bars represent standard error.