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Hwan Choi, Keshia M. Peters, Michael MacConnell, Katie Ly, Eric Eckert, Katherine M. Steele

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# Impact of ankle foot orthosis stiffness on Achilles tendon and gastrocnemius function during unimpaired gait

## Authors and Affiliations:

Hwan Choi<sup>1</sup>, Keshia M. Peters<sup>1</sup>, Michael MacConnell<sup>1</sup>, Katie Ly<sup>2</sup>, Eric Eckert<sup>3</sup>, Katherine M. Steele<sup>1</sup>

<sup>1</sup>Mechanical Engineering, University of Washington, Seattle, WA, USA

Hwan Choi

Mechanical Engineering

University of Washington

3900E Stevens Way NE, Box 352600

Seattle WA, 98195, USA

Tel: 206-475-0772

FAX: 206-685-8047

E-mail: hwanc@uw.edu

<sup>1</sup>Mechanical Engineering, University of Washington, Seattle, WA, USA

Keshia M. Peters

Mechanical Engineering

University of Washington

3900E Stevens Way NE, Box 352600

Seattle WA, 98195, USA

Tel: 206-221-6153

FAX: 206-685-8047

E-mail: rumbek@uw.edu

<sup>1</sup>Mechanical Engineering, University of Washington, Seattle, WA, USA

Michael MacConnell

Mechanical Engineering

University of Washington

3900E Stevens Way NE, Box 352600

Seattle WA, 98195, USA

Tel: 509-423-4207

FAX: 206-685-8047

E-mail: mbm87@uw.edu

<sup>2</sup>Biophysics, University of Washington, Seattle, WA, USA

Katie Ly

Department of Physics

University of Washington

3910 15th Ave NE, Box 351560

Seattle, WA 98195, USA

Tel: 206-359-5365

FAX: 206-685-0635

E-mail: lykatie@uw.edu

<sup>3</sup>Human Centered Design & Engineering, University of Washington, Seattle, WA, USA

Eric Eckert

Human Centered Design & Engineering

University of Washington

3900E Stevens Way NE, Box 352315

Seattle WA, 98195, USA

Tel: 409-876-0038

FAX: 205-543-8858

E-mail: eric95@uw.edu

**Corresponding Author:**

Katherine M. Steele

Mechanical Engineering

University of Washington

Stevens Way, Box 352600

Seattle, WA, USA 98195

Tel: 206-685-2390

FAX: 206-685-8047

E-mail: kmsteele@uw.edu

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**Abstract**

Ankle foot orthoses (AFOs) are designed to improve gait for individuals with neuromuscular conditions and have also been used to reduce energy costs of walking for unimpaired individuals. AFOs influence joint motion and metabolic cost, but how they impact muscle function remains unclear. This study investigated the impact of different stiffness ankle foot orthoses (AFOs) on medial gastrocnemius muscle (MG) and Achilles tendon (AT) function during two different walking speeds. We performed gait analyses for eight unimpaired individuals. Each individual walked at slow and very slow speeds with a 3D printed AFO with no resistance (free hinge condition) and four levels of ankle dorsiflexion stiffness: 0.25 Nm / °, 1 Nm / °, 2 Nm / °, and 3.7 Nm / °. Motion capture, ultrasound, and musculoskeletal modeling were used to quantify MG and AT lengths with each AFO condition. Increasing AFO stiffness increased peak AFO dorsiflexion moment with decreased peak knee extension and peak ankle dorsiflexion angles. Overall musculotendon length and peak AT length decreased, while peak MG length increased with increasing AFO stiffness. Peak MG activity, length, and velocity significantly decreased with slower walking speed. This study provides experimental evidence of the impact of AFO stiffness and walking speed on joint kinematics and musculotendon function. These methods can provide insight to improve AFO designs and optimize musculotendon function for rehabilitation, performance, or other goals.

(Word Count: 226)

**Keywords:** ankle foot orthoses, ultrasound, musculoskeletal modeling, stiffness, Achilles tendon, gastrocnemius, gait

## **1. Introduction**

Ankle foot orthoses (AFOs) can be used to improve walking function (Figueiredo et al., 2008). Most AFOs are passive, exerting a torque about the ankle based upon the stiffness of the AFO's structure. In unimpaired individuals, passive AFOs have been shown to reduce metabolic costs of walking by up to 7% (Collins et al., 2015). However, this reduction is less than that predicted by various models and researchers have hypothesized this discrepancy is due to complex musculotendon dynamics during gait (Collins et al., 2015; Sawicki and Khan, 2016). Since the amount of force a muscle can produce is dependent on muscle length and velocity (Hill, 1953; Hill, 1938), if an AFO changes a muscle's length in unanticipated ways, AFOs may have counter-intuitive impacts on muscle force and metabolic costs. However, the impact of AFOs on the length of key muscles, such as the gastrocnemius, has not been experimentally evaluated.

The ankle plantarflexors play a critical role supporting and propelling the body during gait (Neptune et al., 2008; Steele et al., 2010). Previous studies have shown the contributions of the ankle plantarflexors to motion can be significantly altered when wearing AFOs (Collins et al., 2015; Delafontaine et al., 2017). For example, a case series of children with cerebral palsy evaluated the impact of AFOs on muscle contributions to mass center accelerations and found that AFOs significantly reduced gastrocnemius contributions to support and propulsion during gait (Ries, 2017). The gastrocnemius is a bi-articular muscle that generates large forces during activities of daily living (Finni et al., 1998, Sawicki and Ferris, 2009). It inserts into the Achilles

tendon (AT), the thickest and strongest tendon in the human body (Muffulli, 1999, Giddings, 2000), and stores and releases mechanical energy during gait (Cavagna et al., 1977; Farris and Sawicki, 2012). Given the importance of the gastrocnemius and AT to efficient gait, it is important to understand how AFOs will impact their function.

Quantifying muscle and tendon length during dynamic tasks like walking is challenging. The overall musculotendon unit (MTU) lengths can be estimated with musculoskeletal modeling using information about an individual's joint kinematics, musculoskeletal geometry, and musculotendon moment arms (Delp et al., 1990). Prior studies have evaluated the impact of various types of AFOs on MTU lengths in individuals with neurologic injuries, demonstrating AFOs can impact MTU stretching and shortening during gait, depending on AFO stiffness and other properties (Choi et al., 2016; Choi et al., 2015; Thompson et al., 2002). However, these methods are limited to estimating changes in overall MTU lengths, and do not differentiate between the relative muscle and tendon lengths.

Ultrasound can be used to experimentally monitor the relative length of the gastrocnemius and AT (Lichtwark and Wilson, 2006; Kalsi et al., 2016; Ishikawa et al., 2005). These methods have demonstrated that changes in AT length help reduce changes in gastrocnemius length, allowing it to operate near isometric conditions during gait (Fukunaga et al., 2001). If AFOs decrease ankle plantarflexor activity, they may alter the efficiency and function of the gastrocnemius and AT during gait (Figueiredo et al., 2008; Lichtwark and Wilson, 2006; Cronin et al., 2010; Kalsi et al., 2016). Using a mathematical model, one prior study suggested decreased gastrocnemius activity with an AFO may alter changes in muscle and tendon length, compromising muscle force

production and metabolic costs during gait (Sawicki and Khan, 2016); however, changes in gastrocnemius function with AFOs or other assistive devices has not been experimentally investigated.

In this study, we integrated ultrasound and musculoskeletal modeling to evaluate changes in AT and gastrocnemius length during gait with AFOs. Based upon prior research, we hypothesized that (1) as AFO stiffness increases, gastrocnemius muscle activity decreases leading to greater stretching of the gastrocnemius and a decrease in peak AT length during gait, and (2) as walking speed decreases, the gastrocnemius lengthens at a slower rate during stance, decreasing the sensitivity of musculotendon dynamics to different AFO conditions. By examining the impact of both AFO stiffness and walking speed on AT and gastrocnemius function, this research may inform future AFO design and prescription for both unimpaired individuals and individuals with neurologic injuries.

## **2. Methods**

### **2.1. Participants**

We recruited 9 unimpaired individuals with no history of lower-extremity orthopedic surgery (5 males and 4 females) to participate in this study. One subject (male) was excluded due to excessive out-toeing during gait, resulting in a total of 8 participants whose data were analyzed for this study (mean  $\pm$  standard deviation, height:  $170.0 \pm 7.9$  cm, weight:  $66.5 \pm 8.5$  kg, and age  $25.3 \pm 4.5$  years). Institutional Review Board approval was obtained for this study and participants consented prior to testing.

## 2.2. AFO Fabrication

A combination of 3D scanning, 3D printing, and traditional AFO hardware were used to create an AFO whose stiffness could be easily changed during the experiment (Fig. 1). At the first visit, the participants' dominant foot and shank were scanned (HandySCAN 300, Creaform) while their leg was maintained in anatomical position. To create an articulated AFO, a commercially-available metal hinge joint (Camber Axis Technic, Becker) was scanned and integrated with the AFO design. The acquired scanned data was post-processed using mesh design software (MeshMixer, Autodesk) with input on shape and structure provided by a certified orthotist. The hinge joint was aligned to the medial malleolus (Lundberg et al., 1989) to minimize error between the AFO and participant's rotational ankle axes. Similar to current AFO fabrication methods, the hinge joint for the lateral malleolus was positioned parallel to the medial hinge joint. The foot of the AFO covered from calcaneus to metatarsal providing free motion of the phalangeal joint. The bottom of the calcaneus and 5<sup>th</sup> metatarsal joints were flattened to provide stability and comfort. Two foam inserts were added to the lateral and medial hinge joints to prevent abrasion.

The model was trimmed to the shape of the AFO and assigned a 4 mm thickness, similar to current articulated AFOs. Two supports, used to attach elastic polymer bands to adjust AFO stiffness, were positioned posteriorly on the shank and foot segments in line with the AT. AFOs were printed with a fused deposition modeling 3D printer (FlashForge Creator Pro, FlashForge) with polylactide (PLA) filament (print settings: 100% infill, 0.2 mm layer height, 2 shells). A sample of one of the AFO models is available on



WEBSITE (to be posted upon publication). The elastic polymer bands were fabricated with polyurethane (PMC 780 Dry, Smooth-On) to control the stiffness of the AFOs. Four polymer bands were created for the four AFO stiffness conditions. The stiffness conditions ( $0.25 \text{ Nm} / ^\circ$ ,  $1 \text{ Nm} / ^\circ$ ,  $2 \text{ Nm} / ^\circ$ ) were selected based on the reported range of current clinical prescriptions and a stiffness value ( $3.7 \text{ Nm} / ^\circ$ ) previously reported to reduce metabolic costs during unimpaired gait (Collins et al., 2015). Using a universal testing machine, tensile force of the polymer bands was measured and the moment from the AFO (M) was calculated by multiplying the orthogonal distance from the Camber joint rotational axis to the elastic polymer band (MA), by the tensile force (F) of the elastic polymer band. Small clips on the posterior supports of the AFO were used to adjust the MA and match the desired stiffness levels. Velcro straps around the shank and the foot were used to secure the AFO.

### 2.3. Gait Analysis

Each participant walked on a single-walking-belt treadmill with five AFO conditions: free hinge mode (no resistance) and AFO stiffnesses of  $0.25 \text{ Nm} / ^\circ$ ,  $1 \text{ Nm} / ^\circ$ ,  $2 \text{ Nm} / ^\circ$ , and  $3.7 \text{ Nm} / ^\circ$  to resist ankle dorsiflexion. We asked each participant to identify their dominant leg used for tasks, such as kicking, and fabricated their AFO for their dominant leg. Since a previous study demonstrated MTU length and joint kinematics vary with walking speed, each participant walked on the treadmill at two non-dimensional walking speeds. Non-dimensional walking speed was defined as  $\bar{v}^* = v / \sqrt{\text{gravity} \times \text{leg length}}$ , and this study used very slow ( $\bar{v}^* = 0.19$ ) and slow ( $\bar{v}^* = 0.29$ )

speeds on the treadmill (Liu et al., 2008). These speeds were selected to match the common speeds of individuals with neurologic disorders such as stroke or cerebral palsy. The order of each stiffness condition and non-dimensional walking speed was randomized for each participant. Participants acclimatized to each walking speed and AFO condition for a minimum of 2 minutes, after which a recording of at least 25 gait cycles was collected for each condition.

For gait analysis, an eight-camera infrared motion capture system (Vicon, Motion Systems Ltd.) was used to acquire three-dimensional marker data at 100 Hz based on a modified Helen-Hays marker set with 40 markers (Kadaba et al., 1990). Since ankle kinematics and AT length can be sensitive to the location of foot markers, extra measurements were taken to evaluate the location of each marker with respect to the shoe and bony landmarks (see Supplementary Material for sample data collection worksheet). A pressure switch (MA-153, Motion Lab Systems) placed beneath the calcaneus was also used to evaluate potential motion between the shoe and AFO during gait. All subjects wore the same type of shoe (UM311F102, Sintetico) and the pressure sensor demonstrated that the shoe and AFO remained in contact throughout the gait cycle.

To evaluate the impact of AFO stiffness on AT length, an ultrasound (Logicscan 128, Telemed) recording at 61.4 Hz with a linear transducer (LV7.5/60/128Z-2, 59mm, Telemed) was attached to the shank and used to track the location of the medial gastrocnemius (MG) MTJ. An L-shaped ultrasound holder was fabricated with a 3D printer and used to provide a stable mount for the transducer and consistent marker locations. Three reflective markers were attached to the ultrasound holder to provide a

coordinate system for tracking transducer position relative to the lab coordinate system. The location of the markers relative to the origin and axis of the ultrasound images was established by aligning the  $x'$  axis to the proximal side of ultrasound image and the  $y'$  axis to the lateral side of ultrasound image (Fig. 2 b).

Gastrocnemius activity was measured using surface electromyography (EMG) data (Trigno<sup>TM</sup> Wireless EMG, Delsys). EMG data was recorded at 1111 Hz, with hardware signal processing including a band-pass filter (20-460 Hz). Using custom Matlab software, the EMG signals were rectified and low-pass filtered (6 Hz, 4<sup>th</sup> order Butterworth) to create a linear envelope for analysis. Muscle activity was normalized to the averaged maximum value across all trials walking with the hinged AFO condition. Participant 2's EMG data was excluded due to poor signal quality.

A custom-made signal generator was used to synchronize the ultrasound, motion capture, EMG systems and pressure sensors. We used the ultrasound to synchronously initiate data collection by transmitting a 5 Volt signal to the signal generator, which then sent trigger signals to the motion capture and EMG systems.

#### 2.4. Musculotendon Dynamics

OpenSim, an open-source musculoskeletal modeling and simulation software platform (Delp et al., 2007), was used to calculate gait kinematics, AT length, and MG length. A generic musculoskeletal model with 37 degrees of freedom and 92 muscle actuators (Hamner et al., 2010) was scaled to each subject based on anatomical landmarks. The degrees of freedom in the model included a ball-and-socket joint at

each shoulder, a hinge joint at each elbow, a combination saddle and condyloid joint at each wrist, three translations and three rotations of the pelvis, a ball-and-socket joint between the pelvis and the torso located at the third lumbar vertebrae, a ball-and-socket joint at each hip, a combination joint of translations and rotations at each knee, and a revolute joint at each ankle. We created an ultrasound object in OpenSim that allowed six degrees of freedom (three translations and three rotations) with respect to the tibia segment of the musculoskeletal model. For each trial, the distance between experimental reflective marker trajectories from gait analysis and virtual markers on the OpenSim generic model was minimized to calculate joint angles and position of the ultrasound transducer using inverse kinematics. The MG MTU length was calculated from the path between origin and insertion of the muscle during the gait cycle (Choi et al., 2016; Choi et al., 2015; Thompson et al., 2002).

Ultrasound images during each trial were acquired as movie data with a size of 823 by 512 pixels. In accordance with prior studies, these images were used to manually track the MTJ position using ImageJ (NIH, Maryland), an open-source image processing software (Randhawa and Wakeling, 2013). The horizontal pixel size of the ultrasound image was correlated with the ultrasound linear sensor array to convert from the number of pixels to physical distances. The MTJ trajectories were manually identified as the point where the deep MG aponeuroses and external AT intersect on the ultrasound images (Rosso et al., 2012). The two-dimensional pixel position representing the MTJ position was transformed into the ultrasound transducer coordinate system in OpenSim to visualize the position of the MTJ relative to the tibia during gait (see video in Supplementary Material II).

The insertion of the MG MTU from OpenSim and the MTJ trajectories from ultrasound were transformed into OpenSim's global coordinates to calculate changes in AT length during gait (Fig. 2). The MG muscle belly length was calculated by subtracting the AT length from MTU length at every time point. To compare between participants, changes in MTU, MG muscle, and AT lengths were normalized to MTU length when the scaled musculoskeletal model was in anatomic position - full knee extension and neutral ankle angle, as done in previous studies (Choi et al., 2016; Choi et al., 2015; Arnold et al., 2006). MG muscle velocity was calculated by numerically differentiating the normalized muscle length with respect to time.

### *2.5. Data Analysis*

We used R (R Core Team, 2012) and lme4 (Bates et al., 2014) to perform a linear mixed effects analysis to evaluate how different AFO stiffness and walking speed impacts ankle angle, AT length, MTU length, MG muscle length, and MG velocity during mid-terminal stance (10-60% of the gait cycle). AFO stiffness and walking speed were included in the model as fixed effects with random effects for the intercepts of each participant. Visual inspection of residual plots did not reveal any obvious deviations from homoscedasticity or normality. P-values were obtained by likelihood ratio tests of the full model against the model without the effect in question between participants and walking speeds. Significance was defined as  $\alpha < 0.05$ .

## **3. Results**

AFO stiffness impacted joint kinematics and musculotendon lengths while walking on a treadmill. As an example, Figure 3 illustrates the impacts of AFO stiffness on kinematics and musculotendon lengths for a representative participant walking at a slow speed. As AFO stiffness increased, this participant exhibited increased hip extension in terminal stance, a slight increase in knee extension in stance, and decreased dorsiflexion in terminal stance. These changes in knee and ankle kinematics contributed to a 3.3, 11.6, 20.3, and 31.4 Nm increase in the AFO moment with AFO stiffness conditions of 0.25, 1, 2, and 3.7 Nm / ° compared to the hinged AFO condition. For this participant, while peak MG MTU length was similar across AFO stiffness conditions, peak AT length decreased and MG muscle length increased in mid-stance with increasing AFO stiffness. Changes in AT length contributed to more of the change in MTU length (57-58%) than MG length during stance across AFO conditions.

Across participants, as AFO stiffness increased, peak knee extension angle ( $p < 0.0001$ ) and peak ankle dorsiflexion angle ( $p < 0.0001$ ) significantly decreased, with greater AFO moments and increasing AFO stiffness ( $p < 0.0001$ ) for both the slow and very slow walking speeds (Fig. 4). Peak gastrocnemius MTU length ( $p < 0.0001$ ) and peak AT length ( $p < 0.0001$ ) during mid- to terminal-stance significantly decreased as AFO stiffness increased (Fig. 5 a, d), although the effect size was small. On average, there was a 0.41 % change in MTU length and 2.89 % change in AT length between the hinge and 3.7 Nm / ° conditions at slow walking speed and 0.81 % and 2.14 % change in MTU and AT lengths at the very slow walking speed. AT length at heel contact significantly increased as AFO stiffness increased ( $p = 0.0002$ ) and walking speed decreased ( $p < 0.0001$ ) (Fig. 5 b).

There was a significant increase in peak MG length ( $p = 0.0016$ ) with increasing AFO stiffness at both walking speeds across participants (Fig. 5 c), although there was more variability among participants when comparing MTU and AT lengths (see Supplementary Figure 1 for results of each participant). For example, during slow walking, three participants had decreased peak MG lengths as AFO stiffness increased beyond  $2 \text{ Nm} / ^\circ$  while the other participants' peak MG length continued to increase. During very slow walking, two participants had a shortened peak MG length with an AFO stiffness of  $3.7 \text{ Nm} / ^\circ$  compared to  $2 \text{ Nm} / ^\circ$ , but the other participants had consistent increases in peak MG length with increasing AFO stiffness. Increasing AFO stiffness did not impact peak MG activity ( $p = 0.5967$ ) and peak MG eccentric velocity ( $p = 0.5967$ ) at either walking speed (Fig. 5 e, f).

Walking speed significantly reduced peak MG length ( $p = 0.0009$ ), peak MG activity ( $p = 0.0025$ ), and peak MG eccentric velocity ( $p < 0.0001$ ) (Fig. 5 c, e, f). However, peak knee extension angle ( $p = 0.4381$ ), peak ankle dorsiflexion angle ( $p = 0.4264$ ), peak AFO moment ( $p = 0.7464$ ), peak MTU length ( $p = 0.3510$ ), peak AT length ( $p = 0.1100$ ) were not significantly affected by walking speed. These results suggest walking speed significantly impacted the force generating capacity of the MG, with minimal changes in overall kinematics.

#### **4. Discussion**

This study provides experimental evidence of the impacts of AFO stiffness and walking speed on joint kinematics and musculotendon function using 3D motion capture,

ultrasound, and musculoskeletal modeling. Our results demonstrated that AFO stiffness and walking speed significantly impacted joint kinematics and musculotendon function during unimpaired gait, with trends of increasing changes in AT and MG lengths with increasing AFO stiffness. Prior studies (Lichtwark and Wilson, 2006; Cronin et al., 2010; Kalsi et al., 2016) have used ultrasound to investigate the operating length of the MG during unassisted walking, and our results found similar musculotendon lengths as these prior studies during the no resistance (hinged) condition. Similar to these prior studies, we found the passive elastic properties of the AT helped reduce muscle lengthening velocity during gait (Ishikawa et al., 2005; Cronin et al., 2013).

This research demonstrated that AT length was significantly impacted by AFO stiffness during unimpaired gait. Although the difference was small, peak AT length decreased as AFO stiffness increased, suggesting the AT may store less energy while walking with an AFO. A prior study suggested that unimpaired individuals use a consistent net ankle moment regardless of AFO assistance (Collins et al., 2015). While kinetics were not available for this experiment, the changes in MG activity and AT length in this study support these prior hypotheses. Future studies that can also incorporate ground reaction forces could provide further insight into the impacts of AFO stiffness on musculotendon and gait dynamics.

Evaluating rate of change in peak MG length can also provide insights into the mechanisms underlying the interplay between walking speed and AFO stiffness. MG function is influenced by walking speed, with decreasing contributions to support and propulsion at slower walking speeds (Liu et al., 2008; Neptune et al., 2008). Our results demonstrated that slower walking speeds decreased MG's eccentric velocity (Neptune



and Sasaki, 2005; Farris and Sawicki, 2012; Arnold et al., 2013), peak MG length, and MG activity. These results suggest that for individuals who use a slower walking speed with an AFO may not receive the level of energy savings expected due to changes in MG operating length and velocity. Further, at slower walking speeds, similar levels of energy saving may not be necessary.

This study also demonstrated the variability in responses to AFOs, even among unimpaired individuals. As AFO stiffness increased, some participants' increased MG activity while others continued to decrease MG activity with increased AFO stiffness. Since no training or coaching was given to the participants in this study, the differences reflect variable responses to the potential assistance provided by a passive AFO. Prior studies of unimpaired individuals walking with active and passive AFOs have also reported variable changes (Collins et al., 2015; Cain et al., 2007), which may be related to different strategies when walking with an AFO, individual differences in size or strength, or other factors. Providing specific instructions when introducing the AFOs may help to optimize results across participants.

The methods described in this manuscript can also be used to evaluate musculotendon function for individuals with neurological impairments, such as individuals with cerebral palsy. Muscle contracture, characterized by short and tight muscles, are common for these individuals and can contribute to pathologic movement (Barber et al., 2017). Since AFOs are worn all day, understanding whether they stretch or shorten the MG could have important implications for rehabilitation (Zhao et al., 2011; Theis et al., 2015; Pin et al., 2006). Prior studies demonstrated that walking with currently prescribed AFOs or inclined treadmill walking can increase MG belly operating

length (Hösl et al., 2016;). This study demonstrated that, although there were only small changes in MTU length with different AFOs, the MG had larger changes in length and using ultrasound to quantify changes in MG length may guide AFO prescription for rehabilitation goals. Determining the magnitude of change in AT or MG length which are necessary for specific goals, such as reducing metabolic costs or stretching the MG, remain important unanswered questions.

While ultrasound provides insight into AT and MG muscle belly length, there are further musculotendon properties that may influence function with AFOs that cannot be quantified with ultrasound alone. In this study, we oriented the ultrasound probe so that the MTJ was positioned in the middle of the screen, so that MG belly length could be measured. We did not measure fascicle length or pennation angle, which would require a more proximal placement of the transducer. Examining MG fascicle length and pennation angle could provide insight on how AFO stiffness changes muscle force-generating capacity (Randhawa and Wakeling, 2013; Zajac, 1989). Further, muscle force is dictated by sarcomere length, which cannot be estimated from ultrasound. While it is often assumed that muscle adapts to operate near optimal sarcomere length, this assumption has not been extensively evaluated in humans and may also not hold true for individuals with neuromuscular disorders (Mathewson et al., 2015).

This study examined MG and AT lengths with varying AFO stiffness in a limited sample of eight unimpaired individuals walking at slow walking speeds. While this study was designed to evaluate peak MG and AT length, the impact of other factors such as changes in MG EMG magnitude may require a larger number of participants. More participants' may also clarify sources of inter-subject variability observed in this study

and evaluate responses at self-selected and faster walking speeds. We did not provide specific guidance to the participants about how to walk with the elastic polymer bands on the AFOs, rather participants adopted their own strategies. We tested over the course of a single day, which also limited opportunities for evaluating learning and long-term adaptation to a new device. These limitations represent important areas for future investigation that can further inform the design and prescription of AFOs for performance and rehabilitation goals.

## **5. Conclusion**

This study evaluated how AFO stiffness impacts kinematics and MG function during unimpaired gait. We found that, as AFO stiffness increased, peak AFO dorsiflexion moment increased, while peak MTU and AT lengths decreased. With slower walking speeds, peak MG activity, length, and velocity also significantly decreased; however, there was significant variability between participants. Future studies are required to evaluate how AFO stiffness impacts musculotendon and walking function for individuals with neurologic injuries. These methods and results can assist in the design and prescription of AFOs to improve gait and muscle function during daily life.

(Word Count: 3998)

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***Conflict of Interest Statement***

The authors declare no conflict of interest of this research.

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Figure 1: AFO fabrication method: (a) Scanning a participant's foot and shank with a 3D scanner. Socks with patterns and markers help improve the accuracy of the scan. The scans were used to customize the (b) AFO model with computer aided design software. The (c) AFO was printed with PLA and a polymer band was used to adjust stiffness between conditions. The stretch of the polymer band, measured with markers, determined the force ( $F$ ), which was multiplied by the moment arm ( $MA$ ) to calculate the AFO moment ( $M$ ) resisting ankle dorsiflexion



Figure 2: Experimental set-up: (a) Example position of the AFO, ultrasound transducer, and marker positions. (b) The musculoskeletal model illustrates the position of the ultrasound transducer and coordinate systems used to determine the position of the MTJ and quantify AT and MG lengths.

Figure 3. Average joint kinematics at the (a) hip, (b) knee, and (c) ankle for Participant 4 walking at a slow speed. There were minimal changes in (d) MTU length with AFO stiffness, while (e) peak AT length decreased and (f) peak MG length in terminal stance increased as (g) peak AFO moment increased. This participant showed minimal changes in (h) MG velocity, while (i) MG activity decreased. AT and MG lengths were normalized to the length of the MTU when the scaled musculoskeletal model was in anatomic position.

Figure 4. Effects of AFO stiffness and walking speed on (a) peak knee extension angle, (b) peak ankle dorsiflexion angle, and (c) peak AFO moment during single limb support. Black horizontal lines in each box depict median value, upper and lower boundary of each box represents upper and lower quartiles respectively. Upper and lower error bars represent maximum and minimum values, respectively. Arrows represent a significant change ( $p < 0.001$ ) with increasing AFO stiffness from the linear mixed-effects regression models.

Figure 5. Effects of AFO stiffness and walking speed on (a) normalized peak AT length, (b) normalized AT length at heel contact, (c) normalized peak MG length during terminal stance, (d) normalized peak MTU length, (e) peak MG activity during stance, and (f) normalized peak MG eccentric velocity. AT and MG lengths were normalized to the length of the MTU when the scaled musculoskeletal model was in anatomic position. Black horizontal lines in each box depicts median values, upper and lower boundary of each box represents upper and lower quartiles respectively. Upper and lower error bars represent maximum and minimum values, respectively. Circles in figure e and f depicts outliers ( $> 2$  SD from mean). Inclined arrows and vertical arrows represent a significant change ( $p < 0.001$ ) with increasing AFO stiffness and walking speed respectively from the linear mixed-effects regression models.







