THE RELATIONSHIP BETWEEN PLANTARFLEXOR MOMENT ARM, MUSCLE ACTIVATION PATTERNS AND GAIT VELOCITY IN THE ELDERLY

A Thesis
by
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Abstract

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Previous research suggests a link between the onset of functional dependence, mortality, and reductions in gait velocity among elderly, potentially associated with the effects of aging on the musculoskeletal and nervous systems. Yet underlying processes contributing to these reductions are not well known. The purpose of this study was to investigate ankle structure and lower limb muscle activation to identify differences which could be related to reduced gait velocity seen in aging. Plantarflexor moment arm (PFMA) was estimated from medial/lateral images of the foot. Sixteen subjects (age = 69.81 ± 4.72) performed a Six-Minute Walk Test while surface electromyography (EMG) was recorded from the TA, SOL, MG, and LG of the right leg. After EMG normalizing to the within trial peak, muscle bias was calculated as the area of m1 (MG or SOL) EMG over the sum of the area of m1 and m2 (LG) EMGs during the stance phase. Relationships between variables of interest (gait velocity, muscle moment arm and muscle bias) were investigated using

correlational analysis. No association was found between gait velocity and PFMA ($r = -0.13$, $p = 0.627$) but a low positive correlation was found between effective mechanical advantage (EMA) and medial gastrocnemius (MG-LG) bias during the stance phase ($r = 0.42$, $p = 0.108$). The present study does not confirm links between moment arm and gait velocity or stance phase muscle bias in elderly observed previously. Elderly subjects might not modify neuromuscular control similar to what has been shown in young individuals with lower EMA. Previously suggested relationships between these variables may have task-intensity dependencies relative to the groups studied. Further investigation of muscle bias measures influencing gait velocity were considered. Low positive correlations were found between gait velocity and stance phase MG-LG bias ($r = 0.42$, $p = 0.11$), stance phase soleus (Sol-LG) bias ($r = 0.39$, $p = 0.14$), and pre-stance MG-LG bias ($r = 0.44$, $p = 0.09$). A moderate positive correlation (significant) was found between gait velocity and pre-stance Sol-LG bias ($r = 0.54$, $p = 0.03$). Increases in Sol-LG bias during pre-stance may indicate efforts to stiffen the ankle joint in preparation for single leg support which was shown to positively influence gait velocity in this group. Yet, given the difference between correlations among variables compared to previous studies, it is plausible to suspect that adaptations of the neuromuscular system may not have been present in this group. Indicating the possibility of dedifferentiation and loss of complexity within a younger elderly group that was physically active. To this point, the influence of ankle joint leverage and lower leg neuromuscular activation patterns in elderly gait decline remain unclear.

Acknowledgments

I would first like to thank my committee: Dr. Needle and Dr. Fasczewski for their guidance in further developing the literature review and methodology for this thesis. I also appreciate the time both of you have spent to encourage my professional development and help me toward future opportunities. I would like to give a special thanks to my advisor: Dr. van Werkhoven for all of his hard work throughout this thesis process. Herman, I really appreciate the time you have devoted to help me get this thesis to a finalized document. I would like to thank some fellow graduate and undergraduate students: David Schumacher, Jasmine Cash, Sarean Gaynor-Metzinger, Jonatan Seith, and especially Kevin Page for their help in pilot testing and data collection.

Last, I would like to give a special thank you to my family and Katie. It has been a challenging experience but I appreciate all of your support throughout my life to push me toward things you believe I am capable of achieving.

Dedication

This project is dedicated to my family, in memory of Uncle Chris, to thank them for the continued love and support they provide to their children, grandchildren, and to each other. I would not be where I am without you, and for that I am forever thankful.

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Introduction

The growing population of elderly people over the age of 65 poses challenges for the healthcare system. Elderly will comprise 20 percent of the United States population by 2030 (Kelmar, Gerberding, & U.S..Department, 2007). Evidence of associations between physical capabilities and all-cause mortality have been found in older community-dwelling populations, with grip strength, walking speed, chair rising, and standing balance showing importance in predicting mortality rates in people older than 70 years of age (Cooper, Kuh, Hardy, & Mortality Review Group, 2010). Other evidence provides a link between walking speed and the onset of functional dependence in activities of daily living. For adults aged 65 to 74, maximum walking speed has been shown to be the most sensitive in predicting onset, while preferred speed the most sensitive for those ≥ 75 years (Shinkai, 2000). Components of intra-individual and environmental factors substantiate difficulties in understanding mobility decline . For adults > 70 years, a 2015 World Health Report on Aging and Health indicated that impairments of the musculoskeletal system exceed other noncommunicable diseases (Briggs et al., 2016). Muscle strength and balance are commonly studied in mobility impairments as these closely relate to a walking task (Rantakokko, Mänty, & Rantanen, 2013). Investigating musculoskeletal factors underlying predictors in mobility decline may provide assistance in developing more robust strategies for early detection and intervention.

Previous research highlights a longitudinal loss in muscle strength among elderly men and women. Elderly experience a three times greater loss in strength compared to muscle mass when tracked over a three year period (Goodpaster et al., 2006). Components of muscular, neural, and skeletal systems may contribute to this loss in strength. Sarcopenia, the age-related loss in muscle mass and strength, continues to be a challenging pathology studied within aging populations (Marcell, 2003). Mechanisms thought to contribute to this disease include disuse atrophy, changes in protein metabolism and hormones, mitochondrial dysfunction, inflammation, impairments of muscle tissue regeneration pathways, alterations in redox homeostasis, and changes is neuromuscular function (Bowen, Schuler, & Adams, 2015; Marcell, 2003). The nervous system guides the voluntary control of movement. It may also be a contributor to strength losses observed in sarcopenia as aging affects the peripheral nervous system through impairments of axonal pathways (Orssatto, Wiest, & Diefenthaeler, 2018; Öztürk et al., 2013). Additionally, strength is generated through forces imparted across joints. Therefore, the structural properties of the joint contribute to the overall mechanical advantage of the muscles to produce movement.

Across a range of locomotor speeds, the ankle and hip joints have been found to be the highest contributors to overall gait power in healthy subjects under the age of 30 years old. In walking speeds which ranged from 0.75 to 2.0 m/s, the ankle joint contributes between 40 and 44% of overall positive power (Farris & Sawicki, 2012). This relationship is notable as walking speed decreases with age along with the ability to generate power at the ankle joint (Winter, Patla, Frank, & Walt, 1990). Studies on elderly gait show reductions in plantarflexor power, plantarflexor range of motion, and increases in hip extension and flexion occurring during the late stance phase (Judge, Davis, & Ounpuu, 1996; Kerrigan, Todd, Croce, Lipsitz, & Collins, 1998). These gait redistributions were found to parallel step length, as one study shows a 10 percent reduction in step length with advancing age (Judge et al., 1996). Previous studies have also compared the differences in isometric and concentric ankle performance between young and old subjects. Both confirmed older subjects generate lower plantarflexor torque about the ankle (Candow & Chilibeck, 2005; Simoneau, Martin, & Van Hoecke, 2005). Several studies have specifically considered the influence of the PFMA on physical performance (e.g. Baxter, Novack, van Werkhoven, Pennell, & Piazza, 2012; Lee & Piazza, 2009; Raichlen, Armstrong, & Lieberman, 2011; Scholz, Bobbert, van Soest, Clark, & van Heerden, 2008). PFMA is the perpendicular distance measured between the line of action of the plantarflexor muscles and the center of the ankle. This measure is critical in determination of ankle joint torque (Figure 1).

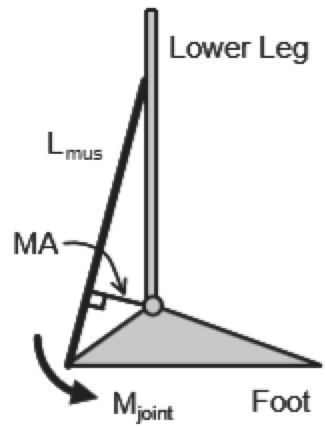

Figure 1. Definition of PFMA (shown as MA). The perpendicular distance between the ankle joint center and Achilles tendon line of action (Nagano & Komura, 2003).

In studies on young adults, this measure has been shown to be related to sprinting abilities, with sprinters exhibiting smaller moment arms than controls (Baxter et al., 2012; Lee & Piazza, 2009), and there has also been evidence that smaller moment arms are correlated with a better running economy (Raichlen et al., 2011; Scholz et al., 2008). A few studies have also investigated how PFMAs could affect gait in the elderly. Lee & Piazza (2012) indicated a strong correlation between the Six-Minute Walk Test and PFMA for elderly that were slow walkers, suggesting that PFMA may have a strong influence on gait velocity in elders who are unable to compensate through strength adaptation at the level of the muscular or neural systems (Lee & Piazza, 2012). More recently, Rasske & Franz (2018), showed reductions in peak ankle moment during the push-off phase of gait are correlated with smaller PFMAs in elderly subjects. The elderly also exhibited 11 percent smaller dynamic PFMAs than younger counterparts during walking (Rasske & Franz, 2018). These results indicate that the PFMA could play a critical role in determining the ability of the elderly to perform locomotor activities. The exact mechanism by which this occurs in unclear.

The association between PFMA and gait velocity could in part be explained by the inability of elderly to compensate through strength adaptation at the muscular and neural level (Lee & Piazza, 2012). Differences in muscle activity have previously been linked to varying PFMAs in young individuals. Sedentary individuals with lower effective mechanical advantage (EMA), or ratio of PFMA to external MA of the foot, relied more upon the medial head of the gastrocnemius (MG) during walking (Ahn, Kang, Quitt, Davidson, & Nguyen, 2011). Although the proposed relationship between moment arm and gait velocity was not addressed in this study with younger individuals (Ahn et al., 2011), it is possible that variations in muscle activation could partly explain differences in gait velocity within elderly. Because differences in moment arm may be linked to differences in muscle activation, it is plausible that elderly who struggle to maintain greater walking speeds, have smaller moment arms that are associated with changes in muscle activation patterns. Younger individuals might be able to adapt to a biomechanical constraint via adaptations of motor control and muscle size, whereas elderly may not have the ability to do so. Therefore, the purpose of this master's thesis is to investigate elderly ankle structure and muscle activation patterns to identify differences which may lead to reduced gait velocity seen in aging.

Specific Aims

- Investigate the association between PFMA and gait velocity in elderly individuals

- Investigate the association between PFMA and triceps surae muscle activation patterns during gait

Hypotheses

- Subjects with smaller PFMAs will walk at a slower preferred gait velocity
- Subjects with smaller EMA will walk with more bias toward the medial gastrocnemius head

Literature Review

Aging affects a number of human organ systems which influence health and may lead to lower quality of life. For the scope of this review, aging is confined to its effects on the musculoskeletal and neural systems. Furthermore, the concern focuses on how changes in these systems impact locomotion in aging populations as they are at the highest risk of mobility-related comorbidities (Cooper et al., 2010).

Aging and Gait Velocity

Physical capabilities such as gait velocity have been linked with mortality in aging populations (Cooper et al., 2010). Former studies support this as the risk of mortality increases as velocity decreases. For the age group 70 to 79 years, subjects with faster rates of decline in gait velocity (0.03 m/s per year) are at a 90 percent higher risk of mortality than those in the slow declining group (0.02 m/s per year) (White et al., 2013). In addition, subjects in the slower declining group started at 1.37 m/s while subjects in the faster declining group started at 0.9 m/s. This is important as the fast declining group gait velocity was reduced by 21 percent over an eight year period (White et al., 2013). Others have quantified five and ten year survival rates for subjects age 65 to ≥ 85 years, showing that survival rates tend to increase as gait velocity increases from 0.4 m/s to 1.4 m/s (Studenski et al., 2011). In combination, these studies support the need for research in elderly locomotion. Gait velocity declines by some percentage every year with the magnitude depending on the individual and survival rates depend on how well mobility is able to be maintained over time through management and intervention strategies.

Gait velocity may decline with age due to reductions in step length (Himann, Cunningham, Rechnitzer, & Paterson, 1988). Ankle plantarflexor power generation during the late stages of the stance-phase contributes to 52 percent of the variation in step length (Judge et al., 1996). Elderly exhibit lower anterior-posterior ankle generated mechanical work during plantarflexion in terminal stance leading to less propulsion and shorter swing phases (Ko, Hausdorff, & Ferrucci, 2010). Reductions in plantarflexor strength and power have been correlated with reduced gait velocity in elderly men and women (Bendall, Bassey, & Pearson, 1989). The redistribution of gait mechanics has been discussed previously (DeVita & Hortobagyi, 2000; Judge et al., 1996; Kerrigan et al., 1998). The effects of aging on gait mechanics leads to a redistribution of power and torque at the hip and ankle joint. Significant differences in plantarflexor power, plantarflexor range of motion, and peak hip extension have been observed between young and elderly and comfortable versus fast walking velocities (Kerrigan et al., 1998). Elderly adults rely more on hip extensors and less upon the ankle plantarflexor muscle group during walking (DeVita & Hortobagyi, 2000). The discussion as to why this occurs in still ongoing.

Aging and the Nervous System

The nervous system, specifically the somatic system associated with voluntary control of movement, may contribute to muscle degradation leading to functional declines with increasing age. The peripheral nervous system is subject to a loss of motor neurons incurring damages to axonal pathways through a process called “dying back” (Öztürk et al., 2013). The combination of aging and peripheral nervous system exposure leads to neuronal cell death in motor neurons of primarily type II motor units (MU) (Gordon, Hegedus, & Tam, 2004). This process occurs at a more accelerated pace after the sixth decade of life and elderly over the age of 70 show evidence of 50 percent fewer motor neurons than younger adults (Manini, Hong, & Clark, 2013; Pannese, 2011). Furthermore, this deficit is more significant in the lower limbs and concurrent with this event are noted reductions in muscle mass and force generating capacity (Manini et al., 2013;

McKinnon, Montero-Odasso, & Doherty, 2015). To combat the denervation of muscle fibers due to “dying back”, the neuromuscular system undergoes a process of remodeling. Muscle fibers within the motor unit of a deteriorating motor neuron become denervated and will only be re-innervated by dendritic sprouting of an adjacent motor neuron (Piasecki, Ireland, Jones, & McPhee, 2016). Since primarily type II MUs are prone to denervation, the orphaned muscle fibers are re-innervated by motor neurons from type I MUs (Manini et al., 2013). As the process of remodeling in motor units proceeds with age, it becomes less efficient in re-innervating orphaned fibers. The accumulation of denervated fibers is the result of unsuccessful re-innervation due to limited axonal sprouting. Since the re-innervating capacity of the nervous system is impaired, the orphaned muscle fibers atrophy and become non-functional (Aare et al., 2016). Clark et al. (2013) found reductions in muscle activation and a trend of reduced muscle cross-sectional area in the thigh muscles of elderly men after a three-year period. These findings were accompanied by reductions in knee extensor power (Clark et al., 2013), suggesting that the cyclic efficiency of the nervous system is a determinant of muscle size and strength losses seen in aging.

Inadequate remodeling may require adaptations within other motor units to maintain muscle force, essentially increasing motor unit size and slowing contractile speeds (Doherty & Brown, 1997; Power, Dalton, & Rice, 2013). This may have a large effect on muscle force production. Previous research shows differences in maximal voluntary contractile torque, twitch tension, and speed of contraction between young and elderly subjects with equal voluntary activation and firing rates (Kirk, Copithorne, Dalton, & Rice, 2016). Other research shows reduced firing rates and doublet patterns in healthy elderly adults, which paired with longer time to reach peak force in ankle dorsiflexor muscles during rapid contraction (Klass, Baudry, &

Duchateau, 2008). Similarly, elderly subjects' responses to stimulation of the gastrocnemius muscles were delayed and of lower magnitude when compared to young (Petrella, Cunningham, Vandervoort, & Paterson, 1989). These findings support neural degradation due to aging, which impair the upper limits of motor neuron function, neuromuscular mechanics, and provide insight in to the efficiency of the nervous system to adapt with age is thus a determinant of muscle force production.

Neural control of locomotion is challenging to describe for any age population. This is due to the variability and redundancy of motor control. Individual patterns to achieve locomotion, even at a small level, may differ. Studying the elderly may increase that difficulty as they are shown to have higher sensorimotor variability (Christou, 2011; Lin & Faisal, 2018) and variability of coordination patterns (Vernooij, Rao, Berton, Retornaz, & Temprado, 2016). Elderly exhibit higher coactivation in their lower extremities, possibly due to impairments of peripheral pathways leading to more reciprocal inhibition in the spinal cord and co-contraction from antagonist muscles (Aagaard, Suetta, Caserotti, Magnussion, & Kjaer, 2010), which negatively affects gait velocity (Lee, Chang, Choi, Ryu, & Kim, 2017). This leads to reduced force and power generation due to opposing forces from the muscles (Ortega & Farley, 2015).

Other research has focused on activation patterns of the triceps surae during gait. In a study on energetic efficiency in walking, young subjects (28 ± 5 years) were found to rely more on the MG shortening for propulsive force at the end of the stance phase at a 20 percent faster pace. The MG was found to contract 20 percent faster in the push-off phase during these faster walking conditions. Fascicle lengths were observed to decrease with prolonged walking. In combination, increasing contractile speeds and decreasing fiber lengths may have led to lower force capability of the MG in this study (Cronin, Avela, Finni, & Peltonen, 2013). This was accompanied by consistent contractile behavior in the SOL muscle for all conditions, suggesting that the SOL may play an important functional role in prolonged walking and faster paced as the ability of the MG decreases (Cronin et al., 2013). Lending insight to the sustainability of the somatic system to accommodate energy efficiency during prolonged walking and walking at different paces. The SOL comprises more type I muscle fibers allowing it to use less energy to complete the same task (Cronin et al., 2013). Prolonged walking, lasting more than a few minutes, shows increases in the compliance of the Achilles tendon possibly causing need for motor control changes to optimize efficiency. Prolonged walking causes a 10 percent decrease in MG reliance and 10 percent increase on the SOL (Cronin, Peltonen, Sinkjaer, & Avela, 2011). Which suggests that the nervous system is able to compensate for neuromuscular and tendinous changes through motor control strategies to maintain adequate movement patterns (Cronin et al., 2011).

The skeletal structure has also been shown to have an effect on muscle activation during gait. Specifically related to the triceps surae, heel length or EMA, which is affected by the insertion point of the muscles may influence motor control during gait. One study shows that young subjects with shorter heels walk with MG activation bias. In the same study, subjects with longer heels walked with unbiased activation between the MG and LG muscles. The subjects with shorter heels were also found to have larger MG muscles which may have been an adaptation to counterbalance shorter heel lengths (Ahn et al., 2011). The exact mechanism for this adaptation is unclear. One plausible insight may be to highlight the differences between postural and phasic muscles of the human lower limb. The SOL, primarily comprised of type I muscle fibers, functions as the principal plantarflexion contributor in postural control yet suppression of its muscular activity has been noted in late stance (Sinkjaer, Andersen, & Larsen, 1996). Others have suggested that the contributions of the SOL during gait in elderly remain within early to mid-stance in order facilitate single leg support but SOL contribution during terminal stance may be reduced in these adults (Schmitz, Silder, Heiderscheit, Mahoney, & Thelen, 2009). Therefore, it may be suggested that coordination of phasic plantarflexors in younger individuals, particularly during terminal stance may be a top-down control mechanism to direct reliance on larger muscles within the plantarflexors to facilitate propulsion during gait. Further, this mechanism of lower limb coordination may be altered due to older individuals' ability to compensate for age-related changes within the peripheral nervous system. Permitting examination of responses from postural and phasic muscles of the elderly lower limb in response to lowered leverage around the joint.

Musculoskeletal Mechanics

Muscle moment arms are known for their role in joint mechanics and relationship to muscle force and torque generation about a joint. Specific to the ankle, the PFMA serves as a class I lever connecting the ankle center of rotation and muscle force line of action (Sherman, Seth, & Delp, 2013). Lesser known is the importance of the PFMA in gait and mobility. Previous research shows positive correlations between PFMA and maximal isometric plantarflexion torque. Additionally, stronger correlations were noted between PFMA and torque than were noted for plantarflexor muscle volume and torque at increasing angular velocities (Baxter & Piazza, 2014). This finding suggests that further investigation of PFMA is reasonable as muscle leverage is an equal determinant of muscle strength.

Several studies have investigated the link between PFMA size and locomotor performance. Heel and toe length have been observed to differ between human sprinters and non-sprinters. Sprinters had shorter heels and longer toes, which may have allowed for a higher gear ratio and more propulsive force (Baxter et al., 2012; Lee & Piazza, 2009). Equally, larger moment arms negatively affect joint moment development, power, and work output at higher ankle angular velocities. This effect is due to the non-linear force-velocity relationship in muscle and to the relationship between joint angular velocity and muscle shortening velocity. This relationship is described by the equation 1:

(Equation 1) $$L_{mus} = MA * \Theta$$

Where L_{mus} is the muscle shortening velocity, MA is the moment arm, and Θ is the joint angular velocity. According to this equation, muscle shortening velocity during plantarflexion is the product of the corresponding joint angular velocity multiplied by a factor of moment arm, suggesting that larger muscle shortening velocities are required for larger moment arms (Nagano & Komura, 2003). This is important to consider across the range of human locomotion as it suggests that PFMA may have an inverse effect on performance, which is dependent on the angular velocity of the joint.

Plantarflexor contributions to moment and power are redistributed during elderly gait (DeVita & Hortobagyi, 2000) but little is known about moment arm effects on gait within this population. Previous support shows that reductions in gait velocity are accompanied by decreases in ankle plantarflexor range of motion, moment, and power (Winter et al., 1990), where moment arm may be a likely contributor to these reductions. Strong correlations have been found between the Six-Minute Walk Test (6MWT) and PFMA length for elderly subjects. These findings were limited as this slower group was also found to be older and have a higher body mass index than their comparative group (Lee & Piazza, 2012). Nevertheless, the idea that moment arm is a limiting factor in gait has been confirmed more recently by Rasske & Franz (2018). They concluded that peak ankle moments during the toe-off phase were correlated to smaller PFMAs in older adults. In the same study it was shown that older adults have 11 percent smaller PFMAs during walking when compared to younger subjects (18 to 36 years) which may contribute, in part, to slower gait velocities (Rasske & Franz, 2018). Aging may affect the structure of the triceps surae and ankle joint in two ways described previously. The first is that a loss of muscle mass effectively reduces the PFMA by moving the muscle line of action closer to the ankle (Sugisaki et al., 2010). The second way involves a proximal migration of the insertion point on the calcaneus. Studied in subjects with an age range of 10 to 40 years, the Achilles tendon insertion was shown to drift more superior as aging progressed by year (Kim, Martin, Ballehr, Richey, & Steinberg, 2011).

Summary

There is an increasing occurrence of gait velocity decline in the elderly population. The underlying mechanisms thought to contribute to this have been described, yet individual components remain unclear. Elderly subjects are found to adapt their gait strategies to rely more on the hip and less on the ankle joint during the propulsive phase of gait (DeVita & Hortobagyi, 2000; Judge et al., 1996). Shown recently is the importance of the ankle PFMA in elderly gait (Lee & Piazza, 2012; Rasske & Franz, 2018). Younger subjects also show distinct adaptations to shorter heel lengths where they are able to compensate through motor control and muscular size (Ahn et al., 2011). Considering an elderly population between the ages of 65 and 80 years old, these adults may be less likely or even unable to adapt muscular size in compensation for shorter heel lengths. Within the triceps surae group, this may leave only the neural system to compensate for poorer mechanical advantage leading to reductions in gait velocity seen with aging.

Methods

Subjects

Seventeen elderly adult volunteers between the ages of 65 and 80 were recruited from Appalachian State University and local areas surrounding Boone, NC. All subjects indicated no history of musculoskeletal disorders, lower extremity injury, stroke, heart attack, or rheumatoid arthritis experienced within the past year (Lee & Piazza, 2012). Subjects were brought into the Biomechanics Laboratory on one occasion, which lasted approximately one hour. Upon arrival, they were informed of the benefits and risks of participating in the study and provided their informed consent along with passing an American College of Sports Medicine pre-participation questionnaire. Subjects then completed the Physical Activity Scale for the Elderly (PASE). After initial height, mass, and leg circumference measures were obtained, external pictures of their lower leg were taken for moment arm. Following measurements, subjects were equipped with electromyographic (EMG) sensors on the MG, LG, SOL, and TA of the right leg and the strength of the lower leg was assessed. Initial measurements were followed by the Six-Minute Walk Test. EMG was collected during the lower leg strength test and 6MWT.

Moment Arm Estimates

PFMAs for each subject were estimated using external measures described previously (Scholz et al., 2008; van Werkhoven & Piazza, 2017). Briefly, the PFMA was estimated as the mean perpendicular distance from the lateral (Lmal) and medial (Mmal) malleoli to the most posterior portion of the Achilles tendon (Ach). This measure is achieved using external photographs of the subject’s foot which was placed on a reference box. Subjects were instructed to align their foot with the lateral edge of the block and position their knee above the ankle. Marks were then placed on the most prominent portion of the first metatarsophalangeal joint and malleoli. All images were processed in Matlab (The Mathworks, Inc., Natick, MA). EMA was calculated as the ratio between PFMA and mid-foot length – distance from ankle center of rotation to first metatarsophalangeal joint (Ahn et al., 2011). PFMA and EMA estimated by external measures are illustrated in figure 2.

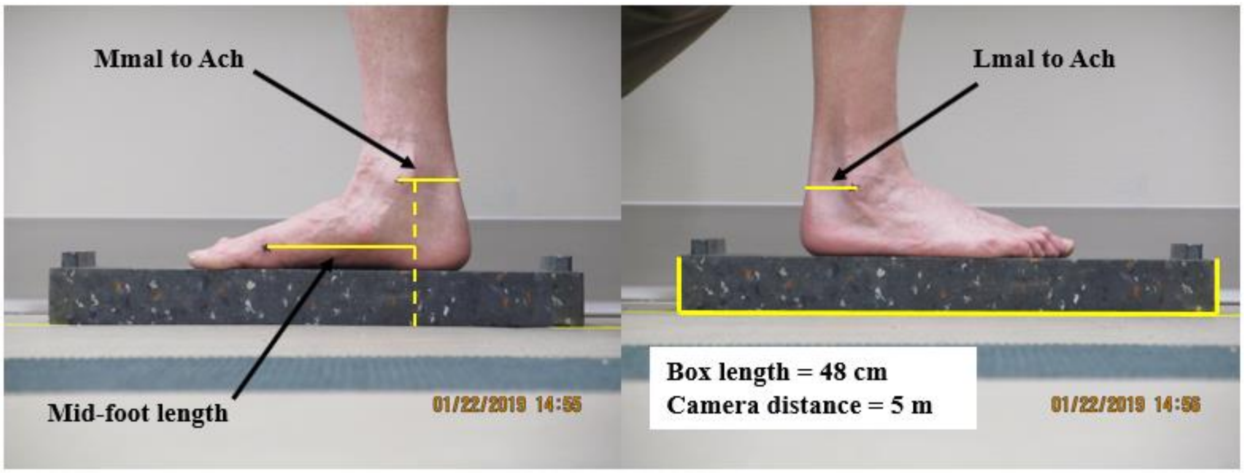

Figure 2. Images of medial (left) and lateral (right) lower leg with foot placement on reference box. PFMA calculated as mean distance from Mmal and Lmal to Ach. Mid-foot length estimated as distance from first metatarsophalangeal joint to estimated ankle joint center. EMA calculated as ratio of PFMA to mid-foot length.

Electromyography

EMG was recorded from the TA, LG, MG, and SOL muscles during strength assessments and the Six Minute Walk tests using wireless sensors (Delsys Trigno Wireless System, Natick, Massachusetts, USA). EMG electrodes with dimension of 27 x 37 x 15 mm with four 5 mm x 1 mm contact points (spaced at 10 mm) in a bipolar configuration were used to collect data at 1000 Hz. Electrodes were placed according to guidelines from SENIAM (seniam.org). Briefly, the TA was first identified by having the subject dorsiflex the foot while pressure was applied toward plantarflexion. Sensors were placed lateral to the anterior border of the tibia at one third the distance between the fibular head and tip of the medial malleolus. The gastrocnemii were identified by having the subject plantarflex from a prone position as pressure was applied toward dorsiflexion. The gastrocnemius heads were identified by observing their separation at the proximal portion of the Achilles tendon. MG sensor placement was on the most prominent bulge of the muscle on the medial side. LG sensor placement was one third the distance between the fibular head and heel. The SOL was identified by having the subject plantarflex while applying pressure toward dorsiflexion. SOL placement was two thirds the distance between the medial condyle of the femur and medial malleolus lateral to the Achilles tendon. The skin was abraded and cleaned with isopropyl alcohol after being shaved of excess hair. Electrodes were attached using sensor adhesives (Trigno™ Sensor Skin Interface, Natick, Massachusetts, USA) to ensure no movement of the sensor during testing conditions. EMG signals were bandpass filtered (20 to 450 Hz) using a zero-phase fourth order Butterworth filter before being full-wave rectified and lowpass filtered (10 Hz) to create a complete linear envelope (Moritz, Greene, & Farley, 2004; Sano et al., 2013). These signals were then normalized to the within trial peak for each muscle.

Lower Leg Strength

Lower leg strength was assessed using a dynamometer (HUMAC NORM, Computer Sports Medicine Inc.; Stoughton, Massachusetts). Subjects were first familiarized with the equipment and reminded of the intended use. Subjects were then placed in a seated position with their hip at 60 degrees flexion, knee fully extended, ankle in a neutral position, and foot securely fastened in the plantarflexion/dorsiflexion input arm adapter. Subjects were instructed to plantarflex at maximum effort for three seconds while receiving verbal encouragement from the tester. One practice rep was given to allow subjects to familiarize themselves with the test. This process was repeated three times with a minute rest between reps to obtain an average maximal voluntary isometric contraction value (MVIC).

The Six Minute Walk Test

The 6MWT incorporates preferred gait velocity while maintaining low expense and time requirements. The test-retest reliability of the 6MWT found previously is high between testing site, gender, and age group (r = 0.95 over one-week period) (Harada et al., 1999). 6MWT results for inactive elderly people have been found to be significantly lower than their active counterparts. These results were also moderately correlated with other measures of functional mobility such as gait speed (r = -0.73), standing balance (r = 0.52), and chair stands (r = 0.67). 6MWT performance also correlated with subjects' perceptions of health and self-efficacy pertaining to physical function (Harada, Chiu, & Stewart, 1999). Other research shows associations between the 6MWT and functional measures. It is reported that both strength and power reductions at the ankle and knee joints are strongly associated with results of the 6MWT (Bean et al., 2002).

The Six-Minute Walk Test was conducted in the Biomechanics Laboratory at Appalachian State University. Previous studies have used a walking surface ranging from 20-50 meters ("ATS Statement," 2002). Guidelines from the American Thoracic Society recommend a straight course layout which was used in this study. Due to space constraints in the facility, the Six-Minute Walk Test was conducted on a 15-meter straight course. Sciurba et al. (2003) found no significant effect of a 50 to 164 (15 to 50 meters) feet straight course layout on 6MWT distances (Sciurba et al., 2003). Subjects were instructed to walk for six minutes around two cones at their preferred walking velocity making 180-degree pivots around each cone and focusing to stay within the one-meter width of the track. The instructions were then demonstrated before the test began. Every meter of the testing surface was marked using tape to calculate 6MWT distance. 6MWT results were obtained by counting the number of laps completed by the subject plus the distance the subject was at when the test was terminated. Total distance over the six minutes was used to determine average gait velocity for the 6MWT. Gait velocity was also collected once every minute using two timing gates set ten meters apart to validate 6MWT results. These values were highly correlated ($r = 0.98$, $p < 0.001$).

EMG samples were collected continuously during the 6MWT. The sensors were synchronized with an accelerometer (Delsys Trigno Wireless System, Natick, Massachusetts, USA) placed on the posterior right heel and ground reaction forces (GRF) from three imbedded force plates (Bertec Corp.; Columbus, Ohio) located in the middle of the laboratory walkway. Gait events were determined for the right leg using accelerometer and GRF vertical magnitudes. Peak EMG amplitudes during stance phases were normalized to their within-trial peak and averaged across the six minutes. Muscle biases were calculated during pre-stance and stance phases. Pre-stance bias was calculated using data from 100 milliseconds before heel-strike. Stance phase bias was calculated from data collected between each heel-strike and toe-off. Muscle bias was calculated using a modified version of the muscle coactivation equation as used by Peterson & Martin (2010) (Equation 2). Briefly, the bias of a muscle is given by dividing the area of EMG for that muscle (m1; MG or Sol) by the sum of the areas of collective EMG signals for m1 and the reference muscle m2 (m2; LG).

(Equation 2) $$Muscle\ Bias = \frac{\int_{FC}^{TO} EMG_{m1}dt}{\sum \int_{FC}^{TO} EMG_{m1}dt + \int_{FC}^{TO} EMG_{m2}dt}$$

Physical Activity Scale for the Elderly

The Physical Activity Scale for the Elderly (PASE) is a simple instrument used to quantify the activity levels of individuals ≥ 65 years of age. Scores are calculated by determining the weight and frequency of 12 activities from individual’s responses. Test-retest reliability of this instrument is 0.75 (95% CI = 0.69 to 0.80). The total score of PASE shows positive associations with the International Physical Activity Questionnaire ($r = 0.742$, $p < .001$), Short Physical Performance Battery ($r = 0.622$, $p < .001$), Short Form-36 Quality of Life Questionnaire ($r = 0.432$, $p < .001$) (Ayvat, Kilinç, & Kirdi, 2017), and monthly temperature ($r = 0.83$) (Washburn, Smith, Jette, & Janney, 1993).

Statistical Analyses

Associations among variables were investigated using correlational analysis. In line with study hypotheses, correlations were assessed between gait velocity, PFMA, and muscle bias indices. Assumptions of normality were checked using the Shapiro-Wilk test and visual inspection of quantile-quantile plots. Univariate outliers were checked using box-plots. Criteria for removal required violation of the interquartile rule for outliers. Assumptions of homoscedasticity were examined through inspection of scatterplots. Effect sizes were categorized using Pearson correlation coefficients where $r \leq 0.3$ indicated no association, 0.3 to 0.5 indicated low association, 0.5 to 0.7 indicated moderate association, and $r \geq 0.7$ indicated high association (Maher, Markey, & Ebert-May, 2013; Mukaka, 2012). *A priori* level of significance was set to 0.05. All statistical tests were done using R language (RStudio, Inc., Boston, Massachusetts).

Results

Seventeen subjects completed the study. One subject was removed due to noise in MG EMG signals. Subject characteristics are summarized in Table 1 ($n = 16$, 8 Women/8 Men). There was no association found between gait velocity and PFMA ($r = -0.13$, $p = 0.63$) (Figure 3) but a low positive correlation was found between EMA and MG-LG bias during the stance phase ($r = 0.42$, $p = 0.11$) (Figure 4).

Table 1. Summary of Subject Descriptive Statistics (mean ± st. dev.)

	mean	st. dev	min.	max.
Height (m)	1.68	0.11	1.51	1.82
Mass (kg)	70.85	18.53	47.1	105.6
BMI (kg/m^2)	24.73	4.02	18.06	32.23
Age (yr)	69.81	4.72	65	79
PASE	157.88	59.89	64	280
Gait Velocity (m/s)	1.27	0.14	0.99	1.46
PFMA (mm)	47.1	4.7	38.5	53.8
EMA	0.39	0.05	0.3	0.51
Peak Torque (Nm)	84	40	28	174
Muscle Bias:				
Stance				
MG-LG	0.53	0.11	0.3	0.7
Sol-LG	0.56	0.11	0.35	0.71
Pre-stance				
MG-LG	0.52	0.15	0.21	0.77
Sol-LG	0.53	0.14	0.33	0.79

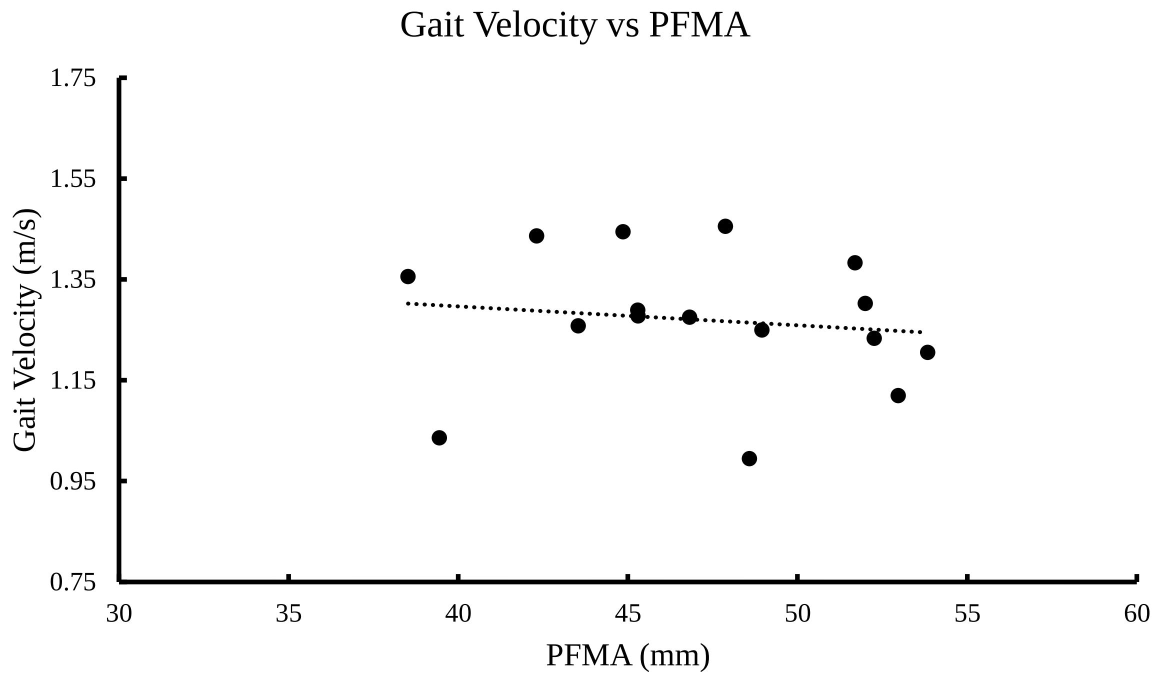

Figure 3. Gait velocity versus PFMA. Gait velocity was not associated and did not significantly correlate with PFMA ($r = -0.13$, $p = 0.63$)

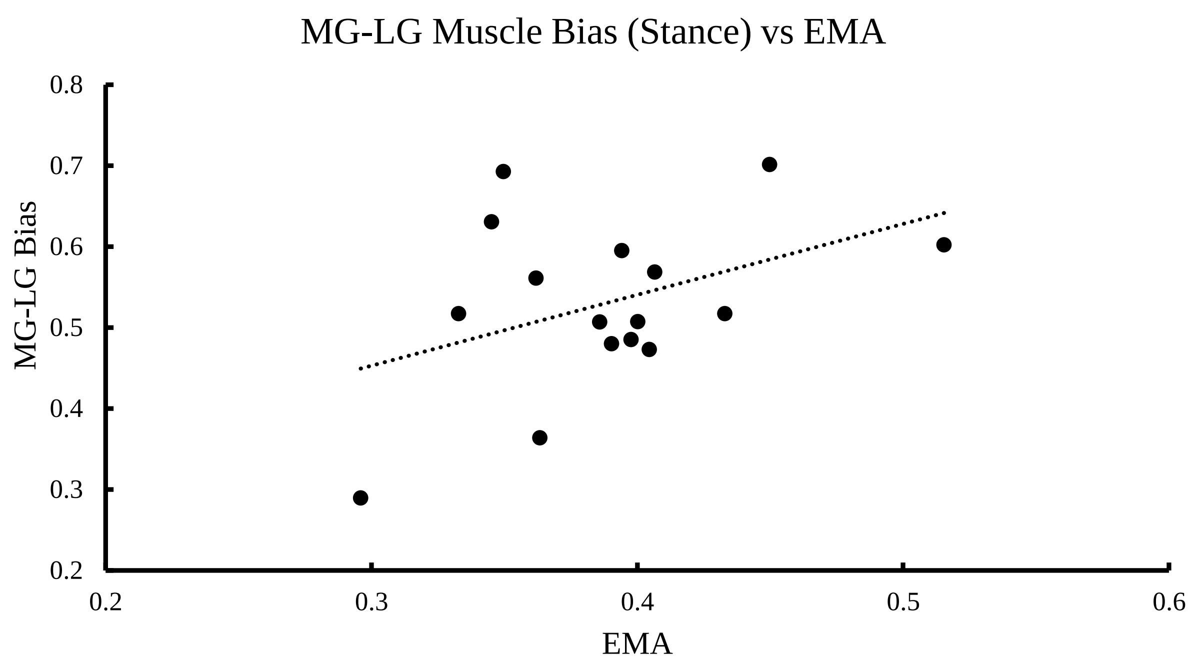

Figure 4. MG-LG muscle bias versus effective mechanical advantage during stance. There was a low positive correlation between MG-LG muscle bias and effective mechanical advantage ($r = 0.42$, $p = 0.11$).

Further analyses were conducted to assess the associations between gait velocity and muscle biases during pre-stance and stance phases. Low positive correlations were found between gait velocity and stance phase MG-LG bias ($r = 0.42$, $p = 0.11$) (Figure 5), stance phase Sol-LG bias ($r = 0.39$, $p = 0.14$) (Figure 6), and pre-stance MG-LG bias ($r = 0.44$, $p = 0.09$) (Figure 7). A moderate (significant) correlation was found between gait velocity and pre-stance Sol-LG bias ($r = 0.54$, $p = 0.03$) (Figure 8). All correlations between variables of interest are summarized in Table 2.

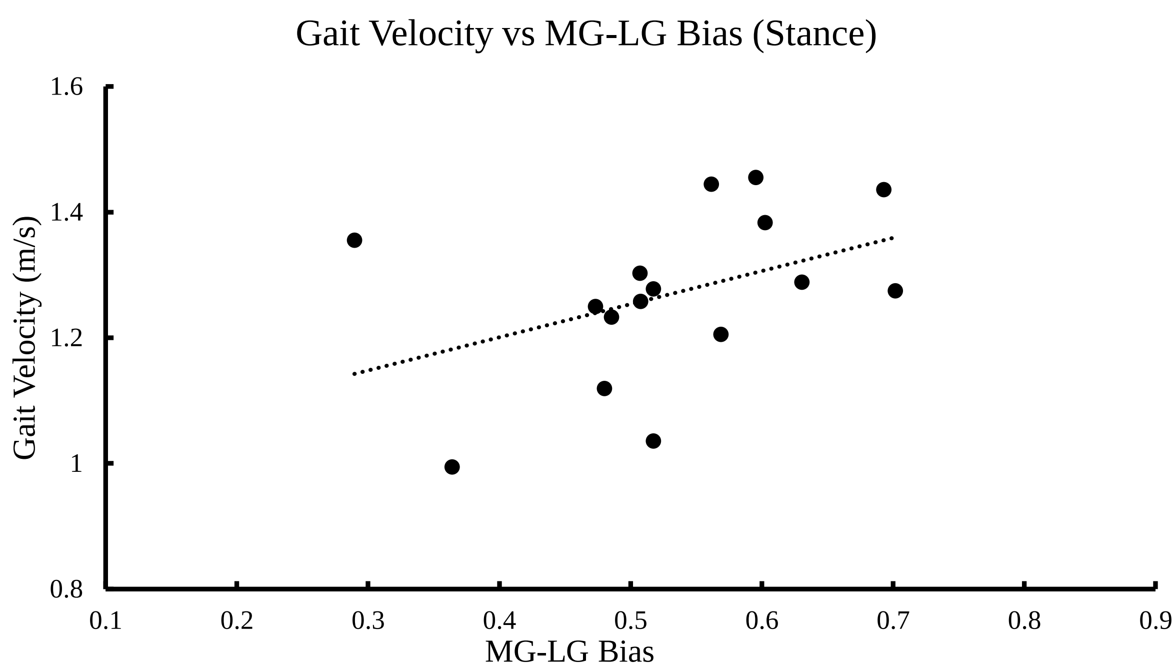

Figure 5. Gait velocity versus MG-LG muscle bias during the stance phase. There was a low positive correlation between gait velocity and MG-LG muscle bias during the stance phase ($r = 0.42$, $p = 0.11$).

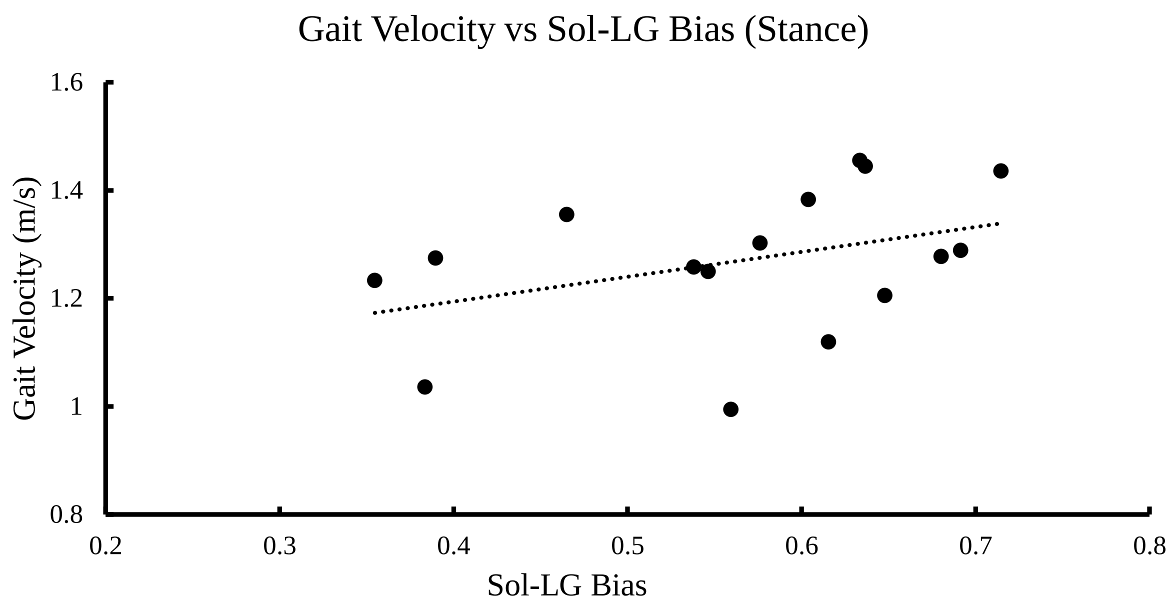

Figure 6. Gait velocity versus Sol-LG bias during stance. There was a low positive correlation between gait velocity and SOL bias during the stance phase ($r = 0.39$, $p = 0.14$).

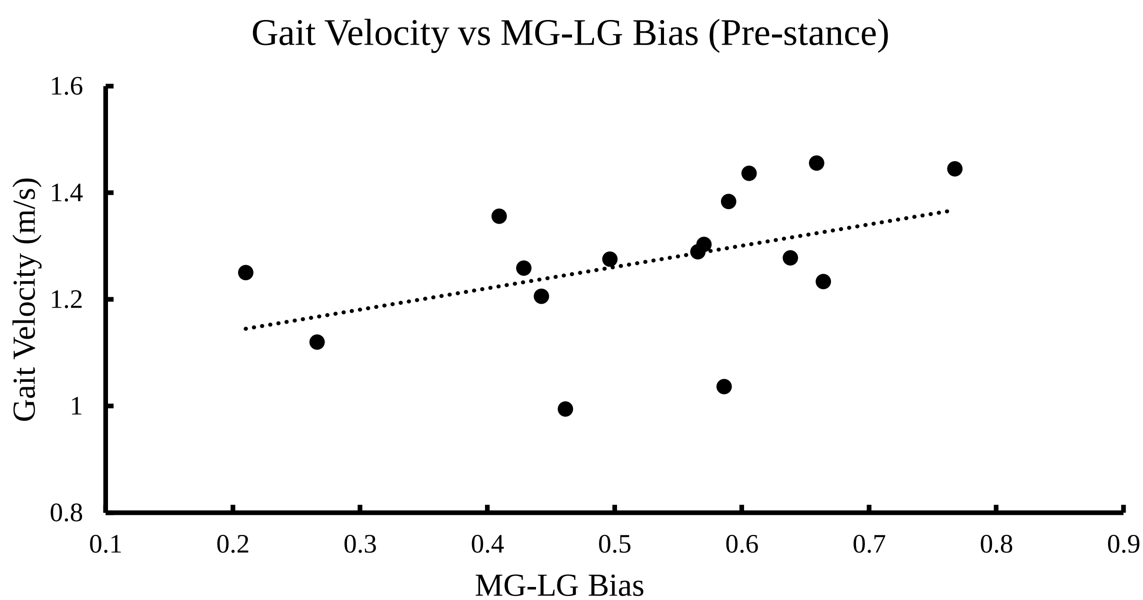

Figure 7. Gait velocity versus MG-LG bias during pre-stance. There was a low positive correlation between gait velocity and MG bias during pre-stance ($r = 0.44$, $p = 0.09$).

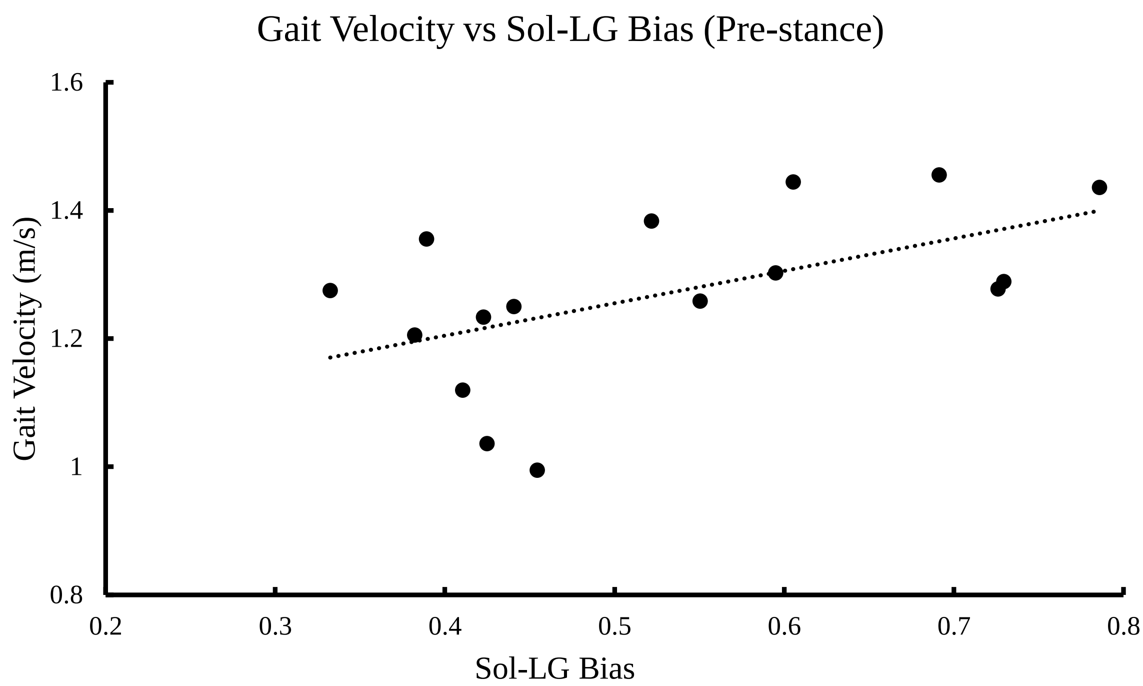

Figure 8. Gait velocity versus Sol-LG bias during pre-stance. There was a moderate (significant) positive correlation between gait velocity and SOL bias during pre-stance (r = 0.54, p = 0.03).

Table 2. Correlations between variables. Expressed as coefficient with p-value in parentheses (* indicates p < 0.05, ** indicates p < 0.01). TG means velocity was measured by timing gate.

	Velocity (6MWT)	Velocity (TG)	PFMA	EMA
Velocity (6MWT)	1.00		-0.13 (0.63)	0.13 (0.62)
Velocity (TG)	0.98 (< 0.001**)	1.00	-0.24 (0.37)	0.04 (0.88)
MG-LG (pre)	0.44 (0.09)	0.46 (0.07)	-0.17 (0.54)	0.03 (0.93)
Sol-LG (pre)	0.54 (0.03*)	0.56 (0.02*)	-0.22 (0.42)	-0.08 (0.78)
MG-LG (stance)	0.42 (0.11)	0.3 (0.26)	0.11 (0.68)	0.42 (0.11)
Sol-LG (stance)	0.39 (0.14)	0.33 (0.21)	0.14 (0.62)	0.06 (0.83)

Discussion

Gait Velocity and Moment Arm

This study did not find an association between gait velocity and PFMA in an elderly sample. This is contrary to the first hypothesis and differs from previous literature that found a strong and significant relationship ($r = 0.82$, $p = 0.004$) between the two (Lee & Piazza, 2012). Lee & Piazza (2012) only found the relationship after applying *post-hoc* clustering techniques to 6MWT velocities. Further, when assessing differences between groups, the authors concluded that the relationship occurred in the slower group where subjects were found to be older ($p = 0.034$) and had a higher body mass ($p = 0.021$) (Lee & Piazza, 2012).

Differences in subject characteristics between these two studies may aid in explaining the non-association observed presently. First, the age of subjects in the present study was 69.81 ± 4.72 years which is observably less than the slower group (77.7 ± 6.0 years) studied by Lee & Piazza (2012) given by one-sample t-test ($p < 0.001$). BMI was also observed to be lower in the present study (24.73 ± 4.02 kg/m^2) when compared to values from Lee & Piazza (29.81 ± 3.69 kg/m^2) given by one-sample t-test ($p < 0.001$). Since subject age and BMI was different between studies, the association of gait velocity and moment arm may have been limited by a failure to capture a more heterogenous sample of elderly aged 65 to 80 years with a wider range of BMI. Second, the method of moment arm measurement differed between studies. *In vivo* methods to measure moment arm at the ankle are MRI (Center of Rotation, Finite Helical Angle, Instantaneous Helical Angle), tendon excursion (TE), hybrid ultrasound and motion capture, and external measures. Due to time and equipment limitations, the present study estimated moment arm using external methods described by Scholz et al. (2008) and van Werkhoven & Piazza (2017). Estimates obtained from this study were 47.1 ± 4.7 mm which are in range of Scholz et al. (2008) (48.5 ± 3.6 mm) and van Werkhoven & Piazza (2017) (44.0 ± 6.0 mm) using this external method. Previously, moment arm size using MRI methods has been reported by Fath, Blazevich, Waugh, Miller, & Korff (2010) (51.7 ± 4.3 mm, neutral ankle), Hashizume et al. (2012) (41 ± 5.9 mm; 49 ± 4.0 mm, neutral ankle), and Maganaris, Baltzopoulos, & Sargeant (2000) (44 ± 3.0 mm to 55 ± 3.0 mm, -15 to 30 degrees). Although the estimates from these studies were primarily observed in younger populations, the estimates of the elderly sample in the current study are well within ranges observed previously. Estimates of the TE method used by Lee & Piazza (2012) were acknowledged as being unexpectedly small (32.0 ± 3.2 mm). Nevertheless, when compared to the present study, variation in the estimates obtained are relatively similar. Therefore, differences in observed relationships between studies were unlikely influenced by differences in methods of moment arm measurement.

Third, the potential differences between levels of physical activity between these studies should be mentioned. The cohort of this study scored 157.88 ± 59.89, measured by the Physical Activity Scale for the Elderly (PASE). Normal values of the PASE for men are 144.3 ± 58.6 (age = 65 to 69 years) and 102.4 ± 53.7 (age = 70 to 75 years) while women range from 112.7 ± 64.2 (age = 65 to 69 years) and 89.1 ± 55.5 (age = 70 to 75 years) (Washburn et al., 1993). One-sample t-tests indicated that present study mean values for both genders were similar to values reported previously for ages 65 to 69 years and higher than values reported for 70 to 75 years (Men: $p < 0.01$, Women: $p < 0.01$). Lee & Piazza (2012) did not report a specific tool to measure physical activity levels, however, they stated that their subjects ranged from sedentary to very physically active. Population-based research in adults older than 60 has indicated physical activity as a protection factor against gait decline (Busch et al., 2015). Since subject age differed between studies it may be plausible to suspect differences in physical activity. Yet, without methods to compare activity levels across studies, differences in gait velocity and PFMA relationships observed between studies cannot be explained at this point.

Previous work on the topic of moment arm in lower limb mechanics and gait has been primarily focused in younger runners and sprinters (Baxter et al., 2012; Karamanidis et al., 2011; Lee & Piazza, 2009; Raichlen et al., 2011; Scholz et al., 2008). Shorter moment arms are strongly correlated with lower rates of energy consumption during running (Raichlen et al., 2011; Scholz et al., 2008) but moment arm size was not associated with walking economy (Raichlen et al., 2011). Human sprinters exhibit smaller PFMAs than non-sprinters (Baxter et al., 2012; Lee & Piazza, 2009) by a magnitude of up to 25 percent (Lee & Piazza, 2009). These studies provide evidence of the PFMA role during intensive locomotor tasks where the proposed characteristics improve variable gearing of the ankle, and consequentially, optimize pre-stretch and shortening velocity of the plantarflexor muscles (Carrier, Heglund, & Earls, 1994). Ankle joint leverage has also been studied in clinical populations. Patients with moderate/severe diabetic peripheral neuropathy (DPN) exhibited significantly larger EMA than controls when walking at 0.6, 1.0, 1.2, 1.4, and 1.6 m/s. Diabetic patients without peripheral neuropathy also exhibited significantly larger EMA than controls while walking at speeds of 0.6, 0.8, and 1.4 m/s (Petrovic et al., 2017). Causal explanations of increased EMA may largely be due to smaller external moment arms within these patients. Diabetic patients, and those with DPN, may modify distance of applied ground reaction force to reduce the force requirements from the plantarflexors to achieve a walking task. This, as the authors propose, may have a causal relationship with increased cost of walking observed previously in diabetic populations through mechanisms of inadequate utilization of tendon elastic properties (Petrovic et al., 2017).

Evidence of the PFMA role in performance and clinical populations suggests that exposure of its significance may be largely dependent on the intensity of the locomotor task as well as the abilities of the population in question. Nevertheless, the idea that the leverage around the ankle may be constraining in clinical populations remains present. Furthermore, this constraint proves difficult to identify due to the redundancy of the musculoskeletal and nervous system to adapt movement patterns. Insignificance of the PFMA in the present study may have been largely task-intensity dependent. Considering the physical activity of the cohort studied, the intensity of a six-minute walking test at a preferred pace was likely inadequate in challenging functional capacity. Additionally, the proposed age of major gait decline in elderly women was found to be 71 years (Kirkwood et al., 2018), whereas the average age of subjects in the present study lies below that point. Similar to non-associations between moment arm and walking economy in young (Raichlen et al., 2011), preferred gait velocity in this cohort was highly conserved and influences of the PFMA were not evident.

EMA and Muscle Activation

The present study found a low positive correlation between EMA and MG-LG bias during the stance phase meaning that bias values generally increased with EMA, although this result was non-significant ($r = 0.42$, $p = 0.11$). This was contrary to the second hypothesis and differed from previous reports in sedentary young where a strong negative relationship was found ($r = -0.71$, $p = 0.01$) (Ahn et al., 2011). In comparisons between studies the EMA of the present study was 0.39 ± 0.05 whereas Ahn et al. (2011) found values of 0.26 ± 0.02 for the biased group (MG-LG bias > 0.67) and 0.32 ± 0.07 for the unbiased group (MG-LG bias < 0.67). MG-LG stance bias for the present study was 0.53 ± 0.11. Previous values from the biased and unbiased walkers were 0.74 ± 0.06 (biased) and 0.57 ± 0.06 (Ahn et al., 2011).

Similar to the present study, Ahn et al. (2011) estimated the PFMA externally by measuring distances from the medial and lateral malleoli to the posterior surface of the calcaneal tendon. Mid-foot length was estimated by averaging the distances measured from the lateral malleolus to the fifth metatarsophalangeal joint and medial malleolus to the first metatarsophalangeal joint (Ahn et al., 2011). Subtle differences are noted with the present study. Mid-foot length was estimated as the horizontal distance between the ankle center of rotation and first metatarsophalangeal joint and therefore may be overestimated compared to measures by Ahn et al. (2011). This is counterintuitive that the present study would yield larger EMAs as it is assumed that mid-foot lengths would generally be larger and therefore the ratio of internal to external moment arm would be smaller for this elderly group. This can be explained by PFMA size differences between studies that resulted from differences in measurement methods. Internal moment arm lengths for biased (33.1 ± 2.6 mm) and unbiased (41.4 ± 9.2 mm) walkers were generally smaller than those observed in the present study (47.1 ± 4.7 mm) which may explain EMA differences. The present study measured muscle bias during a preferred walking velocity while Ahn et al. (2011) measured bias across a range of velocities from 0.2 to 1.5 m/s. Qualitative assessment of the effect of velocity on muscle bias suggests that bias may converge as velocity increases from 0.2 to 1.5 m/s in younger individuals. Given the velocity results of the present study this may suggest that the insignificance of EMA and muscle bias may have been due to the use of a preferred gait velocity task.

The largest disparity between the present study and that of Ahn et al. (2011) is the difference in age. Elderly with smaller moment arms were expected to rely more on the MG muscle during a walking task. The results of the present study indicate that this assumption was inaccurate. The low, but insignificant correlation, suggested a positive association between EMA and muscle bias meaning elderly may activate muscles of the lower leg differently in response to varying mechanical advantage at the ankle. Further, the bias ranges for this elderly group were lower than young biased walkers and somewhat closer to those of the unbiased group. Sedentary young individuals with poorer leverage around the ankle are thought to rely on relative thicker muscles to achieve a walking task. Concomitant with increased bias, these individuals also exhibited adapted muscle size within their MG which was posited to be an outcome of larger force requirements due to lower EMA (Ahn et al., 2011). Mechanisms of contractile strength have been studied in younger individuals' responses to heavy resistance training. Fourteen weeks of resistance training increased individual muscle fiber area and pennation angle in the vastus lateralis. These increases lead to overall increases in physiological cross sectional area, thus improving force-generating capacity (Aagaard et al., 2001). Therefore, activation differences seen by Ahn et al. (2011) may have been adaptations of younger individuals to rely on the MG where structural adaptations were increased relative to other members of the triceps surae in response to reduced leverage around the ankle. It is a limitation of this study that muscle sizes were not quantified. Elderly subjects, prone to sarcopenia, might not be able to maintain adapted muscular size, similar to young, and might therefore adapt neuromuscular patterns that may facilitate gait velocity. This might explain the positive correlation observed between EMA and muscle bias which was opposite to findings by Ahn et al. (2011). To further explain our results, the present study also considered the influence of other potential triceps surae biases that may occur in elderly gait. Present findings suggest trends of low positive correlations between gait velocity and stance phase MG-LG bias ($r = 0.42$, $p = 0.11$) and Sol-LG bias ($r = 0.39$, $p = 0.14$). In pre-stance (100 ms before heelstrike), there was a moderate positive correlation between gait velocity and Sol-LG bias (r = 0.54, p = 0.03) and low positive correlation with MG-LG bias (r = 0.44, p = 0.09).

The present study shows an association that may indicate muscle activation patterns that positively influence gait velocity. Higher Sol-LG in pre-stance was significantly associated with faster gait velocities but the mechanisms are unclear. One possible explanation for increased pre-stance SOL activity is to facilitate single leg support through mid-stance. In younger individuals SOL activity induced by electrical stimulation resulted in increased vertical support during mid-stance (Francis, Lenz, Lenhart, & Thelen, 2013). It is possible that increased pre-stance Sol-LG bias in elderly may be an effort to stiffen the ankle joint or to retain this biomechanical role similar to younger groups (Bailey et al., 2018). The present study did not partition the EMG signals to consider biases during loading, mid-stance, or terminal stance. Therefore, specific muscle biases may have occurred at different points in the stance phase. Subjectively, this association between gait velocity and increased Sol-LG bias observed in the present study may be a similar neuromuscular adaptation for older adults to maintain gait velocity through increased support in mid-stance (Schmitz et al., 2009).

Since the strength of associations failed to show significance in all but one comparison, it is also possible that aging may mitigate muscle biases. Ahn et al. (2011) considered subjects to be biased if their MG-LG bias ratio exceeded 0.67. Bias results of the present study lie within the ranges of unbiased walkers (bias = 0.34 to 0.66), therefore may indicate adapted muscle activation patterns may not be sustained with age. A possible explanation of this may be entwined within the coordination dynamics in and between brain, muscular, and behavior levels of the neuro-musculo-skeletal-system (Sleimen-Malkoun, Temprado, & Hong, 2014). Particularly, the observed difference between muscle activation patterns of young and older adults may be due to dedifferentiation caused by aging where functional roles of muscle (i.e. changes in fiber composition) or neural subsystems become more simplified and less specialized to movement tasks (Sleimen-Malkoun et al., 2014). Further support is derived from the hypothesis that aging causes a loss of complex variability which reduces the dimensionality of possible responses to physiological stresses (Lipsitz & Goldberger, 1992). Suggesting that as complexity of the system is reduced, behavior becomes predictable, and a person loses functionality – a decline from young adults to healthy elderly to frailty (Lipsitz, 2002; Sleimen-Malkoun et al., 2014). Therefore, it is possible that results of the present study may differ based upon linkages of dedifferentiation and loss of complexity hypotheses within a physically active elderly group.

Conclusions

The present study did not find similar relationships among PFMA, muscle activation patterns, and gait velocity reported previously. This may have been limited by its failure to capture a heterogenous sample of elderly individuals. Further, the methodology may have suffered from an oversimplification of testing which likely did not successfully challenge functional capacity within this group. Regardless, these inconsistencies provide new indications which may provide a basis for future investigation of mobility constraints in elderly.

The present study cannot negate the possible influence of PFMA as a limiting factor within elderly gait. The absence of association between PFMA and gait velocity in the present study may have been affected by the homogeneity of age and level of physical activity of its subjects, suggesting that constraints of walking at a preferred pace may be masked within this group. This is significant as previous work has proposed the threshold for gait velocity decline to be at 71 years (Kirkwood et al., 2018) along with the importance of physical activity in preventing this loss (Busch et al., 2015). The results of the present study are not consistent with muscle biases observed within younger individuals who exhibit lower EMA at the ankle. However, due to preliminary evidence of MG-LG and Sol-LG biases during gait, the existence or non-existence of adapted motor patterns in elderly may promote further investigation. Future research should seek to clarify the influences of ankle structure and motor patterns on mobility within this population. Given by previous research in lower limb mechanics, the role of the ankle structure may be largely dependent on the intensity of the locomotor task and abilities of the population in question. Furthermore, due to the adaptability of the human musculoskeletal and nervous systems, more intensive methods and multivariate approaches may be necessary to progress the understanding of mobility losses due to aging in order to improve early detection and intervention strategies. To do this, a large body of research suggests impairment of muscular and neural systems may precede the age group considered within the present study (Bowen et al., 2015; Kim & Choi, 2013; Manini et al., 2013; Pannese, 2011), indicating that future work should seek to understand how interactions between muscular, neural, and structural systems are altered over a larger time course.

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Appendix A: PASE - Physical Activity Scale for the Elderly

Instruction

Please complete this questionnaire by either circling the correct response or filling in the blank. Here is an example:

During the past 7 days, how often have you seen the sun?

[0.] NEVER

[1.] SELDOM (1-2 DAYS)

[2.] SOMETIMES (3-4 DAYS)

[3.] OFTEN (5-7 DAYS)

Answer all items as accurately as possible. All information is strictly confidential.

Leisure time activity

1. Over the past 7 days, how often did you participate in sitting activities such as reading, watching TV, or doing handcrafts?

[0.] NEVER (go to question 2)

[1.] SELDOM (1-2 DAYS) (go to question 1.a and 1.b)

[2.] SOMETIMES (3-4 DAYS) (go to question 1.a and 1.b)

[3.] OFTEN (5-7 DAYS) (go to question 1.a and 1.b)

1.a **What were these activities? (open end question)**

1.b **On average, how many hours did you engage in these sitting activities?**

[0.] Less than 1 hour

[1.] 1 but less than 2 hours

[2.] 2 - 4 hours

[3.] more than 4 hours

2. Over the past 7 days, how often did you take a walk outside your home or yard for any reason? For example, for fun or exercise, walking to work, walking the dog, etc

[0.] NEVER (go to question 3)

[1.] SELDOM (1-2 DAYS) (go to question 2.a)

[2.] SOMETIMES (3-4 DAYS) (go to question 2.a)

[3.] OFTEN (5-7 DAYS) (go to question 2.a)

2a. **On average, how many hours per day did you spend walking?**

[0.] Less than 1 hour

[1.] 1 but less than 2 hours

[2.] 2 - 4 hours

[3.] more than 4 hours

3. Over the past 7 days, how often did you engage in light sport or recreational activities such as bowling, golf with a cart, shuffleboard, fishing from a boat or pier or other similar activities?

[0.] NEVER (go to question 4)

[1.] SELDOM (1-2 DAYS) (go to question 3.a and 3.b)

[2.] SOMETIMES (3-4 DAYS) (go to question 3.a and 3.b)

[3.] OFTEN (5-7 DAYS) (go to question 3.a and 3.b)

3.a **What were these activities? (open end question)**

3.b **On average, how many hours did you engage in these light sport or recreational activities?**

[0.] Less than 1 hour

[1.] 1 but less than 2 hours

[2.] 2 - 4 hours

[3.] more than 4 hours

4. Over the past 7 days, how often did you engage in moderate sport and recreational activities such as doubles tennis, ballroom dancing, hunting, ice skating, golf without a cart, softball or other similar activities?

[0.] NEVER (go to question 5)

[1.] SELDOM (1-2 DAYS) (go to question 4.a and 4.b)

[2.] SOMETIMES (3-4 DAYS) (go to question 4.a and 4.b)

[3.] OFTEN (5-7 DAYS) (go to question 4.a and 4.b)

***4.a* What were these activities? (open end question)**

***4.b* On average, how many hours did you engage in these moderate sport or recreational activities?**

[0.] Less than 1 hour

[1.] 1 but less than 2 hours

[2.] 2 - 4 hours

[3.] more than 4 hours

5. Over the past 7 days, how often did you engage in strenuous sport and recreational activities such as jogging, swimming, cycling, singles tennis, aerobic dance, skiing (downhill or cross-country) or other similar activities?

[0.] NEVER (go to question 6)

[1.] SELDOM (1-2 DAYS) (go to question 5.a and 5.b)

[2.] SOMETIMES (3-4 DAYS) (go to question 5.a and 5.b)

[3.] OFTEN (5-7 DAYS) (go to question 5.a and 5.b)

***5.a* What were these activities? (open end question)**

***5.b* On average, how many hours did you engage in these strenuous sport or recreational activities?**

[0.] Less than 1 hour

[1.] 1 but less than 2 hours

[2.] 2 - 4 hours

[3.] more than 4 hours

6. Over the past 7 days, how often did you do any exercises specifically to increase muscle strength and endurance, such as lifting weights or pushups, etc.?

[0.] NEVER (go to question 7)

[1.] SELDOM (1-2 DAYS) (go to question 6.a and 6.b)

[2.] SOMETIMES (3-4 DAYS) (go to question 6.a and 6.b)

[3.] OFTEN (5-7 DAYS) (go to question 6.a and 6.b)

***6.a* What were these activities? (open end question)**

***6.b* On average, how many hours did you engage in these strenuous sport or recreational activities?**

[0.] Less than 1 hour

[1.] 1 but less than 2 hours

[2.] 2 - 4 hours

[3.] more than 4 hours

Household activity

7. During the past 7 days, have you done any light housework, such as dusting or washing dishes?

[1.] NO
[2.] YES

8. During the past 7 days, have you done any heavy housework or chores, such as vacuuming, scrubbing floors, washing windows, or carrying wood?

[1.] NO
[2.] YES

During the past 7 days, did you engage in any of the following activities? Please answer YES or NO for each item.

a. Home repairs like painting, wallpapering, electrical work, etc.

b. Lawn work or yard care, including snow or leaf removal, wood chopping, etc.

c. Outdoor gardening

d. Caring for another person, such as children, dependent spouse, or another adult

Work-related activity

10. During the past 7 days, did you work for pay or as a volunteer?

[1.] NO
[2.] YES (go to questions 10.a and 10.b)

***10a.* How many hours per week did you work for pay and or as a volunteer? ____ hours**

***10b.* Which of the following categories best describes the amount of physical activity required on your job and or volunteer work?**

1. Mainly sitting with some slight arm movement (Examples: office worker, watchmaker, seated assembly line worker, bus driver, etc.)
2. Sitting or standing with some walking (Examples: cashier, general office worker, light tool and machinery worker)
3. Walking with some handling of materials generally weighing less than 50 pounds (Examples: mailman, waiter/waitress, construction worker, heavy tool and machinery worker)
4. Walking and heavy manual work often requiring handling of materials weighting over 50 pounds (Ex: lumberjack, stone mason, farm or general laborer)

Vita

Zachary Ripic is originally from Towanda, Pennsylvania, is the son of Andrew and Shelli Ripic, and brother to Alexis Ripic. He graduated from Towanda Area Jr./Sr. High School in June 2012. He attended Pennsylvania State University from 2012 to 2016 and graduated with a Bachelor of Science in Kinesiology with a Minor in Business. Zack attended Appalachian State University from August 2017 to May 2019 where he earned a Master's of Science in Exercise Science. While at Appalachian State he worked as a research assistant and graduate research associate in the Injury Neuromechanics Laboratory. Zack plans to transition to a PhD program in the coming years to continue his research using biomechanical, neuromechanical, and statistical methods to study movement variability and deficits in human gait and balance disorders.