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Net ankle quasi-stiffness is influenced by walking speed but not age for older adult women

John D. Collins[a,b], Elisa S. Arch[a,c], Jeremy R. Crenshaw[a,c], Kathie A. Bernhardt[d], Sundeep Khosla[e], Shreyasee Amin[f,g], Kenton R. Kaufman[d,*]

[a] Biomechanics and Movement Science Interdisciplinary Program, University of Delaware, Newark, DE, USA
[b] BADER Consortium, Newark, DE, USA
[c] Department of Kinesiology and Applied Physiology, University of Delaware, Newark, DE, USA
[d] Department of Orthopedic Surgery, Mayo Clinic, Rochester, MN, USA
[e] Division of Endocrinology, Diabetes, Metabolism, and Nutrition, Department of Medicine, Mayo Clinic, Rochester, MN, USA
[f] Division of Epidemiology, Department of Health Sciences Research, Mayo Clinic, Rochester, MN, USA
[g] Division of Rheumatology, Department of Medicine, Mayo Clinic, Rochester, MN, USA

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ABSTRACT

Background: Insufficient plantar flexor resistance due to plantar flexor weakness, an impairment common in patient populations, causes substantial gait deficits. The bending stiffness of passive-dynamic ankle-foot orthoses (PD-AFOs) has the capacity to replace lost plantar flexor resistance. Many patients who are prescribed PD-AFOs are older adults. While PD-AFO bending stiffness should be customized for patients, a method to objectively prescribe this stiffness does not exist. Quantifying natural plantar flexor resistance during non-pathological gait could provide a reference value for objectively prescribing PD-AFO bending stiffness.

Research question: This study investigated the effect of age on plantar flexor resistance in 113 participants above the age of 65 years. We did so while also considering the confounding influence of gait speed, an aspect known to be reduced with old age.

Methods: Ambulatory, community-dwelling older adult women (ages 65–91 years) with no current or recent lower-extremity injuries or surgeries underwent an instrumented gait analysis at a self-selected speed. Plantar flexor resistance was quantified via net ankle quasi-stiffness (NAS) defined as the slope of ankle joint moment-angle curve during late stance.

Results: showed that NAS was not significantly influenced by age ($r = -0.11$, p = 0.12), and that the confounding factor of walking speed had a significant, positive relationship with NAS ($r = 0.59$, $p < 0.001$).

Significance: By determining that gait speed, not age, is related to NAS in older adults, this study represents the initial step towards objectively prescribing PD-AFO bending stiffness to achieve a targeted gait speed for older adults with plantar flexor weakness.

1. Introduction

During the period from the first instance of the plantarflexion moment to the maximum dorsiflexion angle, the plantar flexors act eccentrically to control the rate and timing of the ankle's dorsiflexion [1–4]. This plantar flexor function is important for controlling the shank's rotation as well as influencing the kinetics, energetics, propulsion, support, and forward progression of gait [2–5]. Insufficient plantar flexor resistance due to plantar flexor weakness, an impairment common in patient populations including post-stroke [6], is indicative of reductions in the dynamic function of the joint. This impairment typically presents as one of two extremes, excessive dorsiflexion [1] which may present with concurrent knee flexion (or "crouch gait") excessive plantarflexion with knee hyperextension [7] (commonly referred to as a "recurvatum knee gait") during stance. These patterns result in detrimental gait dysfunctions including shorter or asymmetrical step lengths and decreased walking speeds [3,6]. Such decreases in gait function, in particular slower walking speed, have been linked to reductions in overall function and quality of life [8].

Passive-dynamic ankle-foot orthoses (PD-AFOs) are often given to patients with plantar flexor weakness with the intent of using the spring-like bending stiffness of the orthosis to replace a patient's lost

* Corresponding author at: Motion Analysis Laboratory, Division of Orthopedic Research, Charlton North L-110L, Mayo Clinic, Rochester, MN, 55905, USA.
E-mail addresses: kaufman.kenton@mayo.edu, Borglum.barbara@mayo.edu (K.R. Kaufman).

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plantar flexor resistance [9,10]. The PD-AFO bending stiffness should be customized for patients to achieve optimal patient outcomes [11] by properly restoring the joint's dynamic function. Too little bending stiffness will not provide enough resistance to control the shank's rotation, while too much stiffness will provide insurmountable resistance that prevents proper shank rotation. Thus, if the level of bending stiffness is not customized based on each patient's needs, gait deficits will likely persist or could be exacerbated. However, PD-AFO bending stiffness is traditionally provided through a qualitative, craft-based process with limited prescription guidelines [11], and therefore often requires an inefficient trial-and-error process to achieve a suitably-customized PD-AFO. Furthermore, evidence suggests that these traditional orthoses often do not enable patients to achieve optimal function, including improved walking speed and energy cost of walking [11,12].

Technological advancements now enable PD-AFOs to be rapidly and objectively customized and manufactured [13,14], which could overcome the aforementioned limitations of the traditional orthosis fabrication process. However, in order to utilize these technologies, an objective PD-AFO prescription model must be established. Quantifying natural plantar flexor resistance during non-pathological gait could provide a baseline reference value that serves as the basis for an objective PD-AFO bending stiffness prescription model.

Plantar flexor resistance can be quantified via net ankle quasi-stiffness (NAS), which is defined as the slope of the ankle joint moment-angle curve during late stance [9]. NAS has also been termed dynamic ankle stiffness [9] or quasi-stiffness [15]. NAS is a biomechanical measure of the resistance the joint provides when an angular displacement occurs during walking [9]. Thus, NAS is a measure of the spring-like behavior of the ankle joint during walking [16] and is a function of all active and passive structures that cross the ankle joint.

Researchers have previously characterized NAS for various populations including healthy children [9], healthy adults [10,15–19], and several patient groups [9,19,20]. NAS has been shown to increase with walking speed [10,17] and differ for certain patient populations [9,19], but has not been shown to be statistically different between young men and women [16,18] during the period of interest in this study. However, the influence of age on NAS has not been thoroughly explored. Many patients who are prescribed PD-AFOs, such as individuals post-stroke, are older adults. For example, the average age of first-ever stroke is 68.6 years for men and 72.9 years for women [21]. Thus, understanding how NAS changes with age is critical for developing baseline reference values that can be used to prescribe age-specific orthosis bending stiffness to customize PD-AFOs for each individual's needs. One study that compared NAS in young and middle-aged to elderly adults found NAS was slightly, but not significantly, higher in the elderly adults compared to the young and middle-aged group [16]. However, in that study, the elderly adult population, which ranged from 65 to 85 years old, was evaluated as a single group and was compared to younger group also with a large age range (19–50 years old) [16]. The comfortable gait speed of older adults declines by approximately 20–25% from the age of 60–90 years [22,23]. This age span is also associated with a reduction in peak ankle power [22]. Therefore, late aging effects on NAS are likely ignored when all older adults are grouped together, in turn reducing the ability to discriminate changes in NAS from young to older individuals. Given evidence that gait speed [23] and kinetics [22] are influenced beyond the age of 65 years, evaluating how NAS changes as a function of late age is warranted. It has been shown that the plantar flexion moment during stance, a main component of NAS, decreases with increased age [24]. Thus, it is likely that NAS itself may also change with age.

The long-term goal of this study is to develop an objective PD-AFO bending stiffness prescription model for individuals with plantar flexor weakness, such as those with chronic stroke. Because many individuals prescribed PD-AFOs, including those with chronic stroke, are older adults, the purpose of this study was to characterize the NAS beyond the age of 65 years. We hypothesized that, during gait at preferred speeds, NAS would decrease with age. This hypothesis was based on evidence that the maximum plantarflexion moment and plantar flexor angular impulse was smaller for elderly adults compared to young adults [24,25]. In a cross-sectional study of aging, however, it is likely that potential confounders underlie the observed relationships between age and gait variables. Because gait is typically slower for those who are older [23], walking speed was also considered in the analysis to determine if observed age-related trends were due, in part, to the known influence of speed on ankle stiffness [10,17]. Characterizing NAS in older adults will enhance our understanding of ankle function during gait and could serve as the basis of a prescription model for personalization of PD-AFO bending stiffness for a large portion of patients who are prescribed PD-AFOs.

2. Methods

This paper represents an analysis of data collected from the Mayo Clinic Study Assessing Fall Epidemiology and Risk (SAFER) − a study evaluating fall risk in older women. A total of 125 women between the ages of 65 and 91 years old were recruited and participated in an IRB-approved study at the Mayo Clinic in Rochester, MN. Participants were recruited so each 5-year age strata up to 85 + years were represented by approximately 25 individuals. All participants were community-dwelling women who reported the ability to walk at least a city block without a gait aid (e.g. cane, walker). Women who had a lower extremity joint replacement within a year prior to participation were excluded.

Participants underwent a three-dimensional (3-D) instrumented gait analysis where they walked barefoot, overground. Forty-one retro-reflective markers were affixed to participants' extremities, pelvis, and trunk, including the sacrum and bilaterally on the acromion processes, lateral epicondyle of the elbows, center of the dorsum of the wrists, anterior superior iliac spines (ASIS), lateral femoral condyles, lateral malleoli, mid-thigh, mid-shank, heels and the spaces between the first and second metatarsal heads. The markers placed at bony prominences were used for establishing anatomic coordinate systems for the pelvis, thigh, shank and foot. Additional tracking markers were applied to each segment. One set of data corresponding to the standing position (static data) was recorded in order to calculate the location of the joint centers using 4 additional markers bilaterally on the medial femoral condyle and medial malleoli. Participants walked approximately 7.5 m at a self-selected, preferred walking speed. Kinematics of three steps per leg were recorded by 10 cameras at 120 Hz (Motion Analysis Corporation, Santa Rosa, CA), and ground reaction forces were recorded at 600 Hz by two AMTI force plates (Advanced Mechanical Technology, Inc., Watertown, MA) and three Kistler force plates (Kistler Instrument Corp., Amherst, NY).

Kinematic and kinetic data were each filtered at 6 Hz, using a 4th order Butterworth filter. Using a standard inverse dynamics approach, ankle joint angles and moments were calculated for each participant from the gait analysis data in Visual 3D (C-Motion Inc., Germantown, MD). Right-handed coordinate systems were established for the foot and shank segments such that the positive y-axis was directed to the participant's left, positive x-axis was pointed anteriorly, and positive z-axis was pointed superiorly. All angles were referenced to the proximal segment and all joint moments were resolved in the distal segment's coordinate system. With these definitions, and to align with common practices in clinical settings, moments about the ankle were negated so that plantarflexion was represented as positive. Thus, dorsiflexion was a positive angle and negative moment and plantarflexion was a negative angle and positive moment. Each participant's ankle moment was scaled by body mass, leg length (measured as the vertical distance from the greater trochanter to the floor), and gravitational constant (9.81 m/s^2) [26,27] to create a dimensionless moment that could be compared across participants. A total of four to six stance phases of gait from both legs combined were analyzed for each participant. For each stance