INTRODUCTION
Computational modeling and simulation of the musculoskeletal system has significantly increased in recent years [1-3]. From a basic science perspective, torque-driven and muscle-actuated forward dynamics simulations allow the identification of causal relationships between biomechanical factors and specific task performance by varying one model parameter and keeping other parameters unchanged [3, 4]; thus generating knowledge that cannot be revealed through experimental studies alone. Simulations also allow estimation of quantities, such as muscle forces, that are generally difficult to measure. Furthermore, simulations can aid in designing training or rehabilitation programs to improve specific human and/or biomechanical factors related to the specific task performance.

Anticipatory postural adjustments [5] are the postural changes associated with voluntary movements and are critical for maintaining balance during locomotor initiation. Falls are a serious health hazard to the elderly. Older adults that have delayed anticipatory postural adjustments during step initiation are at a greater risk for falls [5]. Through musculoskeletal modeling, we can gain insight into the coordination of anticipatory postural adjustments and the errors that occur. Stepping is one activity that can reliably produce anticipatory postural adjustments and when balance is perturbed a quick step may prevent a fall [6]. Information gained from previous work conducting rapid voluntary step initiation tests have shown that older adults exert lower peak forces than young adults but not necessarily lower peak power [6].

OBJECTIVE
This study seeks to further understand the impact of aging on lower extremity muscle activity contributions to balance during rapid voluntary stepping in young and older healthy adults using computational modeling and simulation approaches. The objective of this study is to determine individual lower extremity muscle force contributions during rapid stepping.

HYPOTHESIS
It is hypothesized that during rapid voluntary step initiation, muscle weakness in older populations will cause dynamic or biomechanical differences that could increase the likelihood of falls when compared to younger populations.

METHOD
Bilateral ground reaction forces, motion capture (marker trajectories), and EMG data (muscles: tibialis anterior, soleus, medial gastrocnemius, vastus lateralis, and biceps femoris long head) were collected from a healthy young subject (age 23, mass = 79.8 kg, height = 176 cm) and an older subject (age 73, mass = 92.5 kg, height = 174 cm) to examine age-related differences during rapid voluntary stepping.

Subjects were asked to complete a simple step reaction time task [7], where they stepped forward with their left foot in response to an auditory cue. There were two stepping locations, short or long, and two response speeds, quick or self-selected.

OpenSim [4] was utilized to develop a 3-D generic musculoskeletal model consisting of 18 degrees of freedom and 60 Hill-type muscles (30 per leg). The generic model was scaled to create subject-specific models matching each individual’s anthropometry. A weighted least squares problem (inverse kinematics) was solved to obtain a kinematically consistent set of joint angles using the measured motions. The joint angles and the measured ground reaction forces were then used to resolve net joint moments into individual muscle forces at each instant in time. This was achieved by solving a static optimization problem, where muscle forces are determined by minimizing the sum of squared muscle activations:

$$\min \ J = \sum_{m=1}^{n} (a_m)^2$$
subject to  \[ \tau_j = \sum_{m=1}^{n} a_m f(I_m^0, l_m, v_m) \] \( r_{m,j} \)

where \( n \) is the number of muscles in the model, \( a_m \) is the activation level of muscle \( m \), \( f(I_m^0, l_m, v_m) \) is its maximum isometric force, \( l_m \) is its length, \( v_m \) is its shortening velocity, \( f(I_m^0, l_m, v_m) \) is its force-length-velocity surface, \( r_{m,j} \) is its moment arm about the \( j \)th joint axis, and \( \tau_j \) is the generalized force acting about the \( j \)th joint axis.

RESULTS
Overall, the experimental EMG patterns were captured by the simulated muscle activations (Fig. 1, left). The older subject’s soleus (SOL) EMG data was slightly variable in both the long quick (LQ) and short quick (SQ) conditions; however, the simulated muscle activations are consistent with published data [8]. In general, the simulated muscle activations and estimated muscle forces for the LQ and SQ step conditions displayed similar behavior between the young and older subject.
The estimated muscle forces (Fig 1., right) show that the young subject mainly relies on ankle plantar flexors, of which the soleus muscle is a primary one, during push-off. In contrast, the older subject relies on both the soleus muscle as well as the medial gastrocnemius (MGAS) muscle, which is both a primary ankle plantar flexor and knee flexor, for push-off.

![Figure 1](image)

Figure 1. Simulated muscle activations (left) and estimated muscle forces (right) for the stance (right) leg during quick short (SQ) stepping conditions for the young (black) and older (red) subjects. The plots are from the stepping/swing (left) leg toe-off (0%, corresponding to 12% of the gait cycle (GC) or early mid-stance) until the stance (right) leg toe-off (100%, corresponding to 62% of the GC or initial swing).

To better quantify differences between the two subjects, total stance leg ankle plantar flexor net joint moment and power contributions were calculated (Fig. 2). Compared to the young subject, the older subject generated lower ankle plantar flexor moment, and subsequently power, in all stepping conditions. A noticeable contrast was also observed in the quick response conditions, with the older subject exhibiting a different ankle moment pattern.

![Figure 2](image)

Figure 2. Total stance (right) leg ankle plantar flexor net joint moment and power contributions in long quick (LQ), short quick (SQ), long self-selected (LSS), and short self-selected (SSS) stepping conditions for the young (black) and older (red) subjects. The young subject exhibits higher ankle moments, and power, compared to the older subject, in all stepping conditions.

**DISCUSSION**

The differences observed in this preliminary analysis could be attributed to reduced muscle function due to aging. The results indicate that the young subject adopts similar strategies during quick and self-selected step initiation, while the older subject exhibits a different mechanism during rapid voluntary stepping.

The stance leg ankle plantar flexor moment contribution, in both quick and self-selected stepping conditions, was different between the young and older subject, with the latter exhibiting smaller plantar flexor moments. This could be attributed to different force generation capabilities observed in the young and older subject (Fig 1., right) during step initiation in response to an auditory cue. Additionally, the power contribution to ankle plantar flexion during push-off was stronger in the young subject, with the older subject exhibiting smaller amplitudes, leading to slower advancement of the stance leg into swing.

The findings of this preliminary modeling work may play an important role in identifying individuals at risk of falls based on the way they initiate a stepping response. These individuals can then be targeted and encouraged to enroll in fall prevention programs lead by physical therapists or orthopedic surgeons. Future work will focus on developing more accurate subject specific models and inclusion of more subjects in the analysis.

**ACKNOWLEDGMENTS**

Funding provided by the Pittsburgh Claude D. Pepper Older Americans Independence Center (P30AG024827). I would like to acknowledge Dr. Mahboobin for his help and guidance throughout the study.

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