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# Age and falls history effects on antagonist leg muscle coactivation during walking with balance perturbations

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ABSTRACT

*Background:* Inspired by a reliance on visual feedback for movement control in older age, optical flow perturbations provide a unique opportunity to study the neuromuscular mechanisms involved in walking balance control, including aging and falls history effects on the response to environmental balance challenges. Specifically, antagonist leg muscle coactivation, which increases with age during walking, is considered a neuromuscular defense against age-associated deficits in balance control. The purpose of this study was to investigate the effects of age and falls history on antagonist leg muscle coactivation during walking with and without optical flow perturbations of different amplitudes.

*Methods:* Eleven young adults [mean (standard deviation) age: 24.8 (4.8) years], eleven older non-fallers [75.3 (5.4) years] and eleven older fallers [age: 78 (7.6) years] participated in this study. Participants completed 2-minute walking trials while watching a speed-matched virtual hallway that, in some conditions, included mediolateral optical flow perturbations designed to elicit the visual perception of imbalance.

*Findings:* We first found that lower leg antagonist muscle coactivation during normal walking increased with age, independent of falls history. We also found that older but not young adults increased antagonist leg muscle coactivation in the presence of optical flow perturbations, with more pervasive effects in older adults with a history of falls.

*Interpretation:* Our findings allude to a greater susceptibility to optical flow perturbations in older fallers during walking, which points to a higher potential for risk of instability in more complex and dynamic everyday environments. These findings may also have broader impacts related to the design of innovative training paradigms and neuromuscular targets for falls prevention.

## 1. Introduction

Roughly a third of people over the age of 65 fall at least once annually and 25–30% of these falls lead to moderate to severe injury (Alexander et al., 1992). The consequences may be functionally devastating for the individual and yield an enormous financial burden on the health care system. Unfortunately, evidence even suggests that the rate of injurious falls is accelerating, with 2.4 million fall-related emergency department visits in 2011, up 46% from 2001 despite only a 17% increase in the older adult population (Centers for Disease Control and Prevention and National Center for Injury Prevention and Control, 2017). Indeed, while numerous studies and clinical trials have attempted to decrease the prevalence and severity of falls, fall rates have been resistant to change (Gardner et al., 2000; Kraemer et al., 2009; Rubenstein et al., 2000). The physiological mechanisms underlying age-related falls risk are likely multifactorial, but may include declines in somatosensory acuity and a decreased neuromuscular capacity to respond to unexpected balance challenges. Accordingly, balance perturbations have become highly prevalent in studying the fidelity of walking balance control in older adults due to their capacity to elicit age-associated differences that are not otherwise apparent during normal, unperturbed walking (Franz et al., 2015; Martelli et al., 2017).

As a potential neuromuscular defense mechanism for deficits in balance control, the concurrent activation of antagonist muscles during walking (i.e., antagonist coactivation), increases in old age (Hortobagyi et al., 2009; Hortobagyi et al., 2011; Hortobagyi and DeVita, 2000; Mian et al., 2006; Ortega and Farley, 2015; Peterson and Martin, 2010). This age-related increase in antagonist coactivation is thought to bolster leg joint stiffness and thereby improve joint stability as a means to mitigate or improve the response to balance disturbances (Finley et al.,

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2012; Hortobagyi and DeVita, 2000). Indeed, evidence from arm reaching tasks shows that antagonist coactivation and limb stiffness concurrently increase with the magnitude of external force perturbations (Wong et al., 2009). Moreover, another study found a significant positive correlation (i.e., r = 0.42) between antagonist coactivation and the quality of standing postural control among older adults (Nagai et al., 2011). However, although intuitive and consistent with the prevailing mechanistic interpretation, it remains unclear whether the coactivation of antagonistic leg muscles in older adults increases further in the presence of walking balance perturbations. This may be particularly prevalent in older adults with a history of falls; compared to non-fallers, older fallers exhibit disproportionate decrements in metrics of walking balance control (Cebolla et al., 2015; Svoboda et al., 2017).

Due to their ability to elicit corrective motor responses, optical flow perturbations (i.e., eliciting the visual perception of imbalance) have increasingly been used to study balance control during walking (Franz et al., 2015; Jeka et al., 2010; O'Connor and Kuo, 2009; McAndrew et al., 2011). Moreover, this class of perturbations is uniquely well-suited to study the neuromuscular control of walking balance in older age; compared to young adults, older adults rely more on visual feedback for motor planning and execution (Franz et al., 2015; Jeka et al., 2010) – an effect that becomes even more pronounced in older adults with a history of falls (Lord and Webster, 1990). Accordingly, optical flow perturbations applied during walking can elicit age-related differences in many metrics of walking balance control that are not otherwise apparent during unperturbed walking, including increased gait variability and decreased local dynamic stability (Francis et al., 2015; Franz et al., 2015). More recently, we incorporated electromyographic (EMG) recordings of leg muscle activities to show that young adults walking with similar perturbations do show evidence of elevated antagonist coactivation compared to normal, unperturbed walking – particularly during limb loading in early stance (Stokes et al., 2017). However, despite longstanding scientific and clinical interest, our understanding of age and falls history effects on the neuromuscular mechanisms involved in walking balance control and the response to optical flow perturbations remains fundamentally incomplete.

Therefore, the purpose of this study was to investigate the effects of age and falls history on antagonist leg muscle coactivation during walking with and without optical flow perturbations of different amplitudes. We used a virtual reality environment to apply continuous mediolateral optical flow perturbations during treadmill walking while recording EMG activities of antagonist upper and lower leg muscle pairs. We hypothesized that: (1) compared to young adults, aging and falls history would increase antagonist muscle coactivation during walking, and (2) these between-group differences would increase in the presence of optical flow perturbations.

## 2. Methods

### 2.1. Participants

Eleven healthy young adults [6 female, mean (standard deviation, SD), age: 24.8(4.8) years, height: 1.72(0.01) m, mass: 67.2(8.8) kg], eleven healthy older adults [6 female, age: 75.3(5.4) years, height: 1.75(0.01) m, mass: 73.4(16.1) kg] and eleven older adults with a history of falls [7 female, age: 78(7.6) years, height: 1.6(0.12) m, mass: 69.3(14.0) kg] participated in this study. Older adults were considered to be fallers if they had fallen one or more times in the past year. For this study, falls counted towards the self-reported total were defined according to the Kellogg International Work Group Definition (1987). Participants also completed a health questionnaire prior to participating, which we used to exclude based on: BMI ≥ 30, sedentary lifestyle, orthopedic or neurological condition, or taking medication that causes dizziness. Visual acuity was examined for all participants to ensure each had sufficient vision. The experimental protocol was approved and conducted in accordance with the University of North Carolina Internal Review Board, and participants provided written informed consent prior to participating. Young adult data from a previously published study were reanalyzed for comparison to older adults.

### 2.2. Experimental protocol

We first recorded all participants' preferred overground walking speed [Non-fallers: 1.19(0.20) m/s, Fallers: 1.03(0.22) m/s] using two photocells (Brower Timing, Draper, UT) as the average of three times taken to walk the middle 2 m of a 10 m walkway. Young adults completed all testing on a dual-belt, force measuring treadmill (Bertec Corp., Columbus, OH) at 1.25 m/s, a speed comparable to their preferred overground walking speed [1.29(0.18) m/s, p = 0.51]. We note that preferred speed differed only between young adults and older fallers ($p < 0.01$). Older adults walked on the same dual-belt treadmill completing all trials at their preferred walking speed. In both instances, the treadmill was surrounded by a semi-circular curved screen measuring 2.24 m high and 2.83 m wide. The participants were then secured in an overhead harness, which was worn for all walking trials. Participants first walked on the treadmill at their preferred walking speed for 5 min in order to acclimate to the treadmill. Next, participants completed four 2-minute walking trials, in randomized order, while watching a speed-matched, virtual hallway rear-projected onto the screen. The walking trials consisted of usual, unperturbed walking and continuous mediolateral optical flow perturbations at amplitudes of 20, 35, and 50 cm. Each perturbation was comprised of the sum of three sinusoids, such that the full amplitude was applied at 0.250 Hz and half that amplitude was applied at 0.125 Hz and 0.442 Hz. The perturbations were consistent with visual feedback associated with head movement, meaning that the hallway's end moved very little compared to the foreground. This allowed us to provoke balance corrections rather than heading corrections.

### 2.3. Measurements and data analysis

A 14-camera motion capture system (Motion Analysis, CA) operating at 100 Hz, recorded the three dimensional positions of markers placed on participants' right and left heels and sacrum. Marker trajectories were then filtered using a 4th order Butterworth filter and a cut-off frequency of 8 Hz. Heel strikes were identified from peaks in the fore-aft position of the heel markers relative to the sacral marker (Zeni et al., 2008). Additionally, after preparing the shaved skin with alcohol, we collected EMG recordings at 1000 Hz from wireless electrodes (Trigno, Delsys, Inc., Boston, MA) placed over the mid-bellies five right leg muscles: medial gastrocnemius (MG), soleus (SOL), tibialis anterior (TA), vastus lateralis (VL) and medial hamstring (MH), following the guidelines of Cram and Kasman (1998). We then processed the data through a custom MATLAB script (Mathworks, Inc., Natick, MA). We bandpass (20–450 Hz) filtered and rectified the data before normalizing the amplitudes to the mean values during usual walking. We then used previously published procedures to calculate muscle coactivation indices (CI) as the normalized shared area of overlap between 10 Hz linear envelopes derived from three different antagonist muscle pairs: MG/TA, SOL/TA, and VL/MH (Falcon and Winter, 1985; Franz and Kram, 2013):

$$\mathrm{CI}(\mathrm{EMG1},\mathrm{EMG2}) = 2 \times \left( \frac{\int \min(\mathrm{EMG1},\mathrm{EMG2})}{\int \min(\mathrm{EMG1},\mathrm{EMG2}) + \int \max(\mathrm{EMG1},\mathrm{EMG2})} \right) \times 100$$

Specifically, we calculated stance and swing phase antagonist coactivation indices using EMG data encompassing 0–61% and 61–100% of the gait cycle, respectively (Perry and Burnfield, 2010).

We analyzed our outcome measures in three steps. First, a mixed, two-way factorial analysis of variance (ANOVA) tested for main effects of and interactions between perturbation amplitude (unperturbed [i.e.,