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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 nonfallers [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.

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Conflict of interest

The authors have no conflicts of interest

Keywords

Balance; Elderly; Neuromuscular; Stability

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 (McAndrew et al., 2011). 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., 2015a; Martelli et al., 2015).

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 and DeVita, 2000; Hortobagyi et al., 2011; Hortobagyi et al., 2009; 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., 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., 2015a; Jeka et al., 2010; O'Connor and Kuo, 2009). 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., 2015a; 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., 2015b). 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 minutes 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 cutoff 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):

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

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., 0 cm], 20, 35, and 50 cm) and group (young, non-fallers, fallers) on step length, step width, their respective variabilities, and the following six antagonist coactivation outcome measures: stance and swing phase MG/TA, SOL/TA, and VL/MH. For each outcome having a significant perturbation main effect in the two-way ANOVA, we performed a post-hoc one-way repeated measures ANOVA to identify which group(s) were responsible for this significant effect. Second, only for the group(s) exhibiting a significant perturbation main effect in their post-hoc one-way ANOVA, we interrogated each muscle's 10 Hz linear envelope to identify the muscle(s) responsible for and more precisely when in the gait cycle those effects occurred. Specifically, in these select cases, a one-way repeated measures ANOVA tested for significant effects of perturbations on the average EMG amplitude within three bins comprising the stance phase of the gait cycle (GC) (early stance: 1–10% GC, mid-stance: 10–30% GC, terminal stance: 30–60% GC) and/or three bins comprising the swing phase of the gait cycle (early swing: 60–73% GC, mid-swing: 73–87%

GC, terminal swing: 87–100% GC), following the conventions of Perry et al. (Perry and Burnfield, 2010). Finally, exploratory independent t-tests compared antagonist coactivation between fallers and non-fallers for each condition. For all analyses, we defined significance using an alpha level of 0.05.

3. Results

3.1. Age and perturbation effects on step kinematics

Table 1 summarizes optical flow perturbation effects on step kinematics as a function of group to provide context for the primary EMG-based outcomes. Group main effects revealed that older fallers averaged shorter and more variable step widths and step lengths than young adults (Table 1). Compared to unperturbed walking, perturbations elicited larger step length and width variabilities and wider steps in all groups, and shorter steps in older but not young adults.

3.2. Group effects on antagonist coactivation

The two-way ANOVA revealed significant main effects of group for stance phase, but not swing phase, MG/TA ($p=0.050$, $\eta_p^2 = 0.18$) and SOL/TA ($p=0.027$, $\eta_p^2 = 0.21$) antagonist coactivation. Specifically, age, independent of falls history, significantly increased stance phase lower leg antagonist muscle coactivation during walking compared to young adults (Figs. 2–3). Here, MG/TA coactivation increased, on average, by 138% ($p=0.027$, Fig. 3) and SOL/TA coactivation increased, on average, by 120% ($p=0.021$, Fig. 2) for older versus young adults. However, on average across conditions, lower leg antagonist coactivation did

not differ significantly between older non-fallers and fallers. We also found no significant main effects of group for stance nor swing phase VL/MH antagonist coactivation.

3.3. Interactions and perturbation effects on antagonist coactivation by group

Our two-way ANOVA also revealed significant main effects of optical flow perturbations on antagonist coactivation for MG/TA (stance: $p=0.002$, $\eta_p^2 = 0.15$; swing: $p=0.004$, $\eta_p^2 = 0.14$) and SOL/TA (stance: $p<0.001$, $\eta_p^2 = 0.26$; swing: $p<0.001$, $\eta_p^2 = 0.22$). In contrast, perturbations had no effect on stance nor swing phase VL/MH antagonist coactivation. In addition, one significant interaction effect emerged that revealed between-group differences in TA/SOL coactivation, specifically during the swing phase, that were not otherwise apparent during unperturbed walking (age $\times$ perturbation, $p=0.015$, $\eta_p^2 = 0.16$). Here, perturbations increased swing phase TA/SOL coactivation by, on average, by up to 86% in older fallers ($p=0.035$) but not in non-fallers compared to unperturbed walking (Fig. 2).

Accordingly, at the largest amplitude perturbation, older fallers walked with 175% larger TA/SOL coactivation than young adults during leg swing ($p=0.042$) (Fig. 2) – an effect precipitated by higher TA (midswing) and SOL (terminal swing) activations (Fig. 1).

Post-hoc ANOVAs, used to explore the perturbation main effects, revealed that optical flow perturbations had no significant effects on any leg muscle antagonist coactivation outcomes in young adults. In contrast, during the stance phase, perturbations significantly increased TA/SOL coactivation in both groups of older adults, by an average of up to 58% (non-fallers: $p=0.002$, fallers: $p=0.002$) (Fig. 2). This finding was precipitated by higher TA activation (fallers and non-fallers) and higher SOL activation (fallers) during early to midstance (Fig. 1). In addition, only in older fallers did perturbations significantly increase TA/MG antagonist coactivation, by an average of up to 54% during stance and 106% during swing (p 's < 0.050) (Fig 3). Specifically, we observed higher MG activity during early stance ($p=0.050$) and higher TA activity during early stance ($p=0.039$) and midswing ($p=0.048$) (Fig. 1).

Finally, during unperturbed walking, we found no significant pairwise comparisons between older fallers and non-fallers for any measure of antagonist coactivation. However, our exploratory comparisons revealed that perturbations elicited up to 109% more stance phase VL/MH coactivation in older fallers compared to non-fallers ($p=0.047$, $\eta^2=0.175$) (Fig. 4).

4. Discussion

We investigated differences in step kinematics and antagonist leg muscle coactivation during walking between healthy young adults, older adult non-fallers, and older adults with a history of falls and changes thereof when responding to optical flow perturbations of different amplitudes. We first hypothesized that aging and falls history would increase antagonist leg muscle coactivation during walking. In partial support of this hypothesis, lower leg muscle coactivation during the stance phase but not the swing phase of walking increased with age, but this effect was independent of falls history. Further, in support of our second hypothesis, we found that between-group differences in antagonist leg muscle coactivation were larger in the presence of optical flow perturbations, with more pervasive

effects in older fallers compared to those without a history of falls. Our results are generally consistent with the widespread notion that antagonist leg muscle coactivation is a neuromuscular strategy to mitigate the effects of balance challenges that could precipitate a fall (Finley et al., 2012; Hortobagyi and DeVita, 2000), a strategy independently effected by aging and falls history.

Consistent with prior studies (Franz and Kram, 2013; Hortobagyi and DeVita, 2000), older adults averaged greater lower leg antagonist leg muscle coactivation than younger adults during walking. However, these effects were relatively modest; older non-fallers walked only with greater stance phase TA/MG coactivation than young adults. Moreover, similar to our activation profiles, these changes have been attributed predominantly to an increased and prolonged activation of the TA during the initial loading response (Franz and Kram, 2013). Greater coactivation between the antagonistic TA and MG could reflect a need for more ankle and/or knee joint stability as weight shifts from one leg to another during each step-to-step transition. These data contribute to the growing evidence that older adults adopt altered neuromuscular control, even during normal, unperturbed walking. However, these changes appear not to be disproportionately adopted by older adults with a history of falls; indeed, we found no difference between older fallers and older non-fallers for any EMG-based outcome measure during unperturbed walking. There is some evidence that older fallers have increased amounts of muscle coactivation between the tibialis anterior and lateral gastrocnemius compared to non-fallers (Marques et al., 2013). Thus, it is possible that we would have discovered differences in coactivation had we recruited older fallers with a more extensive history of several falls.

Consistent with our second hypothesis, optical flow perturbations revealed independent effects of aging and falls history on antagonist leg muscle coactivation that were not otherwise apparent during normal, unperturbed walking. Unlike either cohort of older adults in this study, and despite smaller but nevertheless hallmark perturbation-induced changes in step kinematics, young adults did not increase antagonist leg muscle coactivation in response to optical flow perturbations of any amplitude. This is consistent with our prior work showing that, compared to older adults, young adults are less sensitive to the visual perception of instability provoked by our virtual reality paradigm. In contrast, compared to unperturbed walking, older non-fallers and fallers both significantly increased stance-phase SOL/TA coactivation in the presence of perturbations. Here, compared to the effects of age alone, optical flow perturbations elicited pronounced increases not only in TA but also in SOL activation during early to midstance in older fallers. Moreover, unlike older non-fallers, only older adult fallers increased TA/MG coactivation during the stance phase and both TA/MG and TA/SOL coactivation during leg swing – the latter effect governed by increased TA activation and an earlier onset for SOL activation. At the lower leg, these differences in the response to perturbations between older fallers and non-fallers were admittedly modest. Nevertheless, the more widespread effects of perturbations on lower leg antagonistic muscle coactivation in older adults with a history of at least one fall do allude to possible unique changes in neuromuscular control to increase ankle and/or knee joint stability when accommodating balance challenges during walking.

In contrast to some (Mian et al., 2006; Ortega and Farley, 2015; Peterson and Martin, 2010) but not all (Franz and Kram, 2013; Hortobagyi and DeVita, 2000) prior literature, we found no effect of age alone on upper leg (i.e., thigh) antagonistic muscle coactivation during unperturbed walking nor in the presence of perturbations. Independent of age and falls history, human walking is typically characterized by greater upper leg than lower leg antagonist muscle coactivation, which increases further with speed or uphill grade (Franz and Kram, 2013; Mian et al., 2006; Ortega and Farley, 2015). Here, we also found no significant main effect of optical flow perturbations on VL/MH coactivation. These results suggest that across the study cohort, perturbations may not have increased the demands for sagittal plane hip and/or knee joint stability. However, our exploratory pairwise comparisons discovered, on average, more than twice the VL/MH coactivation in older fallers compared to older non-fallers, but only across the two largest amplitude perturbations. We interpret this finding to suggest that older fallers required additional neuromuscular control from those muscles spanning the hip and knee, thus making them more susceptible than non-fallers to larger amplitude optical flow perturbations. There are few studies available to provide context for this finding. Ochi et al. (2014) found greater upper leg muscle coactivation in fallers compared to non-fallers during the swing phase of a step recovery task. They suggested that this greater muscle coactivation contributed to slower recovery responses observed in fallers compared to non-fallers. Accordingly, our older adult fallers may have exhibited a neuromuscular tradeoff, wherein they increase joint stability at the expense of slower recovery responses from one step to the next.

Antagonist coactivation is frequently present in people with balance deficits and is considered a strategy to increase leg joint stiffness and aid in responding to, or at least in mitigating, balance challenges (Finley et al., 2012; Hortobagyi et al., 2011; Peterson and Martin, 2010). Accordingly, age and falls history effects on antagonist leg muscle coactivation, which we observed most prominently in the presence of optical flow perturbations, likely contribute joint stability in the face of locomotor tasks requiring larger step-to-step balance corrections. That older but not younger adults respond to these perturbations with increased antagonist leg muscle coactivation may have a sensorimotor explanation. For example, aging increases the reliance on visual feedback for motor planning and execution, including that required for walking balance control. Indeed, we have previously reported that older adults are much more susceptible to similar perturbations using outcome measures that include gait variability (Francis et al., 2015; Qiao et al., 2018) and local dynamic stability (Franz et al., 2015a). As a related example, age-associated declines in peripheral sensory perception have also been implicated in contributing to both elevated antagonistic leg muscle coactivation (Ortega and Farley, 2015) and the reliance on visual feedback (Franz et al., 2015a; Jeka et al., 2010). In their seminal paper, Lord and Webster (Lord and Webster, 1990) also found that visual dependence was stronger in older fallers compared to non-fallers, which may help explain their more pervasive response to optical flow perturbations in this study. As an alternative explanation, older adults with a history of falls may present with a fear of recurrent falls, which has been independently linked to increased antagonist leg muscle coactivation during walking (Nagai et al., 2012).

There are a few important limitations to our study. First, our continuous optical flow perturbations were prescribed as a complex combination of sinusoids that would be

challenging for subjects to anticipate. However, we cannot exclude the possibility that young adults did not alter their muscle coactivation patterns because they had learned to predict the perturbations. Though, we consider this highly unlikely, as we have no reason to suspect that such a learning effect would be realized by young but not older adults. In addition, although all participants had treadmill walking experience, older adults with less treadmill experience than young adults may increase antagonist muscle coactivation more than during overground walking in response to the same amplitude balance perturbation. We also acknowledge the relatively small number of participants comprising each cohort, and cannot exclude the possibility that we were statistically underpowered to identify all group $\times$ perturbation interaction effects. Nevertheless, we find our results allude to important aspects of aging and falls history effects on neuromuscular control during walking and response to balance perturbations that should inspire future investigation. Finally, older fallers walked at a speed not significantly different from non-fallers, but slower than young adults. Based on our prior work (Stokes et al., 2017), we do not anticipate that walking slower would increase the susceptibility to optical flow perturbations and thus explain our findings unique to older fallers.

5. Conclusion

This study provides insight into aging and falls history effects on the neural control strategies used to maintain walking balance and respond to optical flow perturbations. Our findings suggest that older adults, particularly those with a falls history, may rely more on increased leg joint stiffness than young adults to accommodate balance challenges during walking. Moreover, elevated antagonist upper leg muscle coactivation at larger perturbation amplitudes, unique to older adults with a history of falls, may allude to an increased susceptibility to optical flow perturbations, which may place them at risk for instability in more complex and dynamic everyday environments. Finally, these findings may have broader impacts related to the design of innovative training paradigms and neuromuscular targets for falls prevention.

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Research Highlights

- Antagonist muscle coactivation improves joint stability to preserve balance
- Quantified age and falls history on antagonist leg muscle coactivation during walking
- Optical flow perturbations using virtual reality provoked lateral gait instability
- Antagonistic coactivation increased due to aging and falls history
- Larger between-group differences in the presence of optical flow perturbations

Figure 1.

Group mean 10 Hz EMG linear envelopes for antagonist leg muscle pairs in (A) young, (B) older non-fallers, and (C) older fallers over an averaged gait cycle, reported as a percent (%) from heel-strike to heel-strike. Each muscle's EMG amplitudes were normalized to the average value over a complete stride during normal, unperturbed walking. Solid lines indicate unperturbed walking while dashed lines indicate walking with the largest amplitude perturbations. Vertical lines indicate toe-off, signifying the transition between the stance phase and the swing phase. Numerical indicators represent gait cycle phases with significant main effects of perturbation amplitude ($p < 0.05$) (i.e., 1: early stance, 2: midstance, 5: midswing, 6: terminal swing). ML: medial hamstring, VL: vastus lateralis, MG: medial gastrocnemius, TA: tibialis anterior, SOL: soleus.

Figure 2.

Group mean (standard error) antagonist coactivation (%) between soleus (SOL) and tibialis anterior (TA) during the (A) stance and (B) swing phases in young, older non-fallers, and older fallers during unperturbed and perturbed walking. Asterisks indicate significant (i.e., $p < 0.05$) effects of group at a given perturbation amplitude. f and nf indicate significant perturbation effects for older fallers and non-fallers, respectively.

Figure 3.

Group mean (standard error) antagonist coactivation (%) between medial gastrocnemius (MG) and tibialis anterior (TA) during the (A) stance and (B) swing phases in young, older non-fallers, and older fallers during unperturbed and perturbed walking. Asterisks indicate significant (i.e., $p < 0.05$) effects of group at a given perturbation amplitude. f and nf indicate significant perturbation effects for older fallers and non-fallers, respectively.

Figure 4.

Group mean (standard error) antagonist coactivation (%) between vastus lateralis (VL) and medial hamstring (MH) during the (A) stance and (B) swing phases in young, older nonfallers, and older fallers during unperturbed and perturbed walking. We found no significant main effect of perturbation amplitude on upper leg antagonist coactivation during walking. Double asterisks (**) indicate significant exploratory pairwise comparisons between older fallers and older non-fallers.

Table 1.

Mean (standard deviation) limb-level step kinematics and variabilities (cm)

			Optical Flow Perturbation				
	Main Effects	Group	0 cm	20 cm	35 cm	50 cm	
SL	$P_{group} = 0.043$	Young	69.1(2.4)	68.7(2.7)	69.4(3.5)	67.3(2.9)	
	$P_{pert} < 0.001$	Non-fallers	63.5(8.3)	61.9(8.6)	60.9(9.6)	60.4(9.5)	*
		Fallers	61.9(11.5)	59.0(13.7)	58.0(12.0)	57.3(12.2)	*
SLV	$P_{group} = 0.017$	Young	1.8(0.5)	2.2(0.5)	2.5(0.9)	2.8(1.4)	*
	$P_{pert} < 0.001$	Non-fallers	2.7(0.7)	3.5(1.3)	4.0(2.5)	4.3(2.6)	*
		Fallers	2.7(0.9)	3.7(1.5)	4.2(1.9)	5.1(2.1)	*
SW	$P_{group} = 0.892$	Young	12.0(3.7)	12.4(4.0)	13.2(4.7)	13.3(4.3)	*
	$P_{pert} < 0.001$	Non-fallers	12.9(3.3)	13.9(3.5)	13.9(3.3)	14.3(3.6)	*
		Fallers	11.2(4.3)	12.8(4.2)	13.0(4.2)	13.8(3.7)	*
SWV	$P_{group} = 0.010$	Young	2.1(0.5)	3.6(1.2)	3.7(1.6)	4.4(1.8)	*
	$P_{pert} < 0.001$	Non-fallers	2.8 (0.8)	5.0(1.5)	5.3(1.6)	5.6(1.9)	*
		Fallers	3.3(0.4)	5.8(2.1)	6.6 (3.1)	7.1(3.1)	*

SL: step length, SW: step width, SLV: step length variability (i.e., standard deviation of step length time series), SWV: step width variability (i.e., standard deviation of step width time series).

Asterisks (*) indicate a significant main effect of perturbations within each group ($p < 0.05$)