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Effects of initial foot position on postural responses to lateral standing surface perturbations in younger and older adults

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ABSTRACT

Background: An age-related decline in standing balance control in the medio-lateral direction is associated with increased risk of falls. A potential approach to improve postural stability is to change initial foot position (IFP).
Research questions: In response to a lateral surface perturbation, how are lower extremity muscle activation levels different and what are the effects of different IFPs on muscle activation patterns and postural stability in younger versus older adults?
Methods: Ten younger and ten older healthy adults participated in this study. Three IFPs were tested [Reference (REF): feet were placed parallel, shoulder-width apart; Toes-out with heels together (TOHT): heels together with toes pointing outward; Modified Semi-Tandem (M-ST): the heel of the anterior foot was placed by the big toe of the posterior foot]. Unexpected lateral translations of the standing surface were applied. Electromyographic (EMG) activity of the lower extremity muscles, standard deviation (SD) of the body's CoM acceleration (SD of CoMAccel), and center of pressure (CoP) sway area were compared across IFPs and age.
Results: Activation levels of the muscles serving the ankle and gluteus medius were greater than for the knee joint muscles and gluteus maximus in the loaded leg across all IFPs in both groups. TOHT showed greater EMG peak amplitude of the soleus and fibularis longus compared to REF, and had smaller SD of CoMAccel and CoP sway area than M-ST. Compared to younger adults, older adults demonstrated lower EMG peak amplitude and delayed peak timing of the fibularis longus and greater SD of CoMAccel and CoP sway area in all IFPs during balance recovery.
Significance: During standing balance recovery, ankle muscles and gluteus medius are important active responders to unexpected lateral surface perturbations and a toes-out IFP could be a viable option to enhance ankle muscle activation that diminishes with age to improve postural stability.

1. Introduction

Fall-related injuries often lead to functional limitation and long-term disability [1]. Older adults display reduced lower extremity strength and increased postural sway during standing balance control that is associated with increased risk of falling and fall-related injuries [2]. Consequently, identifying approaches to improve postural stability during balance recovery is an important goal.

During standing balance recovery in response to external surface perturbations, immediate muscle activation at the ankle, knee, and hip joints play an important role in moving the body's center of mass (CoM) towards the inside of base of support (BoS). Previous studies have demonstrated that joint kinematics and muscle responses at each joint are modified based upon perturbation direction and intensity [3,4]. However, differences in muscle activation levels across lower extremity joints have been relatively unexplored. Therefore, understanding such relationships will provide insight into the primary target for rehabilitation assessments and interventions to improve standing balance recovery in older adults.

In order to regain balance following external perturbations, a modification of body's CoM position relative to the BoS is performed with or without compensatory stepping and/or grabbing a nearby support [5,6]. Compared to younger adults, older adults have greater difficulty in recovering standing balance during both with and without

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compensatory stepping responses. For example, older adults demonstrated difficulties in performing rapid counter-rotation of the trunk and hip abduction to initiate the compensatory loaded limb side-stepping in response to unexpected medio-lateral (M-L) perturbations [7]. To recover balance without stepping following external perturbation, lower torque generation with slower muscle response times at the ankle joint and reduced hip flexion strength in older adults lead to difficulties in balance recovery in response to unexpected anterior-posterior (A-P) perturbations [4,8]. Considering these age-related changes in lower extremity muscle activity and joint movement control, standing balance recovery in response to external perturbations becomes more challenging for older adults.

Neuromechanical factors that contribute to postural stability could be modified via different initial foot positions (IFPs). With bilateral symmetrical IFP, for example, a wide IFP requires less muscle activation at the ankle and knee joints to recover standing balance from unexpected M-L perturbations compared to a narrow IFP [9]. Compared to a parallel IFP, a toes-out IFP provides greater hip abduction and foot eversion that contribute to M-L balance control while maintaining quiet standing [10,11]. Furthermore, we have previously reported that a toes-out IFP increased hip-abductor and adductor muscles activation during a sit-to-stand. Older adults activated these muscles to a greater degree in a toes-out IFP compared to younger adults [12], indicating that modifying IFP could be a strategy to compensate for reduced knee extensor strength and postural stability in older adults. With bilateral asymmetric IFP, abnormalities in standing balance control in the M-L direction including greater CoP excursions, increased peak-to-peak CoP amplitude, and increased angular displacement with large variability in the ankle, knee, and hip joints were observed in older adults compared to younger adults [13]. However, few studies have addressed the effect of bilateral asymmetric IFP on standing balance recovery in response to external lateral perturbation and assessed the age-related differences.

To date, the effect of IFPs on leg muscle activation patterns and postural stability in younger and older adults in response to external lateral perturbation remains unclear. If modifying the IFP engages the hip and ankle musculatures that are associated with maintaining postural stability, rehabilitation design may consider such approach to target specific muscles. Therefore, the focus of this study was to investigate the effects of changes in IFPs on lower extremity muscle responses during feet-in-place balance recovery following unexpected external lateral surface perturbations. The specific purposes were to (1) compare muscle activation level across ankle, knee, and hip musculatures during balance recovery, (2) determine the effect of IFPs on muscle activation patterns and postural stability and, (3) characterize how aging influences these relationships. We expected to see greater activation levels of ankle muscles than knee and hip muscles based on previous findings that indicated that the central nervous system appeared to recognize the need to stabilize the joint closest to perturbation first with a quick response [3]. We also hypothesized that a toes-out IFP would increase foot evertor and hip abductor muscle activity and reduce M-L postural sway, and older adults would show decreased activation of these muscles due to age-related decline in ankle and gluteus muscle strength with increased postural sway [14,15].

2. Methods

2.1. Participants

Ten healthy younger adults (5 men, 5 women, 20 ± 1 years) and ten healthy older adults (5 men, 5 women, 77 ± 7 years) participated. Participants were excluded from this study if they (1) had deficits or disorders that could affect balance control, 2) had a history of dizziness and imbalance, 3) had a history of musculoskeletal, neurological, visual or vestibular disorder, (4) had recent illnesses or injuries, (5) had body mass index (BMI) within the obesity range (BMI is 30 kg m^{-2} or higher). The BMI mean (Std. Deviation) of older adults was 23.53 (3.02) kg m^{-2} and younger adults was 22.68 (1.84) kg m^{-2}. All procedures were approved by the University of Texas at Austin's Institutional Review Board and were in accordance with the Helsinki Declaration of 1975.

2.2. Apparatus and setup

A 10-camera motion capture system (VICON Motion Systems Ltd, UK) was used to record the whole body movement at 120 Hz. Thirty-nine reflective markers were placed on the body according to the full body Plug-In Gait [12]. Ground reaction forces (GRFs) and center of pressure (CoP) were recorded by the force-instrumented treadmill (M-GAIT, Motekforce Link, Amsterdam, Netherlands) at 1200 Hz.

Treadmill lateral perturbation was delivered by sending a single square pulse signal via a computer software (D-Flow 3.24, Motekforce Link, Amsterdam). Each perturbation consisted of a 0.043 m standing surface horizontal translation for 0.74 s with maximum velocity of 0.106 m/s and equal and opposite maximum acceleration of magnitude 0.374 m/s^2 (see Fig. 1).

A Bagnoli EMG System (Delsys, Inc) was used for acquisition of muscle activity following the perturbation onset. Any possible anticipatory muscle activation before the perturbation was not analyzed. Adhesive pre-gelled Ag/AgCl surface EMG electrodes (Delsys Inc., Boston, MA) were placed bilaterally on lower limb muscles: tibialis anterior (TA), soleus (Sol), fibularis longus (FL), rectus femoris (RF), biceps femoris (BF), gluteus maximus (Gmax), and gluteus medius (Gmed). The positioning of the electrodes was in accordance with Rainoldi et al. [16].

2.3. Experimental procedure

Participants performed three maximum voluntary isometric contractions (MVIC) for each muscle for the normalization of electromyography (EMG) peak amplitude (% EMG max). For the Sol and TA, MVICs plantar- and dorsi- flexion were performed against a strap secured over the foot (ankle joint 90 °; knee joint 90 °). For FL, MVIC foot eversion with plantar flexion against manual resistance over the foot and knee was performed. For RF, MVIC knee extension was assessed with the participant seated on a chair (hip joint 90 °; knee joint 90 °) with a cuff around the ankle. For BF, the participant lay on a table in prone position and performed MVIC knee flexion at 45 ° against manual resistance applied to the ankle. For Gmax, the participant was in prone position with legs extended (hip joint 0 °; knee joint 0 °) and performed MVIC hip extension against manual resistance applied to the ankle. For Gmed, the participants lay on their sides with the upper trunk and pelvis aligned in a straight line and performed MVIC hip abduction against manual external resistance at the lateral side of the knee [17].

Following the MVIC testing, participants took a 5–10 min of rest to recover from potential fatigue induced by the MVIC task [18]. Next, participants stood on a force-instrumented treadmill with three different IFPs (See Fig. 2). Before the perturbation trials, a 20-second quiet standing trial was recorded to evaluate baseline standing balance in each IFP. Participants were instructed to distribute body weight evenly over both legs. Next, participants performed three non-consecutive external lateral perturbation (left or right direction) trials for each of the three IFPs in random order with instruction to "please stand on the foot position asked for each trial and react naturally to maintain your balance. Try to maintain upright posture following perturbation until being asked to stop". GRFs underneath each foot were monitored by the tester and treadmill translation was delivered after visual confirmation of evenly distributed body weight over both legs. Perturbation onset latency and direction, and foot position for each trial were randomized to minimize prediction of perturbation parameters and order effects.

The boundaries of the three IFPs of each participant were marked on contact paper on the standing surface to ensure consistency of the foot position. Because this study focused on lower extremity muscle activation levels and patterns, participants were instructed to hold a light-weight stick in front of their chest to minimize arm movements [19].

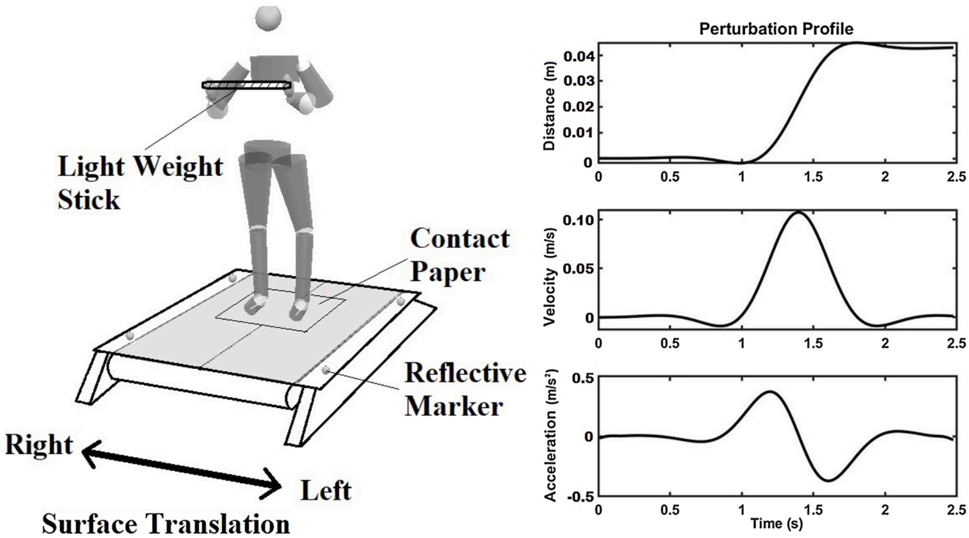

Fig. 1. Experimental setup for M-Gait system (left) and perturbation profile (characteristics of displacement, velocity, and acceleration) of a trial (right). The force platform-instrumented treadmill developed a lateral (left or right direction) surface translation. Participants stood on the contact paper on the treadmill with different initial foot position (IFP) for each trial. The boundaries of the three IFPs of each participant were marked on the contact paper to ensure consistent foot placement. Participants were instructed to hold a lightweight stick in front of their chest to minimize arm movements. The markers on the treadmill were used to determine the end of perturbation.

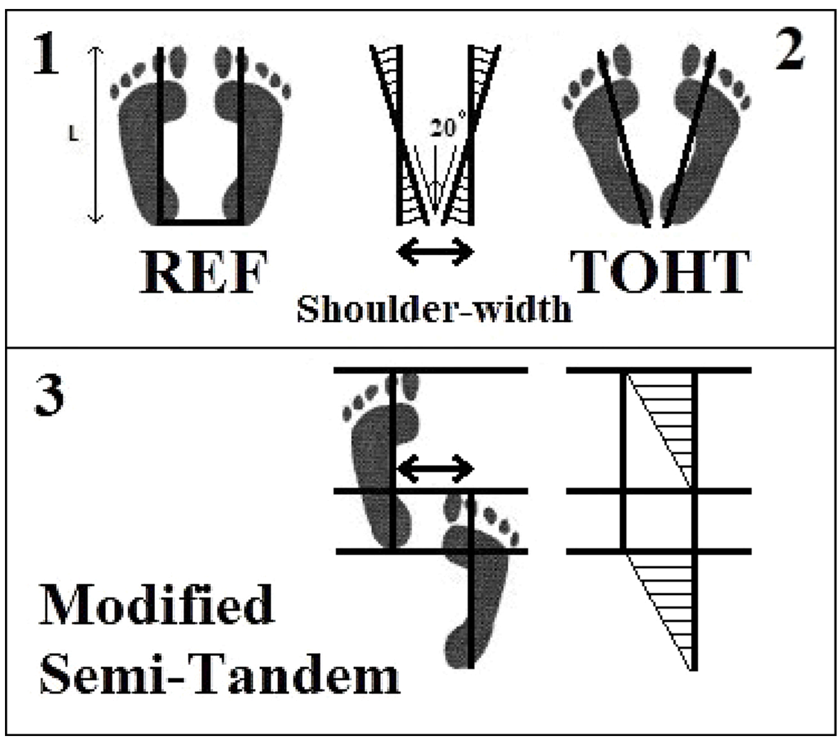

Fig. 2. Three different initial foot positions. REF: Reference foot position. The feet were placed shoulder-width apart and parallel to one another; TOHT: Symmetric toes-out angle of 20 ° with the same base of support area as REF; Modified Semi-Tandem (M-ST): The heel of the anterior foot was placed by the big toe of the posterior foot with both feet shoulder-width apart with the same base of support area as REF.

Participants were instructed to look straight ahead at an object projected on the screen before the perturbations and keep their arms as still as possible during the perturbations.

Three initial foot positions

1 Reference (REF): The feet were placed shoulder-width apart and parallel to one another symmetrically.
2 Toes-out with heels together (TOHT): Toes were turned out symmetrically with a toe-out angle of 20 ° in the frontal plane. The heels were positioned close together to achieve a similar total BoS area to REF.
3 Modified Semi-Tandem (M-ST): The heel of the anterior foot was placed by the big toe of the posterior foot with shoulder-width apart to achieve a total BoS area that was similar to REF. The posterior leg was chosen according to the participants' preference.

2.4. Data processing and analyses

2.4.1. Kinetic and kinematic data

Kinematic and kinetic data were Butterworth low-pass filtered at 6 Hz and 25 Hz, respectively [3,20]. The body's center of mass (CoM) trajectory was calculated using Nexus 1.8.5 Software (Vicon, Oxford Metrics, UK). GRFs in the A-P and M-L directions were calculated and normalized to body mass (kg). To measure the CoP sway area, the 95 % confidence ellipse area enclosed by trajectories underneath both feet was calculated and normalized to the time interval [21].

Standard deviation of the body's CoM acceleration (GRF / body mass), denoted as SD of CoMAccel [22], in the A-P and M-L directions and CoP sway area were used to quantify postural sway before perturbations (quiet standing period) and after the end of perturbations during the stabilization phase. The stabilization phase was defined as the time period between the end of perturbation and the beginning of the quiet standing phase. The beginning of the quiet standing phase was determined by the instant when the difference between the CoP and CoM was steady within ±0.5 cm in both the A-P and M-L directions [3,17].

The perturbation onset was defined as the instant when the platform marker acceleration exceeded zero [23]. The end of perturbation was defined as the moment in which the velocity of markers on the treadmill reached zero (see Fig. 1).

2.4.2. EMG

The surface EMG response after the onset of perturbations was filtered with a 20–450 Hz band-pass filter. A 2nd order Butterworth low pass filter with 20 Hz cutoff was applied as a digital smoothing algorithm after full wave rectification [24]. The EMG signals were then normalized to the MVIC EMG to obtain % EMG peak amplitude for each muscle in both legs. Trials were excluded if anticipatory muscle contractions were observed before perturbation onset. Each muscle's onset time following perturbations was determined as the time at which the EMG exceeded three standard deviations (SD) of the mean baseline during a quiet standing trial [25].

The average of three trials for each IFP was calculated for all data analysis. All data were analyzed in Matlab 9.3 (Matworks Inc., Natick, MA, USA).

2.5. Statistical analysis

A three-way mixed-model design (within: muscles and IFPs × between: age) ANOVA was used to analyze the effects of muscles, IFPs, and age on EMG peak amplitude, peak timing, and burst duration. To quantify postural sway during balance recovery, a two-way mixed-model analysis (within: IFPs × between: age) of covariance (ANCOVA) was used to compare SD of CoMAccel and CoP sway area using weight and height as covariates [26]. A Tukey's test was used for post hoc analysis. SPSS (Chicago, IL) was used for all statistical analysis with an alpha level of 0.05 set a-priori.

3. Results

All data are presented as mean ± SD in the text and tables. None of the participants needed to step to recover their balance across any IFPs over all trials. In addition, no trials were excluded due to anticipatory EMG activities.

3.1. EMG

3.1.1. EMG peak amplitude

There was a main effect of muscles on the EMG peak amplitude (muscles: $F (6, 413) = 29.84$, $p < 0.01$). Different levels of EMG peak amplitude during the perturbation phase (Fig. 3) were observed in the loaded leg muscles (the leg opposite to the perturbation direction). EMG peak amplitude of the RF, BF, and Gmax was significantly smaller compared to TA, Sol, FL, and Gmed during balance recovery in all IFPs (all $p < 0.01$) (Fig. 4).

There was a main effect of IFPs on the EMG peak amplitude (IFPs: $F (2, 417) = 6.92$, $p = 0.01$). EMG peak amplitude in TOHT was greater than REF ($p = 0.01$). There was an interaction between muscles and IFPs (muscles * IFPs: $F (12, 399) = 3.66$, $p = 0.04$). The Sol and FL EMG peak amplitude in the loaded leg was greater in TOHT (Sol: $p < 0.01$, FL: $p = 0.04$) compared to REF. There was an interaction between muscles and age (muscles * age: $F (6, 406) = 2.98$, $p = 0.04$). Older adults demonstrated reduced EMG peak amplitude of the loaded FL muscle compared to younger adults ($p = 0.04$, see Fig. 3).

There was no main effect of age and no interaction between age and IFPs on the EMG peak amplitude.

3.1.2. EMG peak timing and burst duration

There was a main effect of age on the EMG peak timing (age: $F (1, 418) = 4.73$, $p = 0.04$). EMG peak timing in older adults occurred later than younger adults. Interaction was also observed between age and muscles (age * muscles: $F (6, 406) = 3.08$, $p = 0.04$). Older adults showed a delayed FL peak timing compared to younger adults ($p = 0.01$, see Fig. 5). There was no main effect of muscles and IFPs, and no interaction between age and IFPs or between IFPs and muscles for EMG peak timing.

There was no main effect and no interaction on muscles, IFPs and age for EMG burst duration.

3.2. Postural sway before and after perturbations

3.2.1. Quiet standing balance before perturbations

There was a main effect of IFPs on SD of CoMAccel in the A-P direction (IFPs: $F (2, 57) = 3.792$, $p = 0.029$). Post hoc analysis showed that SD of CoMAccel in the A-P direction in M-ST was greater than REF ($p = 0.027$). No interaction was observed in the A-P direction.

In the M-L direction, there was a main effect of IFPs and age on SD of CoMAccel (IFPs: $F (2, 57) = 12.99$, $p < 0.01$, age: $F (1, 58) = 5.40$, $p = 0.03$). Post hoc analysis showed that M-L SD of CoMAccel was greater for M-ST than other two IFPs (both $p < 0.01$) and was greater for older adults ($p = 0.03$). An interaction between IFPs and age for M-L SD of CoMAccel was detected (IFPs * age: $F (2, 57) = 7.77$, $p < 0.01$). Older adults showed increased M-L SD of CoMAccel in M-ST compared to younger adults ($p < 0.01$).

There were main effects of IFPs and age on CoP sway area (IFPs: $F (2, 57) = 12.34$, $p < 0.01$, age: $F (1, 58) = 14.77$, $p < 0.01$). Post hoc analysis showed that the CoP sway area was greater in M-ST compared to TOHT ($p < 0.01$) and REF ($p < 0.01$). Older adults showed greater CoP sway area than younger adults ($p < 0.01$). An interaction between IFPs and age for CoP sway area was detected (IFPs * age: $F (2, 57) = 12.20$, $p < 0.01$). Older adults showed greater CoP sway area in REF and M-ST than younger adults (both $p < 0.01$) (Table 1).

3.2.2. Post-perturbations

There was a main effect of IFP and age for SD of CoMAccel in the A-P (IFPs: $F (2, 57) = 4.33$, $p = 0.02$, age: $F (1, 58) = 4.41$, $p = 0.04$) and M-L (IFPs: $F (2, 57) = 3.82$, $p = 0.03$, age: $F (1, 58) = 5.29$, $p = 0.03$) directions. Post hoc analysis showed that SD of CoMAccel was smaller for TOHT than M-ST in both the A-P ($p = 0.03$) and M-L ($p = 0.02$) directions. Older adults showed greater SD of CoMAccel than younger adults in both the A-P ($p = 0.04$) and M-L ($p = 0.03$) directions. No interaction was observed for A-P and M-L SD of CoMAccel.

There were main effects of IFPs and age for CoP sway area (IFPs: $F (2, 57) = 5.10$, $p = 0.02$, age: $F (1, 58) = 4.75$, $p = 0.04$). Post hoc analysis showed that the CoP sway area was greater in M-ST than REF ($p = 0.03$) and TOHT ($p = 0.02$). Increased CoP sway area was observed in older adults compared to younger adults ($p = 0.04$) and no interaction was detected.

4. Discussion

The purpose of this study was to investigate activation level differences between lower extremity muscles and the effects of IFPs on muscle activation patterns and postural stability in response to unexpected lateral surface perturbations in old and younger adults. We found that activation levels of ankle muscles were greater than knee and hip muscles across IFPs. Compared to younger adults, older adults had greater postural sway in the M-L direction in M-ST before perturbation. Following perturbation, older adults demonstrated decreased FL muscle

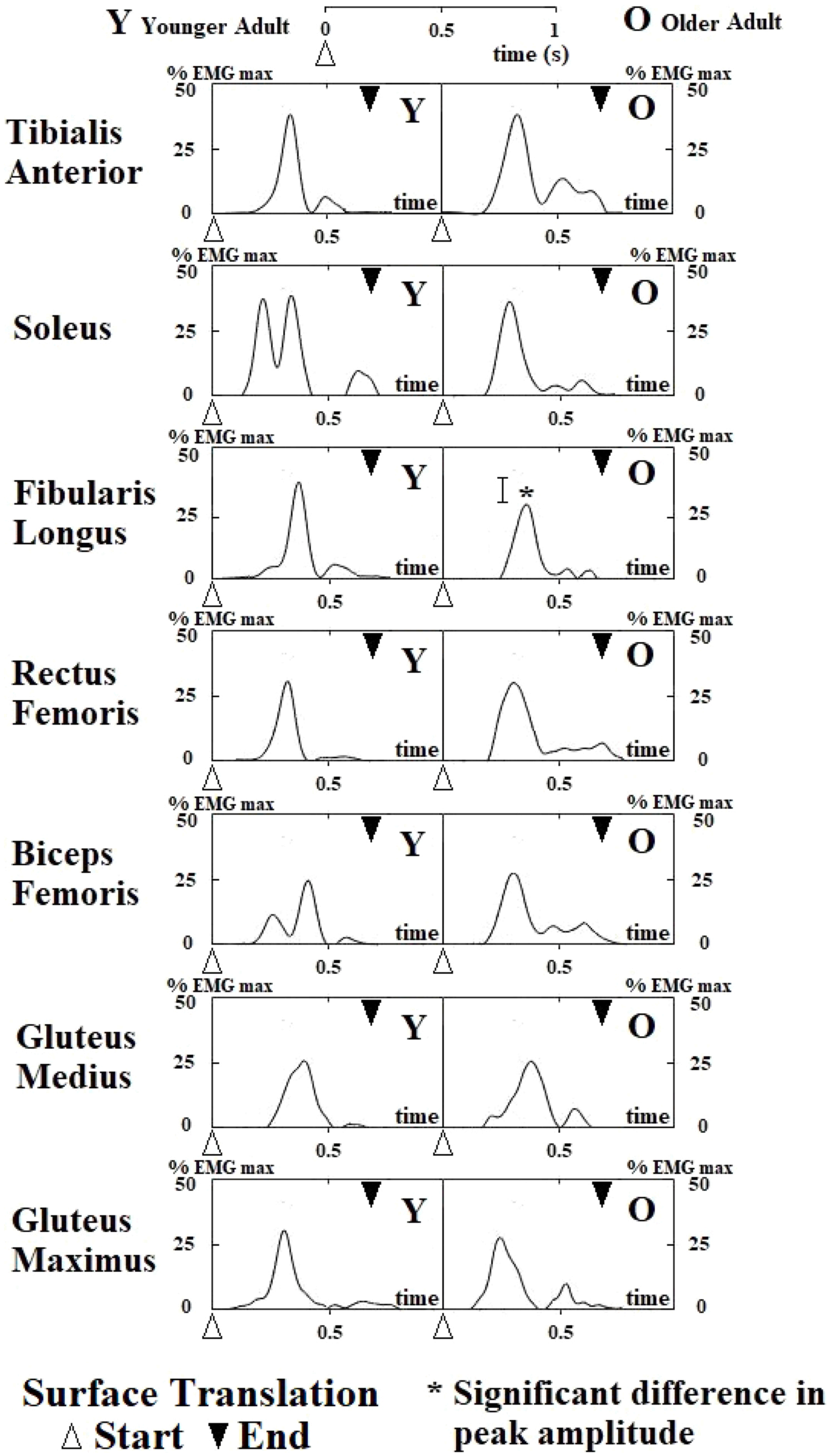

Fig. 3. Normalized EMG activation (% EMG max) of all muscles in the loaded leg during balance recovery in toes-out with heels together initial foot position from representative younger and older adults. * represent a statistically significant difference ($p < 0.05$).

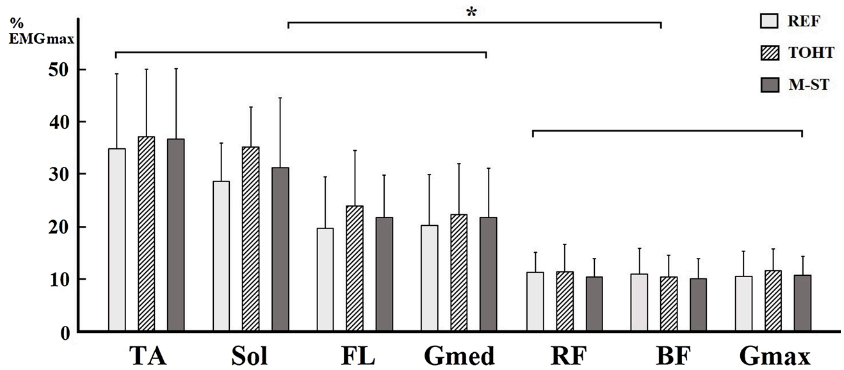

Fig. 4. Normalized EMG activation (% EMG max) of all muscles in the loaded leg across all initial foot positions. * represent a statistically significant difference ($p < 0.05$) between the muscles. TA: tibialis anterior, Sol: soleus, FL: fibularis longus, RF: rectus femoris, BF: biceps femoris, Gmax: gluteus maximus, and Gmed: gluteus medius. REF: Reference, TOHT: toes-out heels together, M-ST: modified semi-tandem (M-ST) initial foot positions.

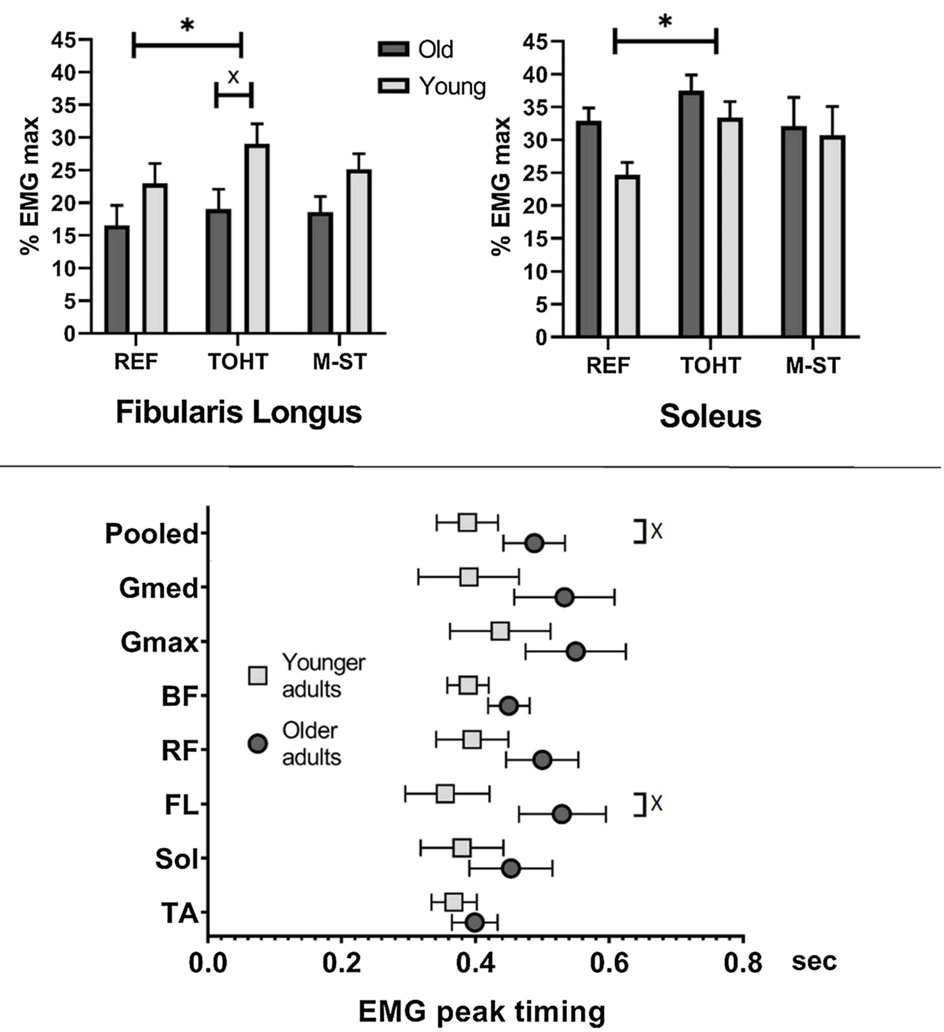

Fig. 5. EMG activation of the fibularis longus and soleus muscles (percentage of maximum voluntary contraction, %EMGmax) (top panel) and EMG peak timing (bottom panel) in older and younger adults during standing balance recovery. * represent a statistically significant difference ($p < 0.05$) between two initial foot positions. ^{x} represent a statistically significant difference ($p < 0.05$) between younger and older adults. REF: reference foot position, TOHT: toes-out with heels together, M-ST: modified semi-tandem. TA: tibialis anterior, Sol: Soleus, FL: fibularis longus, RF: rectus femoris, BF: biceps femoris, Gmax: Gluteus maximus, Gmed: gluteus medius, Pooled: a combined set of all seven muscles. The error bars represent the standard error.

activation with greater postural sway in both the A-P and M-L directions across all IFPs. In addition, TOHT showed the smallest M-L postural sway with increased activation of the Sol and FL during the stabilization phase.

4.1. Muscle activation level differences

We observed greater EMG peak amplitude at the ankle (TA, Sol, and FL) and Gmed muscles compared to knee muscles (RF and BF) and Gmax in the loaded leg during standing balance recovery. This could be due to the relatively low perturbation intensity used in our study. Previous studies showed that distal muscles at the ankle joint provide immediate joint torque to regain balance following surface perturbations [27] and proximal muscles at the hip joint are recruited as the intensity of perturbation increases [3]. Thus, participants in our study could achieve balance recovery by increasing ankle joint muscles and Gmed activation

Table 1
Postural Sway before and after lateral perturbations.

IFP	Group	SD of CoM Acceleration (mm/s^2), Fx/m (A-P) Pre	Post	SD of CoM Acceleration (mm/s^2), Fy/m (M-L) Pre	Post	CoP Sway Area (mm^2/s) Pre	Post
REF	Young	14.0 ± 1.63	43.0 ± 7.46	7.00 ± 1.11	100.0 ± 12.3	2.29 ± 0.27	74.04 ± 5.13
	Old	22.0 ± 4.10	97.5 ± 21.3[*]	16.5 ± 4.28	214.5 ± 47.9[*]	4.62 ± 0.48[*]	90.04 ± 13.44
	Total	18.0 ± 2.33[c]	70.3 ± 12.6	11.8 ± 2.41	157.3 ± 27.4	3.46 ± 0.38[c]	82.04 ± 7.24[c]
TOHT	Young	26.5 ± 4.02	42.0 ± 6.59	14.5 ± 2.17	88.5 ± 11.0	4.67 ± 0.58	65.49 ± 4.89
	Old	27.5 ± 9.87	80.0 ± 15.1[*]	28.5 ± 7.92	122.0 ± 18.7	6.22 ± 0.69	97.99 ± 10.95[*]
	Total	27.0 ± 5.19	61.0 ± 9.15[c]	21.5 ± 4.31	105.3 ± 11.2[c]	5.44 ± 0.47[c]	81.74 ± 6.92[c]
M-ST	Young	0.02 ± 0.01	91.7 ± 7.05	9.00 ± 1.00	160.5 ± 13.3	6.18 ± 0.75	78.09 ± 6.95
	Old	0.07 ± 0.06[*]	266.5 ± 28.7	35.5 ± 5.80[*]	271.5 ± 81.4	18.95 ± 3.47[*]	114.92 ± 43.77[*]
	Total	44.3 ± 10.7[a]	179.1 ± 17.7[b]	22.3 ± 4.18	216.0 ± 42.1[b]	12.56 ± 2.26[a,b]	96.50 ± 13.84[a,b]

Mean ± Standard Error. SD (standard deviation). m (mass), CoM (center of mass). CoP (center of pressure). REF (reference), TOHT (toes-out with heels together), M-ST (modified semi-tandem), A-P (anterior-posterior), M-L (medio-lateral).

[a] Significantly different than REF ($p < 0.05$).
[b] Significantly different than TOHT ($p < 0.05$).
[c] Significantly different than M-ST ($p < 0.05$).
[*] Significantly different than younger adults ($p < 0.05$).

without greater involvement of muscles at the knee muscles and Gmax. Although ankle joint muscles provide immediate balance recovery responses as the closest responders to surface perturbation [3], relatively greater degree of muscle activation might be required to produce sufficient ankle joint torque due to the smaller moment arms and physiological cross-sectional areas of the ankle muscles compared to proximal muscles at the knee and hip joints [28]. These results highlighted the importance of neuromuscular control of the ankle muscles because it appeared to be one of the first responders to increased postural sway [29]. In addition, greater activation level of the Gmed compared to muscles at the knee joint and Gmax confirmed its active role in control of medio-lateral postural stability. This is consistent with previous studies that have shown a significance relation between Gmed muscle activity and balance recovery following external lateral perturbations [30,31]. The hip abd/addutor, trunk, and ankle musculature are key factors that contribute to the control of CoM motion in the frontal plane in response to a lateral balance loss [3,32]. Indeed, our finding demonstrated that the Gmed in the loaded leg increased its activation level even during feet-in-place balance regulation.

4.2. Effects of IFP on standing balance control in old and younger adults

4.2.1. Modified semi-tandem IFP

Compared to REF, M-ST showed greater A-P and M-L SD of CoMAccel and CoP sway area prior to perturbation onset and greater CoP area during the stabilization phase. During REF, A-P balance is primarily controlled by dorsi- and plantar-flexion at the ankle joint, while hip adduction and abduction contributes to M-L balance control [3]. During M-ST, however, balance control in each direction is normally achieved by simultaneous control over both ankle and hip joints [33]. Our result demonstrated that EMG peak timing of the FL is delayed in M-ST compared to REF. Thus, more challenging inter-joint coordination and relatively slow ankle movement control appeared to result in greater postural sway during M-ST compare to REF. Considering older adults commonly show reduced ability to re-distribute lower-limb joint moments [34,35], difficulties in inter-joint coordination in older adults likely contributed to greater postural sway during M-ST. Indeed, our findings suggested that M-ST is a more challenging IFP for older adults to maintain postural stability during quiet standing. This result warrants further investigation on whether balance training in older adults with more challenging posture could lead to better outcomes.

4.2.2. Toes-out IFP

During the stabilization phase, toes-out IFP showed smaller A-P and M-L SD of CoMAccel compared to M-ST, whereas there was no significant difference between REF and M-ST. We found that the activation of the Sol and FL was enhanced in TOHT following perturbation onset. The FL muscle stabilizes the ankle joint of the loaded leg by everting the foot to prevent increasing lateral plantar pressure induced by lateral perturbations. Therefore, it provides more evenly distributed plantar pressure under the foot and this may result in improved M-L balance [36]. The Sol and FL generate ankle plantar flexion moment that typically counteracts the forward toppling of the body's CoM to maintain A-P balance [37]. During toes-out foot position, ankle plantar/dorsi flexion moments also contribute to GRFs in the M-L direction that are responsible for controlling the lateral motion of the body's CoM [38]. Thus, TOHT could provide a biomechanically advantageous foot position for maintaining postural stability following perturbations and it is plausible that greater activation of the Sol and FL in toes-out IFP might have contributed, at least in part, to smaller SD of CoM Accel in both the A-P and M-L directions during the stabilization phase compared to other IFPs. Results from this study showed that older adults had delayed peak timing and decreased activation of the FL muscle following perturbations compared to younger adults, reflecting age-related abnormalities in neuromuscular control for eversion of the ankle during balance recovery in response to external lateral perturbation. Because Sol and FL muscle activation was increased during TOHT, modifying IFP could be a viable approach for rehabilitation assessments and interventions focused on improving balance control in older populations. Our results indicate that toes-out IFP could be a viable option to better engage ankle muscles while postural stability training in older adults.

4.3. Limitations of this study

Although the order of IFPs and perturbation onset latency and direction were randomized, postural reactions during subsequence trials may be different from the first trial and the potential adaptation effects may not be fully removed [39]. Thus, our results may not be generalizable for first trial responses.

In addition to lower extremity muscles, lower back and abdominal muscles control the pelvis and trunk position and, therefore, contribute to maintaining core stability during standing balance recovery [40]. Thus, the observed differences in the FL and Sol muscles during toes-out may not be the only factors that affect postural stability. Future research should evaluate the influence of upper extremity muscle activity in relation to balance control following unexpected lateral perturbations.

5. Conclusions

During standing balance recovery from unexpected lateral surface perturbations, activation levels of ankle muscles were greater than knee and hip muscles. Compared to younger adults, older adults had reduced

FL activation and greater postural sway. Toes-out IFP appeared to increase activation of Sol and FL muscles and reduce M-L postural sway. These findings provide useful information for designing potential rehabilitation approaches that modify IFP to engage ankle muscles to improve dynamic balance control in older adults.

Declaration of Competing Interest

The authors report no declarations of interest.

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