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Effects of localized muscle fatigue on recovery from a postural perturbation without stepping

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ABSTRACT

Several investigations have demonstrated that localized muscle fatigue (LMF) causes an increase in postural sway measures during quiet stance. Since many falls are likely the result of a postural perturbation, this study investigated the effects of LMF on balance recovery from sagittal plane postural perturbations. Thirty-two participants (16 young, 16 older) were tested. Postural perturbations were administered with ballistic pendulums before and after exercises to fatigue the lumbar extensors and plantar flexors. Measures of balance recovery were based on the center of pressure (COP) and center of mass (COM) trajectories and the maximum perturbation that could be withstood without stepping. A covariate analysis that included initial conditions at the time of the perturbation was also performed. The results demonstrated changes in the COM trajectory that were consistent with an LMF-induced decrement in the ability to recover from the perturbations without stepping. Interpretation of the COP trajectory was presented in light of the COM and indicated a modified postural control strategy following LMF.

1. Introduction

Localized muscle fatigue (LMF) increases center of pressure (COP)-based measures of postural sway during quiet standing [1–5]. For example, Corbeil et al. [3] reported increases in COP mean velocity, mean radius, and median frequency following plantar flexor fatigue. Gribble and Hertel [4] found increases in COP velocity following LMF of the plantar flexors, knee extensors, and hip flexors, separately. Based on evidence that increased postural sway is linked to an increased risk of falling (albeit among older adults) [6,7], these findings may indicate that LMF increases the risk of falling. Although a direct link between fatigue and increased falls has not been demonstrated experimentally, epidemiological studies have suggested such a link [8,9].

Quiet standing is not a particularly challenging task under normal circumstances for most individuals. Because of this, many falls are likely precipitated by postural perturbations. Investigating the effects of LMF on the ability to recovery from a postural perturbation may thus afford improved external validity over studies investigating the effects of LMF on quiet standing. Furthermore, the increased challenge and larger kinematic range induced by perturbations may provide additional insight into the effects of LMF on balance. To our knowledge, no studies have investigated the effect of LMF on the ability to recover from a postural perturbation. Wilson et al. [10], however, reported a shift from the so-called "ankle strategy" after a postural perturbation to more of a "hip strategy" following lumbar extensor fatigue. This shift may represent a neuromuscular adaptation to mitigate potentially deleterious effects of LMF on postural control, but it was unclear how the ability to recover from a postural perturbation was affected.

Therefore, the goal of this study was to investigate the effects of LMF on balance recovery (BR) following a postural perturbation. Both young and older subject groups were included, and two different muscle groups were fatigued, to also investigate any interactive effects of LMF with age and fatigued muscle group. We hypothesized that LMF would produce changes indicative of a diminished ability to recover from a postural perturbation, and that age would exacerbate these changes. We hypothesized that LMF would cause increases in COM-based variables such as peak displacement and peak velocity. Some of these changes would have a strong interaction with age, indicating greater changes in

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the older population. We also hypothesized that COP-based variables would not reveal similar patterns to the COM, but could be consistently interpreted considering changes in the COM-based variables.

2. Methods

Thirty-two participants were recruited from the local community including 16 young (age mean $\pm$ S.D. = 19.4 $\pm$ 1.4 years, mass = 71.4 $\pm$ 11.1 kg, stature = 174.8 $\pm$ 8.3 cm) and 16 older adults (age = 62.2 $\pm$ 5.1 years, mass = 74.0 $\pm$ 10.9 kg, stature = 167.8 $\pm$ 8.9 cm). Each age group had an equal number of males and females. Participants were screened for self-reported musculoskeletal disorders and medications that could affect balance. In addition, older participants were required to pass a medical exam to exclude those with neurological, cardiac, respiratory, vestibular, or musculoskeletal disorders, or any falling incident within the past year. This experiment was approved by the Virginia Tech Institutional Review Board, and participants provided informed consent prior to participation.

Participants visited the laboratory for two experimental sessions separated by approximately 1 week. In each session, participants underwent a series of postural perturbations both before and after fatiguing exercises. Both sessions were identical except that in one session the ankle plantar flexors were fatigued, and in the other session the lumbar extensors were fatigued. Presentation order of the two fatiguing exercises was counterbalanced.

Prior to the perturbations the participants were instructed to "stand in a relaxed manner" with their feet together, eyes closed, and hands clasped together behind their back to eliminate upper extremity movement [11]. Perturbations were administered with padded pendulums (mass $\approx$ 13 kg) positioned in the front and back of the participants (Fig. 1). To apply a perturbation, both pendulums were pulled away from the participants in the median plane, and one was released so as to swing in a ballistic manner until impact. The release point was selected to achieve a specified pendulum velocity just before impact. Perturbation magnitude was defined as the linear momentum just before impact. The rear pendulum was used to administer anteriorly-directed (AD) perturbations and impacted the body just inferior to the scapula [12]. The front pendulum was used to administer posteriorly-directed (PD) perturbations and impacted the body just inferior to the jugular notch of the sternum. Each series of perturbations consisted of equal numbers of AD and PD perturbations, ordered randomly, to prevent anticipation of perturbation direction. Only results for AD perturbations are described here. Earmuffs were worn by the participants to eliminate auditory cues to a perturbation.

The experiment began with an initial series of 20 low magnitude perturbations (10 N s AD, 7 N s PD) that were small enough such that participants could recover their balance without stepping (Fig. 2). Exposing the participants to these initial perturbations allowed any adaptation in BR performance to occur prior to investigating the effects of LMF. Next, a series of AD and PD perturbations beginning at 6 N s and 5 N s, respectively, were applied. Following a successful recovery without stepping, the magnitude was increased by 2 N s for AD and 1 N s for PD perturbations. This was continued until two stepping responses were elicited for a given magnitude. Using this approach, the largest perturbation that could be withstood without stepping was determined. Sixteen constant-magnitude perturbations were then administered at 4 N s and 2 N s below the maximum AD and PD perturbations that could be withstood without stepping, respectively.

Fig. 1. Participants were perturbed in the anterior and posterior directions with padded and weighted ballistic pendulums. Perturbations were administered by pulling the pendulums away from the participant and releasing from a fixed location to yield the desired perturbation magnitude.

The fatiguing protocol involved multiple sets of dynamic exertions designed to fatigue the participants to a desired fatigue level over a fixed duration [1,10]. A fixed duration was used due to the potential of fatigue time to modulate the effects of LMF on balance [13]. The protocol began by obtaining isometric maximum voluntary contractions (MVC) of the targeted muscle group (ankle plantar flexors or lumbar extensors). All MVCs and fatiguing exercises were performed on a seated calf-raise device (New York Barbell, Elmira, NY) for the ankle plantar flexors, or on a Biodex System 3 Pro dynamometer (Biodex Medical Systems Inc., Shirley, NY) for lumbar extensors. Ankle plantar flexor MVCs were performed with the ankle in the anatomical position, and lumbar extensor MVCs while the lumbar spine was flexed 45°. Participants performed one set of concentric contractions of the ankle plantar flexors or lumbar extensors each minute through the duration of the fatiguing protocol. Repetitions were performed at a rate of 23 min^{-1} with resistance set at 45% of the unfatigued MVC. Every 2 min an isometric MVC was performed and the number of repetitions in each set was adjusted in an attempt to decrease the MVC in a linear fashion to 70% of the unfatigued MVC over 14 min of exercise. If the MVC had not dropped below 70% by the end of 14 min, 2 min of exercise were added along with an increase in repetitions. This process was repeated until the participants were fatigued to 70% of their unfatigued MVC. Preliminary work indicated that the MVCs remained below 75% of the unfatigued MVC for 4–5 min following the fatiguing protocol and then slowly increased. After 20 min of rest, MVCs typically range from 80% to 90% of the unfatigued MVC. Immediately following the fatiguing protocol, the maximum perturbation that could be withstood without stepping was again determined in the same manner described above, and 16 fatigued perturbations with the same magnitude as the unfatigued perturbations were administered. All perturbations were administered within 3 min following completion of the fatiguing protocol.

Whole-body kinematic data and ground reaction data were collected during all trials using a Vicon 460 Motion Analysis System (Lake Forest, CA) and a six degree-of-freedom force platform (Model OR6-5, Advanced Mechanical Technology Inc., Watertown, MA), respectively. Participants were instrumented with 16 reflective markers placed at selected anatomical locations. Marker position data were sampled at 100 Hz and low-pass filtered at 5 Hz (4th order zero-phase-lag Butterworth). Bilateral marker positions were averaged across the left and right sides of the body to create a sagittal plane representation. A six-segment kinematic model (feet, shanks, thighs, pelvis, torso/arms, head) with anthropometrically-correct inertial parameters [14,15] was used to approximate the trajectory of the body center of mass (COM). Ground reaction forces and moments were sampled at 1000 Hz, low-pass filtered at 7 Hz (4th order zero-phase-lag Butterworth), and transformed to obtain the COP trajectory in the anteroposterior direction [16]. An in-line load cell (Cooper Instruments and Systems, Warrenton, VA) attached to the pendulum was used to identify perturbation onset time, which was the time at which the load cell force exceeded two standard deviations above the baseline mean.

Measures of BR included the maximum perturbation that could be withstood without stepping, and descriptors of the COM and COP trajectories following a postural perturbation. These descriptors included peak displacement relative to initial position, time-to-peak displacement, peak velocity in the anterior direction, time-to-peak velocity [12], minimum time-to-boundary [17], and time-to-return to within 20% of peak displacement relative to initial position [18]. COM- and COP-based displacement measures were normalized by ankle-to-toe length. COM- and COP-based velocity measures were normalized by multiplying by participant mass (yielding units of momentum) to account for effects of inertia on these measures.

Between-session reliability of the BR measures was assessed using intraclass correlation coefficients (ICC) model (2,1) based on single measurements [19] using the pre-fatigue data in each session. A repeated measures analysis of variance (ANOVA) was used to determine the effects of fatigue (unfatigued, fatigued), muscle (ankle plantar flexor, lumbar extensor), and age (young, older) on BR measures. Only interactions of fatigue $\times$ muscle and fatigue $\times$ age were included as higher order effects in the statistical model. Additionally, variability of perturbation magnitude and instantaneous kinematics of the COM and COP trajectories at the instant of pendulum contact were accounted for in the statistical model by including them as covariates. These instantaneous kinematic parameters were

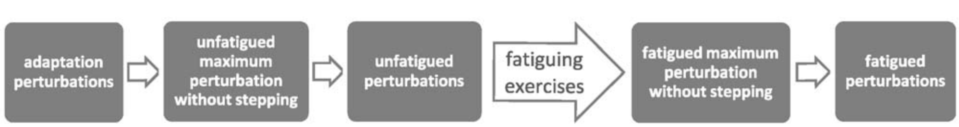

Fig. 2. Overall schematic of experimental protocol. Participants underwent five separate series of perturbations with an intervening fatiguing exercise of the ankle plantar flexors or lumbar extensors.

Table 1
Intraclass correlations (ICC) and least squares mean ± standard error for each balance recovery measure categorized by fatigue level and age group. Significant main and interaction effects are designated using the symbols noted.

Dependent measure	ICC	Young		Older	
		Unfatigued	Fatigued	Unfatigued	Fatigued
Stepping response					
Max. perturbation magnitude (N s)[a]	0.91	7.41 ± 0.14	7.38 ± 0.14	6.31 ± 0.14	5.88 ± 0.14
Center of mass					
Peak displacement (%)[b]	0.97	26.1 ± 0.3	26.6 ± 0.3	25.2 ± 0.3	25.9 ± 0.3
Time-to-peak displacement (ms)[b]	0.93	591.0 ± 7.9	608.4 ± 8.4	538.1 ± 8.5	567.2 ± 8.1
Peak velocity (N s)[b,a]	0.99	16.57 ± 0.08	16.65 ± 0.08	17.96 ± 0.08	18.09 ± 0.08
Time-to-peak velocity (ms)[a]	0.96	145.8 ± 0.8	146.7 ± 0.9	139.5 ± 0.9	139.5 ± 0.8
Min. time-to-boundary (ms)	0.93	572.3 ± 10.2	57.09 ± 10.7	587.9 ± 10.5	585.6 ± 9.9
Return 20% (ms)[b]	0.82	1336 ± 27	1334 ± 28	1295 ± 29	1369 ± 28
Center of pressure					
Peak displacement (%)[c,a]	0.98	47.1 ± 0.3	45.4 ± 0.3	52.5 ± 0.3	51.1 ± 0.3
Time-to-peak displacement (ms)	0.87	346.1 ± 7.9	350.1 ± 8.4	378.8 ± 8.5	369.1 ± 8.1
Peak velocity (N s)[d,c]	0.97	66.69 ± 0.83	66.23 ± 0.88	75.06 ± 0.89	76.81 ± 0.84
Time-to-peak velocity (ms)[d]	0.97	148.0 ± 0.9	142.6 ± 0.9	144.7 ± 0.9	141.3 ± 0.9
Min. time-to-boundary (ms)[c]	0.95	118.8 ± 2.5	122.1 ± 2.6	102.3 ± 2.6	102.6 ± 2.4
Return 20% (ms)[d]	0.91	1096 ± 22	1115 ± 23	987 ± 23	1071 ± 21

[a] Significant main effect of age.
[b] Significant main effect of fatigue.
[c] Significant fatigue × muscle interaction.
[d] Significant fatigue × age interaction.

tested for effects of fatigue and age using a repeated measures ANOVA. Effects were considered statistically significant when $p \leq 0.05$. Significant interactions were further examined using Tukey HSD pair-wise comparisons. Effect size was quantified using Hedge's g [20] where $g = 0.2$, 0.5, and 0.8 were qualitatively interpreted as small, medium, and large effect sizes, respectively.

3. Results

Several measures of BR were affected by LMF (Tables 1 and 2). Four of the six COM-based measures were significantly affected by LMF including a 2.7% increase in peak COM displacement ($p < 0.001$, Hedge's $g = 0.32$), a 4.1% increase in time-to-peak COM peak displacement ($p < 0.001$, $g = 0.47$), a 0.6% increase in peak COM velocity ($p = 0.011$, $g = 0.22$), and a 3.5% increase in time-to-return within 20% of peak COM displacement ($p = 0.002$, $g = 0.28$). In addition, two COP-based measures were affected by LMF: a 3.1% decrease in peak COP displacement ($p < 0.001$, $g = 0.71$) and a 3.0% decrease in time-to-peak COP velocity ($p < 0.001$, $g = 0.75$). The 3.6% decrease in maximum perturbation that could be withstood without stepping with LMF was not significant ($p = 0.086$), but displayed a noteworthy effect size ($g = 0.43$).

Several fatigue × age interactions and fatigue × muscle interactions were observed on the COP-based measures (Fig. 3). Peak COP velocity had a significant fatigue × age interaction ($p = 0.017$) due to a 2.3% increase with LMF in the older group ($g = 0.15$) and no change in the young group ($g = 0.04$). Peak COP velocity also had a significant fatigue × muscle interaction ($p < 0.001$), due to a 3.6% increase with LMF of the lumbar extensors ($g = 0.44$) but no change after LMF of the ankle plantar flexors ($g = 0.20$). Minimum COP time-to-boundary also demonstrated a significant fatigue × mus- muscle interaction ($p = 0.003$), with a 5.2% increase with LMF of the plantar flexors ($g = 0.35$) and no change after LMF of the lumbar extensors ($g = 0.13$). Time-to-return within 20% of peak COP displacement had a significant fatigue × age interaction ($p = 0.006$) due to an 8.5% increase with LMF in the older group ($g = 0.29$), and no change in the young group ($g = 0.07$).

Table 2
Results of the statistical tests including two-way interactions of fatigue × age and fatigue × muscle. Effect size was calculated using Hedge's g, and is displayed for each main effect of fatigue or age.

Dependent measure	Interactions		Main effects and effect size			
	Fatigue × age	Fatigue × muscle	Fatigue		Age	
Stepping response						
Max. perturbation magnitude (N s)	$p = 0.196$	$p = 0.517$	$p = 0.086$	$g = 0.43$	$p = 0.029$*	$g = 0.57$
Center of mass						
Peak displacement (%)	0.471	0.420	<0.001*	0.32	0.596	0.05
Time-to-peak displacement (ms)	0.173	0.998	<0.001*	0.47	0.095	0.14
Peak velocity (N s)	0.540	0.415	0.011*	0.22	0.006*	0.24
Time-to-peak velocity (ms)	0.300	0.712	0.257	0.10	0.042*	0.17
Min. time-to-boundary (ms)	0.932	0.915	0.734	0.03	0.634	0.04
Return 20% (ms)	0.061	0.633	0.002*	0.28	0.845	0.02
Center of pressure						
Peak displacement (%)	0.409	0.041*	<0.001*	0.68	0.004*	0.26
Time-to-peak displacement (ms)	0.109	0.785	0.513	0.06	0.259	0.09
Peak velocity (N s)	0.017*	<0.001*	0.139	0.10	0.041*	0.17
Time-to-peak velocity (ms)	0.034*	0.542	<0.001*	0.81	0.612	0.04*
Min. time-to-boundary (ms)	0.239	0.003*	0.176	0.12	0.070	0.15
Return 20% (ms)	0.006*	0.547	<0.001*	0.39	0.241	0.10

* A statistically significant ($p \leq 0.05$) effect.

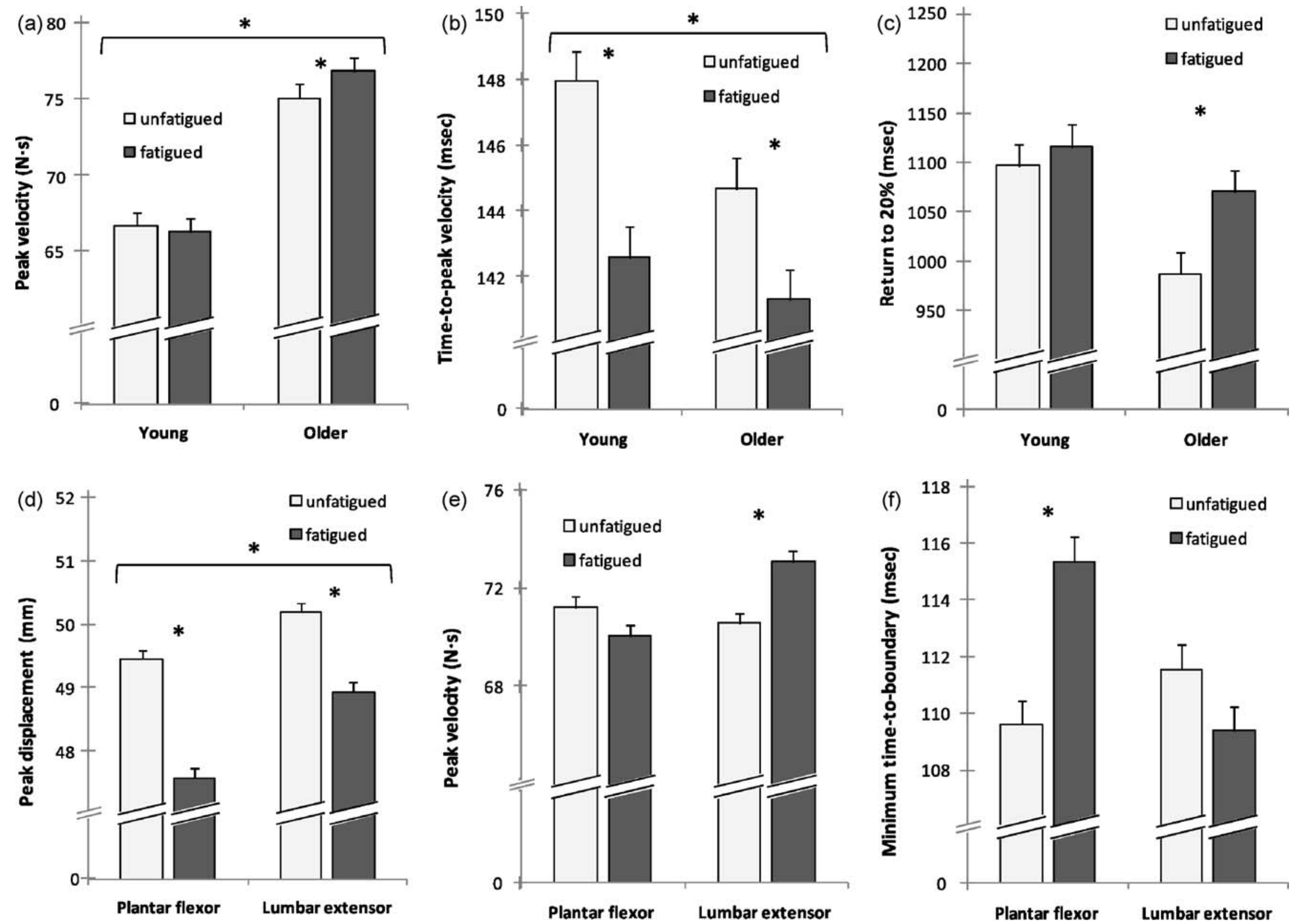

Fig. 3. Least square means (with standard error) of COP-based BR measures that exhibited fatigue × age and fatigue × muscle interactions. *Significant difference ($p \leq 0.05$).

Several measures of BR were affected by age (Tables 1 and 2). The maximum perturbation that could be withstood without stepping was 17.8% lower among the older adults ($p = 0.029$, $g = 0.57$). Older adults exhibited an 8.6% higher peak COM velocity ($p = 0.006$, $g = 0.31$) and a 4.6% lower time-to-peak COM velocity ($p = 0.042$, $g = 0.23$). In COP-based measures, older adults exhibited a 12.0% higher peak COP displacement ($p = 0.004$, $g = 0.35$) and a 14.2% higher COP velocity ($p = 0.041$, $g = 0.17$).

The range of the ICCs calculated for the unfatigued COM- and COP-based measures of BR between sessions was 0.82–0.99 (Table 1). ICC of the maximum perturbation was 0.91. COM- and COP-based positions at the instant of pendulum contact were 3.2–48.2% and 2.7–48.5% of foot length, respectively. COM- and COP-based velocities at the instant of pendulum contact ranged from −3.59 to 3.86 N s and −22.46 to 20.26 N s, respectively. These variables showed no fatigue × age interaction effects. Despite similarities in the instantaneous velocities across age ($p = 0.815$, $p = 0.155$), older adults adopted a posture shifted anteriorly an additional 4.3% of foot length for both COM- and COP-instantaneous positions ($p = 0.011$, $p = 0.010$). Of the 12 total COM- and

Table 3
Linear covariate slopes of perturbation magnitude, and instantaneous kinematics at pendulum impact for each balance recovery measure.

Dependent measure	Covariate slopes		
	Perturbation magnitude	Instantaneous position	Instantaneous velocity
Center of mass			
Peak displacement	0.0239*	0.2290*	0.0176*
Time-to-peak displacement	21.548	828.4*	21.79*
Peak velocity	1.237*	−2.123*	0.3790*
Time-to-peak velocity	−0.3378	−1.449	−0.3700
Min. time-to-boundary	−32.69*	−587.9*	−16.34*
Return 20%	50.12*	893.8*	58.56
Center of pressure			
Peak displacement	0.0209*	−0.6174*	−0.0030*
Time-to-peak displacement	19.19*	409.1*	0.6994
Peak velocity	2.302*	−0.6453*	−118.1*
Time-to-peak velocity	−0.7963	−16.71*	−0.0185
Min. time-to-boundary	−7.395*	70.13*	0.5286*
Return 20%	49.53*	−0.3847*	1254

* The covariate slope is significantly different from 0 ($p \leq 0.05$).

COP-based measures of BR, all but one (time-to-peak COM velocity) were linearly dependent on initial position at the time of perturbation (Table 3). Seven of the 12 BR measures were linearly dependent upon instantaneous velocity, all of which were also linearly dependent upon perturbation magnitude and initial position. Ten BR measures were linearly dependent upon perturbation magnitude (excluding time-to-peak COM and time-to-peak COP).

4. Discussion

The purpose of this study was to investigate the effects of LMF on BR following a postural perturbation. The maximum perturbation that could be withstood without stepping was not significantly affected by LMF, but COM excursion during BR generally increased after LMF. COP excursion exhibited more complex findings in which main effects of age, interactive effects of LMF and age, and interactive effects of LMF and muscle were identified. In general, the results indicated that LMF impaired balance recovery in both groups, changes in the older group larger for some variables, and COP-based variables are best interpreted within the context of COM-based variables.

The COM- and COP-based measures of BR were comparable to other studies employing related measures. Mean values of COM displacement and COM time-to-peak displacement were similar to those reported by Brown and Frank [12]. In addition, the COM minimum time-to-boundary reported here were similar to those reported by Schulz et al. [21]. Discrepancies between the studies may have resulted from the differing perturbation methods, or perhaps because different instructions were given to the participants. The magnitude of changes in COM- and COP-based measures with LMF averaged 0.5–8.5% of unfatigued values and effect sizes were 0.14–0.81. These were comparable to other studies reporting fatigue-induced changes in AP postural sway during quiet standing [5,22,23]. The majority of statistically significant findings had low to medium effect sizes.

COM-based measures of BR exhibited greater excursions with LMF. Both peak COM displacement and time-to-return within 20% of the peak displacement indicated that the perturbed COM not only moved closer to the base-of-support boundary, but was displaced for a longer period of time following LMF. Increases in the peak COM velocity and time-to-peak COM displacement correspond with an increase in peak angular momentum and a delay in reversing the direction of momentum, respectively. When considered together, these changes imply a greater likelihood of stepping and possibly a decrease in the ability to recover without stepping. Consistent with this interpretation, the maximum perturbation that could be withstood without stepping tended to decrease with LMF (effect size $g = 0.47$). However, the increment in perturbation magnitude used to identify this maximum perturbation may have, in retrospect, been too large to detect small effects of LMF.

Most studies employing the COP trajectory during quiet or perturbed stance have used this measure as a surrogate for the COM. COP is relatively easy to collect in the laboratory and clinic, but when used to analyze posture it is best interpreted as a controller of body kinematics used to retain the COM within the base-of-support [24,25]. The COP is influenced by passive components (intrinsic stiffness and damping and tonic muscle activity) and active components (joint torques produced by automatic feedback control and descending cognitive command) used by the postural control system to influence body kinematics. Therefore, it is appropriate to interpret the results of COP-based measures alongside the COM when available, and is treated here as the net torque (time derivative of angular momentum) used to control body kinematics.

Following LMF, a decrease in peak COP displacement occurred with a simultaneous increase in peak COM displacement and a slower return of the COM to the pre-perturbation position. The COP trajectory is a lumped measure of the net torque used to control the angular momentum of the body. As such, the observed increase in these COM-based measures is consistent with a decrease in the maximal net torque as demonstrated by the decreased peak COP displacement. One reason for a decreased torque could be a limited ability to generate ankle plantar flexor or lumbar extensor torque due to muscle fatigue. Another explanation could be a shift in postural strategy following LMF toward the “hip strategy” as demonstrated with LMF in Wilson et al. [10]. Use of a “hip strategy” results in a larger COM displacement and longer time for the body to return to a vertical position in comparison to an ankle strategy [26,27].

Time-to-peak COP velocity decreased by 4 ms with LMF, indicating that LMF resulted in a reduced response time for generating net torque following the perturbation. The average values of time-to-peak COP velocity (mean $\approx$ 144 ms) were slightly less than estimated delays in the central feedback loop of 150–200 ms [28]. Although this change may have little functional or clinical significance, the observed decrease in time-to-peak COP velocity with LMF may be the result of increased pre-perturbation muscle activity, which would provide a faster response by increasing muscle stiffness [25]. We controlled for variations in ankle torque prior to perturbations by including initial COP position as a covariate in our statistical analysis; however, increased agonist–antagonist cocontraction may have occurred following LMF as a proactive postural strategy.

Selected measures of BR exhibited a fatigue $\times$ age interaction effect. Peak COP velocity was higher in the older group and increased only in the older group following LMF. An increase in velocity is consistent with the higher peak COP displacement in the older group since further anterior displacement with no change in time-to-peak displacement necessitates a higher velocity. Perhaps most notable is the 8.5% LMF-induced change in the older group in COP time-to-return within 20% of the peak displacement, while most other changes in the COM- and COP-based measures were much smaller (0.6–4.1%). Similar to the time-to-return to 20% threshold for the COM, the increase in time-to-return to 20% threshold for the COP indicates that a longer time was needed to complete the recovery from the perturbation. However, this increase occurred only in the older group, suggesting that older adults were more challenged during recovery than the young group.

Two COP-based measures (peak velocity, and minimum time-to-boundary) exhibited interactive effects of fatigue $\times$ muscle. Specifically, peak velocity increased only with lumbar extensor fatigue and minimum time-to-boundary increased only following plantar flexor fatigue. Although difficult to assess with this data set, it may be that individuals employed differing postural control strategies following fatigue at each location. For instance, if plantar flexor fatigue resulted in a greater shift toward a hip strategy, decreasing the required ankle torque, this could appear as an increased COP time-to-boundary. Likewise, localized lumbar extensor fatigue has been shown to affect proprioceptive function in locations that are distal to the site of fatigue (e.g. the ankle [29]). Such effects may result in a balance recovery strategy that is evidenced by increased COP velocity.

Covariate analysis demonstrated that uncontrolled parameters (here initial position and velocity) influenced measured responses to the perturbation. Subsequent work using sudden perturbations may benefit by either including covariates in statistical analysis or controlling for these parameters as in Luchies et al. [30]. We attempted to implement a similar criterion to control for these variables, but were unsuccessful given the inherent time between release of the pendulums and contact with the participant. Comparison of the initial parameters between age groups revealed a more anterior posture in the older group. This may be indicative of an anticipatory balance strategy with LMF similar to that discussed in Wilson et al. [10].

Several limitations must be noted in this investigation. First, the participants were aware that they were going to be perturbed. As a result, we cannot infer how their responses would be different after an unexpected perturbation in a natural environment. However, these results of LMF- and age-related changes in postural control may indicate greater effects in more ecologically valid circumstances. Second, we did not employ a quantitative method to monitor preactivation or cocontraction of postural muscles prior to the perturbation. In addition, participants were instructed to recover balance without stepping, which may not correspond to their natural response. However, there may be analogous situations (e.g. in the workplace) when stepping to recover balance is not possible. Lastly, the increments by which the perturbation magnitudes were administered (1 N s) may have been too large to assess maximum perturbations with sufficient accuracy. This becomes of greater importance in the older population and during fatigued perturbations in both groups, as potential changes in BR may be more easily identified with higher resolution of perturbation magnitudes.

In summary, this investigation has examined the effects of LMF during a challenging postural environment and has stressed the importance of interpreting the COP trajectory (postural control) in after examining COM-based variables (postural performance). LMF increased COM excursion following a postural perturbation and elicited alterations to the COP consistent with changes in postural strategy. The results indicated that LMF impaired balance recovery in both groups, and some larger effects in the older group. Changes in these measures during submaximal perturbations imply a higher likelihood of requiring an alternate strategy (such as stepping) with slightly larger perturbations. Interaction effects with age indicated that recovering balance following LMF was more difficult among older individuals. Most effect sizes were considered small, but similar to changes in postural sway with LMF in other studies. This may indicate consistency of performance even during more challenging postural situations.

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Conflict of interest

None of the authors have ties to any activities or industry that could inappropriately influence his or her judgment regarding this research or the results presented in the manuscript.

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