Aging Clinical and Experimental Research

Neuromechanical response of the upper body to unexpected perturbations during gait initiation in young and older adults

-- Manuscript's Last Draft Submitted In This Revised Form --

Manuscript Accepted: 6th May 2020

Published online 23rd May 2020

Corresponding Author:	Lorenzo Rum, Ph.D. Foro Italico University of Rome ITALY
Corresponding Author's Institution:	Foro Italico University of Rome
First Author:	LORENZO RUM, Ph.D.
Order of Authors:	LORENZO RUM, Ph.D.
	Giuseppe Vannozzi, PhD
	Andrea Macaluso, Professor
	Luca Laudani, PhD

Abstract

Background: Control of upper body motion deteriorates with ageing leading to impaired ability to preserve balance during gait, but little is known on the contribution of the upper body to preserve balance in response to unexpected perturbations during locomotor transitions, such as gait initiation.

Aim: To investigate differences between young and older adults in the ability to modify the trunk kinematics and muscle activity following unexpected waist lateral perturbations during gait initiation.

Methods: Ten young (25±2yrs) and ten older adults (73±5yrs) initiated locomotion from stance while a lateral pull was randomly applied to the pelvis. Two force plates were used to define the feet centre-of-pressure displacement. Angular displacement of the trunk in the frontal plane was obtained through motion analysis. Surface electromyography of cervical and thoracic erector spinae muscles was recorded bilaterally.

Results: A lower trunk lateral bending towards the stance leg side in the preparatory phase of gait initiation was observed in older participants following perturbation. Right thoracic muscle activity was increased in response to the perturbation during the initial phase of gait initiation in young (+68%) but not in older participants (+7%).

Conclusions: The age-related reduction in trunk movement could indicate a more rigid behaviour of the upper body employed by older compared to young individuals in response to unexpected perturbations preceding the initiation of stepping. Older adults' delayed activation of thoracic muscles could suggest impaired reactive mechanisms that may potentially lead to fall in the early stages of the gait initiation.

Keywords Gait initiation, Postural control, Perturbation, Trunk, Ageing

Introduction

The initiation of human gait is a naturally destabilizing task which requires the transition from a stable double-limb posture to an unsteady single-limb support while controlling a forward fall [1]. The postural control system guarantees the optimal initial balance conditions by performing anticipatory postural adjustments (APAs) that prepare the first step execution by shifting the body weight towards the stance leg before stepping [2–5]. This preparatory phase begins with a "release" of the centre of pressure (CoP) moving backwards and laterally towards the swing leg that accelerates the whole-body centre of mass (CoM) forward and laterally towards the stance leg [3]; this is followed by the "unloading" of the swing leg whereby the CoP migrates to the stance leg to initiate the first step forward [6].

The upper body, and specifically the head-trunk system, has been shown to contribute loading the stance leg by leaning forward in a co-ordinated way [7, 8] and by bending laterally [9] to help maintaining balance. The upper body as a whole is a multi-articulated structure that requires a complex interaction between passive and active components of the neuromuscoloskeletal system to maintain stability during daily life movements [10]. When postural stability during standing is threatened by balance disturbances that are generated by voluntary movements of the upper limbs, muscles of the trunk and neck act in a modulated anticipatory mode, thereby contributing to preserve whole-body stability [11–13]. During voluntary gait initiation, an anticipatory top-down activation of erector spinae muscles in the swing leg side stabilises the upper body segments before step execution, thus concurring to fulfil the balance requirements of gait initiation [9]. On the other hand, in case of external, unexpected perturbations (i.e. an occasional lateral push or pull), the early postural response of the upper body is based on intrinsic stiffness of muscles and tissues, followed by appropriate long-latency responses that are scaled to perturbation characteristics [14]. Therefore, motor control of upper body demonstrates both anticipatory and reactive components that are typically involved in posture maintanance in every circumstance. Interestingly, previous research has demonstrated that young adults can actively modify their amplitude of anticipatory displacement of the CoP by increasing activation of muscles acting on the ankle joint in response to an external perturbation involving a lateral waist pull potentially leading to fall [4, 15]. In line with that, it is reasonable to argue that the upper body could play a significant role to react external perturbations and help maintaining balance during gait initiation. To the best of the authors' knowledge, however, no previous studies have investigated the upper body coordination and control in response to an unexpected perturbation delivered during the preparatory phase of gait initiation.

The stabilisation of upper body is particularly relevant to the control of whole-body balance in the elderly, as undesirable excessive trunk motion is often related to trip and slip-related falls in this population [16]. Older adults have shown an altered coordination of the upper body segments with respect to young adults during the preparatory phase of gait initiation, involving a rigid and more variable movement of the trunk-neck-head system [8]. Noteworthy, a delayed anticipatory activity of the neck muscles during voluntary gait initiation in older compared to young adults has been demonstrated in a recent study [7], which would suggest an impairment in the active role of the upper body during the preparatory phase of gait initiation, potentially increasing the instability of the whole body. Indeed, prompt changes in trunk kinematics were shown to play a key role in fall avoidance by older individuals when balance was disrupted by a large disturbance requiring a recovery stepping response [17, 18]. It is not known, however, whether active responses of the trunk are preserved or rather impaired in older individuals initiating gait under perturbed conditions. The understanding of how older adults cope with external perturbation at trunk level during the early stages of gait initiation could be particularly of interest for designing training protocols aimed at preventing balance loss and risk of falling.

The purpose of the study was, therefore, to investigate the differences between young and older adults in the ability to modify the trunk motion and muscle activity following unexpected waist lateral pull during the preparatory phase of gait initiation. It was hypothesized that older adults would show a rigid movement and delayed muscle response of the trunk segment thus increasing the unbalancing effect of such mechanical perturbation, compared to their young counterpart.

Methods

Participants

Ten young (age 25 ± 2 years; body mass 60.2 ± 7.4 kg; height 1.65 ± 0.08 m; 8 females, 2 males) and 10 gender-matched community-dwelling older adults (age 73 ± 5 years; body mass 68.3 ± 12.5 kg; height 1.63 ± 0.11 m; 8 females, 2 males) took part in the study. Participants were recruited from the metropolitan area through local advertisement and were selected only if they were defined as "medically stable" according to Greig et al. [19]; exclusion criteria were any neurological, neuromuscular and orthopaedic impairment affecting gait ability or balance. Older participants were classified as at low risk of falling and with moderate concern about falling through the administration of the Berg Balance Scale (score 53±3) and Fall Efficacy Scale-International (score 23±7), respectively [20, 21]. The study was approved by Cardiff Metropolitan University Institutional Review Board and all participants signed an informed consent form.

Equipment and experimental procedure

Two force plates (Kistler 9287, Kistler, Switzerland) positioned at ground level were used to collect ground reaction forces (GRFs) and CoP position under each foot. A 13-cameras motion analysis system (MX System, Vicon Motion Systems, Los Angeles, CA) recorded the 3D position of 34 markers placed on body landmarks at a sampling frequency of 250 Hz. Whole-body kinematics was provided by a 12-segment biomechanical model [22]. The trunk segment was defined by four markers respectively placed on the notch between clavicles, the xiphoid process of sternum, the seventh cervical vertebra and the tenth thoracic vertebra. Surface electromyography (EMG) of cervical and thoracic erector spinae muscles of both right and left sides of the body was collected by means of a wireless EMG device (Trigno Wireless System, Delsys, Boston, MA) with parallel bar electrodes (1 cm width, 1 cm fixed inter-electrode distance). The electrodes were placed and fixed with medical tape 1.5 cm laterally from the spine on the fourth cervical vertebra (CES) and approximately 3 cm out from the vertebral ridge of tenth thoracic vertebra (TES). Before electrode placement, the skin was gently shaved and abraded to increase impendence. EMG and force plates data were sampled at 1000 Hz and time-synchronized with both motion analysis and perturbation devices.

Gait was initiated from a quiet standing position, taking at least four steps forward at self-selected comfortable speed. Before stepping, participants were required to stand still with their feet normally apart and their arms hanging off the side while fixing the gaze on a visual target located three meters away at eye level. The feet location and orientation on the floor were marked with tape to guarantee identical starting position among trials and equal weight bearing between lower limbs was guided by the experimenter cues by double-checking ground reaction force distribution between limbs. Participants initiated the task with their right leg, which was determined to be the dominant side of all subjects by asking which leg they use to kick a ball. The balance perturbation was delivered through a custom-made waist-pulling device that was aligned in the frontal plane to guarantee a discrete waist pulling in the mediolateral (ML) direction and a smooth load motion (Fig. 1). A load of 10% of the subject body mass was connected to a belt attached to the participant waist through an inextensible rope passing by a pulley, and was anchored to a support by an electromagnetic brake; when the vertical component of GRF under the stepping foot increased by 10% of the participant's body weight, the load was released to deliver the balance perturbation (Fig.1). Another electromagnet that connected the load to the rope was turned off to detach the load and end the perturbation when the vertical force under the stepping foot dropped to zero, i.e. at the toe-off of the stepping foot (TO). The perturbation was delivered with a mechanical delay of ~100 ms, thus falling before the ML CoP peak that normally occurs after about 300 ms from the whole movement onset, and pulled the subject waist towards the stepping side opposite to the CoM displacement towards the stance foot that typically occur during this phase.

FIGURE 1 HERE

Participants were required to perform ten consecutive unperturbed gait initiation trials (UP condition), followed by a block of 60 trials with a 20% rate of balance perturbation occurrence, thus including twelve randomly perturbed trials (P condition). During trials without perturbation in this block, the signal from the force plate was used to directly detach the load from the subject, allowing a smooth initiation of gait with no movement constraints. Participants were unaware of when (i.e. in which trial) the perturbation would have occurred, and they were asked to initiate gait as normally as possible and at self-pace. However, they knew that perturbation could have occurred and, hence, a certain degree of expectancy was unavoidable. Ten trials without perturbation were therefore randomly selected for analysis (EXP condition). Five-minute breaks were provided at the end of UP condition and after 30 trials of the random perturbation condition. During the entire block of 60 trials, a safety harness was fitted to the trunk of the subject and attached through a cable to a ceiling-mounted trolley to prevent fall occurrence while not constraining the movement.

Data analysis

Raw kinematic and kinetic data were filtered by a low-pass 4th order Butterworth filter with a 15 Hz cut-off frequency. Net CoP position was calculated from the two force plates by averaging individual plates coordinates of each foot CoP [23] and the relevant “CoP onset” instant was defined as a ML CoP displacement of 3 standard deviations above a mean value calculated over a 500 ms interval manually selected in the baseline period [24, 25]. TO instant was calculated as when the vertical force registered under the stepping foot dropped to zero. The maximum lateral displacement of CoP (ML CoP peak) was identified to define the “release” (i.e. from CoP onset until ML peak) and the “unloading” phases (i.e. from ML CoP peak to TO) that precede the first step execution. To investigate changes in whole-body kinetics following the perturbation, ML GRF peak after CoP onset was identified and normalised by subject’s body weight; then, it was used to compute the rate of force development (RFD) from CoP onset to the peak. To evaluate the changes in upper body motion in response to the perturbation before stepping, absolute angular displacement relative to the laboratory reference frame of the upper trunk segment (subsequently referred to as trunk) was obtained by the biomechanical model adopted by Gutierrez et al. [22] (Fig. 1). The trunk angular displacement from CoP onset to TO was then calculated in the frontal plane.

EMG signals were band-pass filtered (20-450 Hz) with a 4rd order Butterworth filter and then high-pass filtered with a 30 Hz 2nd order Butterworth filter to remove ECG artefacts [26]. The filtered signal was then full-wave rectified and normalized by the mean value computed over a 1000 ms window selected 1000 ms prior to CoP onset in each trial [27]. Amplitude of muscle activity was quantified by calculating the root mean square (RMS) of the normalised EMG signal during the release and unloading phases. The integral of EMG signal over the two phases was also computed and a coactivation index (CI) was used to compare the level of activity between the two sides (right and left) of both CES and TES as follows [28]:

$$CI = \frac{\int EMG_{right}}{\int EMG_{right} + \int EMG_{left}} \times 100$$

where index value of 50% corresponds to an equal contribution of both sides to total activation (maximal coactivation and increase in spine stiffness), while values greater than 50% correspond to a higher contribution of the right side compared to the left side.

Statistical analysis

The effect of adaptation to trials sequence during the P condition on kinetic, kinematic and EMG parameters was evaluated by a mixed ANOVA with the trial sequence as within-subject factor and group as between-subject factor. The results from the analysis showed no significant effect of trial sequence on any parameter for both groups. Consequently, individual mean values were calculated from at least ten trials per each condition for further analysis. A mixed multivariate ANOVA approach with repeated measures (RM MANOVA) was then used to evaluate the effects of perturbation between conditions (within-subject factor: UP, EXP and P) across the two age groups (between-subject factor: young and older) for grouped parameters, such as spatio-temporal parameters of CoP (e.g. ML CoP peak, release and unloading phase duration), GRF and muscle activity variables (e.g. RMS of CES and TES). Univariate approach (RM ANOVA) was used to evaluate frontal trunk angular displacement variable. The assumption of normal distribution was checked using the Shapiro-Wilk test. Log-transformation of no normal data was performed to meet normality assumption for statistical analysis. Sphericity assumption was assessed by Mauchly's test and Greenhouse-Geisser correction of degrees of freedom was used in case of violation. When significant interaction or main effects were found in the multivariate analyses, separate ANOVAs were performed as follow-up testing. In case of significant effect, pairwise comparison analysis between conditions was performed in each group separately to evaluate the effects of within group changes in adaptation to perturbation. The relationship between thoracic muscles response to perturbation and kinematic displacement of the upper body was investigated by a partial correlation analysis with repeated measures within subjects [29] for all P trials between the EMG parameters (RMS and CI of TES) in the unloading phase as independent variables, and the trunk frontal angular displacement as dependent variable. Statistical analyses were performed using SPSS 23.0 software (Chicago, IL, USA) and significance α level was set at 0.05 with Holm-Bonferroni correction. Partial eta squared (η^2_p) was reported as measure of effect size with 0.0099, 0.0588 and 0.1379 corresponding to the benchmarks for small, medium and large effect size, respectively [30].

Results

Averaged kinetic, kinematic and EMG data from UP and P conditions of one representative subject per each group are presented in Figure 2. When young subjects were asked to react to the perturbation, a postural response in terms of increments in CoP displacement, ML GRF, trunk frontal kinematics and right EMG activity was present. On the other hand, older participants showed an altered postural response that was characterized by lower changes in the same parameters.

FIGURE 2 HERE

Table 1 shows group data of the spatiotemporal parameters derived from CoP displacement (ML peak, release and unloading phase durations) and ML GRF. RM MANOVA on spatiotemporal parameters of CoP displacement showed significant interaction (Wilk's $\Lambda = 0.631$, $F(6,68) = 2.938$, $p < 0.05$, $\eta^2_p = 0.206$) and condition effects (Wilk's $\Lambda = 0.203$, $F(6,68) = 13.797$, $p < 0.001$, $\eta^2_p = 0.549$). Separate ANOVAs revealed that the interaction effect was significant in ML CoP peak and release phase duration only ($F(1.512,27.210) = 8.760$, $p < 0.01$, $\eta^2_p = 0.327$; and $F(1.250,22.492) = 6.779$, $p < 0.05$, $\eta^2_p = 0.274$, respectively). Subsequent pairwise comparisons showed that ML CoP peak was significantly different between the three conditions in young, with an increasing trend from UP to EXP to P condition ($p < 0.01$), whereas older participants showed a significant increase in EXP and P compared to UP ($p < 0.001$), with no difference between EXP and P conditions. Release phase duration was greater in P than both UP and EXP in young ($p < 0.01$) but not in older participants. A significant effect of condition was found in the unloading phase duration ($F(2,36) = 15.767$, $p < 0.001$, $\eta^2_p = 0.467$) as it decreased from UP and EXP to P condition in both groups ($p < 0.05$).

RM MANOVA on ML GRF parameters showed significant interaction (Wilk's $\Lambda = 0.646$, $F(4,70) = 4.273$, $p < 0.01$, $\eta^2_p = 0.196$) and condition effects (Wilk's $\Lambda = 0.326$, $F(4,70) = 13.139$, $p < 0.001$, $\eta^2_p = 0.429$) (Table 1). Separate ANOVAs indicated that the interaction effect was significant in ML

GRF peak ($F(1.291,23.229) = 8.644$, $p < 0.01$, $\eta^2_p = 0.324$), with pairwise comparisons revealing a significant increasing trend from UP to EXP to P in young participants ($p < 0.01$), whereas older participants showed greater ML GRF peak in EXP and P compared to UP condition ($p < 0.05$). A significant condition effect was found in RFD of ML GRF from CoP onset to the peak ($F(2,36) = 22.980$, $p < 0.001$, $\eta^2_p = 0.561$). However, pairwise comparison analysis indicated that young participants had greater RFD in P than UP and EXP condition ($p < 0.01$), whereas older participants showed a significant increase from UP to P only ($p < 0.05$).

TABLE 1 HERE

Figure 3 shows trunk frontal angular displacement in the preparatory phase of gait initiation, i.e. from CoP onset to TO. RM ANOVA showed a significant interaction effect ($F(1.456,26.206) = 14.602$, $p < 0.001$, $\eta^2_p = 0.448$) and pairwise comparisons revealed that a greater displacement of the trunk segment towards the direction opposite to the perturbation occurred in P compared to UP and EXP trials in young ($p < 0.001$), whereas it increased from UP to EXP to P in older participants ($p < 0.05$).

FIGURE 3 HERE

Figure 4 and 5 show the EMG RMS of bilateral CES and TES in the UP, EXP and P conditions during the release and unloading phases, respectively. Multivariate analysis showed a significant interaction between group and condition (Wilk's $\Lambda = 0.313$, $F(16,58) = 2.852$, $p < 0.01$, $\eta^2_p = 0.440$), with separate ANOVAs revealing significant differences in right CES muscle activity during the release phase ($F(1.310,23.577) = 5.486$, $p < 0.05$, $\eta^2_p = 0.234$), right TES muscle activity during the release and unloading phases ($F(1.211,21.793) = 8.083$, $p < 0.01$, $\eta^2_p = 0.310$; and $F(2,36) = 11.677$, $p < 0.001$, $\eta^2_p = 0.393$, respectively) and left TES muscle activity during the unloading phase ($F(2,36) =$

6.831, $p < 0.01$, $\eta^2_p = 0.275$). Pairwise comparisons indicated that a significant increase in right CES and TES muscle activity from UP and EXP to P condition during the release phase was observed in young ($p < 0.01$) but not older participants (Fig.4). During the unloading phase, a bilateral increase in TES muscle activity was observed in young from UP and EXP to P condition ($p < 0.001$), whereas older participants significantly increased muscle activation at the right side only from UP to P and to a lesser extent ($p < 0.05$) (Fig.5). In addition, RM MANOVA showed a significant condition effect (Wilk's $\Lambda = 0.111$, $F(16,58) = 7.232$, $p < 0.001$, $\eta^2_p = 0.666$), with separate ANOVAs reporting significant increases in the muscle activity of left TES during the release phase ($F(2,36) = 15.886$, $p < 0.001$, $\eta^2_p = 0.469$) (Fig.4) and bilateral muscle activity of CES during the unloading phase (right: $F(1.042,18.764) = 22.493$, $p < 0.001$, $\eta^2_p = 0.555$; left: $F(1.541,27.736) = 21.536$, $p < 0.001$, $\eta^2_p = 0.545$) (Fig.5) from UP and EXP to P condition in both groups ($p < 0.01$).

FIGURE 4 AND 5 HERE

Although multivariate analysis only showed a significant effect of condition on CIs (Wilk's $\Lambda = 0.361$, $F(8,66) = 5.484$, $p < 0.001$, $\eta^2_p = 0.399$), separate ANOVAs reported a significant interaction effect on the CI of TES during the unloading phase ($F(2,36) = 4.879$, $p < 0.05$, $\eta^2_p = 0.213$). Specifically, the CI was higher in P compared to UP condition in young ($p < 0.001$) to a greater extent than in older participants ($p < 0.01$) (Fig. 6). In addition, young participants showed greater CI of TES during the unloading phase in P than EXP condition ($p < 0.001$). Separate ANOVAs confirmed the results of multivariate analysis by reporting a significant effect of condition on CI during release phase at TES ($F(2,36) = 5.311$, $p < 0.05$, $\eta^2_p = 0.228$) and during the unloading phase at CES ($F(1.485,26.723) = 8.416$, $p < 0.01$, $\eta^2_p = 0.319$). Although no significant interaction effect was found, pairwise comparisons of CI TES during release phase revealed a greater contribution of right TES muscle activity in P compared to UP and EXP in young ($p < 0.05$), whereas no significant changes were observed in older participants. Conversely, a greater contribution of right CES muscle activity during the unloading phase from UP to P was found in older participants ($p < 0.05$) but not in young.

FIGURE 6 HERE

Partial correlation analysis showed that frontal trunk angular displacement was positively correlated with the right TES muscle response to a greater extent in older compared to young (r = .47, $p < 0.001$, and r = .25, $p < 0.05$, respectively). In addition, older participants showed a positive correlation between frontal trunk angular displacement and left TES muscle activity (r = .27, $p < 0.05$), whereas young participants showed a correlation between frontal trunk angular displacement and CI of TES (r = .26, $p < 0.05$).

Discussion

In the present study, for the first time to the authors' knowledge, unexpected waist-pulling lateral perturbations have been delivered in young and older adults during the preparatory of phase of gait initiation, in order to investigate the age-related differences in the upper body contribution to resisting such perturbation and preserving balance.

Kinematic analysis of the upper body motion demonstrated lower trunk lateral bending towards the stance leg before stepping in older participants compared to young participants, suggesting a decreased ability to modify the upper body movement and overcome the perturbation during the preparatory phase of gait initiation in aged individuals. During the preparatory phase of gait initiation, the CoM has to be displaced toward the stance side in an anticipatory manner before the stepping foot lift off, in order to prepare and ensure a smooth transition between bipedal standing and the stance phase of the first step forward [3]. From a biomechanical point of view, the reduced lateral leaning toward the stance side reported in older individuals does not facilitate a smooth migration of the CoM towards the stance limb and can potentially destabilise older individuals in the transition between standing and walking. In addition, such age-related differences would indicate that the older participants' upper body behaved more rigidly in response to the perturbation compared to young participants. This is in agreement with previous studies showing that healthy older people may assume a rigid upper body posture during walking at natural speed [31] and particularly during transitory locomotor tasks such as gait initiation [8] and termination [32]. During both standing and stepping, the CNS executes APAs to ensure postural stability by controlling the body CoM motion over the base of support, and it is especially true while maintaining lateral stability in a fall prevention-related context [33–36]. The APA that precedes stepping in gait initiation maintains the lateral stability by accelerating the CoM towards the stance leg and this mechanism is known to be altered in older individuals [6, 36–39]. When stepping is unexpectedly induced by a forward waist-pulling perturbation, for instance, elderly anticipated step execution without any change in APA duration, whereas young adults spent more time adjusting their posture before the foot lift-off [40]. The present study confirmed these previous results as older adults were unable to modify release phase duration and amplitude (ML CoP peak) as well as GRF in response to an unexpected ML perturbation as young did. Interestingly, the reduced global postural response in older participants was accompanied by a diminished upper body motion. This would suggest that the adaptation of APA amplitude and timing in response to perturbation is linked to the upper body posture, with greater increase in APA amplitude and timing allowing for better postural reaction at the upper body level.

Both young and older participants in the study showed a clear increase in left thoracic muscle activity during the release phase of gait initiation under perturbed conditions. Conversely, young participants showed an increase in right cervical and thoracic muscle activity with a greater right-side contribution at trunk level in the release phase under perturbed condition, whereas older individuals showed a delayed activation of these muscles in the unloading phase only. From a biomechanical perspective, as the ML perturbation induced a shift of the pelvis towards the swing leg (right) side, producing an opposite momentum causing the upper body to move towards the stance leg (left) side, the involvement of right upper body musculature could be necessary to prevent the trunk and head to collapse towards the opposite side due to inertia. The activation of paraspinal muscles during various motor tasks is partly finalised to the optimal positioning of head in space that guarantees a stable frame of reference for planning and executing motor actions [41]. Stabilisation of the head is particularly provided by both anticipatory and reflex control mechanisms in cervical joint muscles that are known to be impaired in elderly, thus altering the ability to compensate for random disturbances [42]. In this study, the delay in cervical EMG activation pattern while responding to the perturbation confirms previous findings on deterioration of the ability to control head muscles in response to unexpected perturbations with ageing. Furthermore, the delayed and more symmetrical thoracic muscle activity in older participants compared to young participants are in line with previous work reporting an impairment of the trunk neuromechanical control when dealing with balance disturbances [16]. Overall, this could suggest a deterioration in neuromuscular control of upper body postural stability responses in older individuals experiencing unexpected balance disturbances while performing locomotor actions that likely requires alternative recovery strategies to avoid falling. Noteworthy, no acute adaptation to perturbation was observed across perturbed trials in both young and older participants. Perturbation-based balance training has been shown to be an effective approach to fall risk reduction in elderly [43], with previous training protocols being mainly focused on upright steady posture and gait. However, it is still unknown whether the practice over time of the present perturbation protocol could similarly improve balance capacity in older adults by restoring an appropriate upper body motor control. Therefore, future research should also explore training protocols that involve gait transitions, i.e. gait initiation, under perturbed conditions as complex interaction of anticipatory and reactive components of postural control can be observed during such circumstances.

The correlation analysis between thoracic muscle response to perturbation and trunk angular displacement endorses the hypothesis of an altered trunk neuromechanical control in older adults. Particularly, the association between frontal trunk motion and bilateral thoracic muscle activation suggests spine stiffening as upper body postural response in aged individuals. On the other hand, increase in unilateral (right) thoracic muscle activation was correlated to major frontal trunk motion in young adults, likely being more adequate to the upper body postural changes elicited by the perturbation mechanical effect. The observation of these two peculiar age-related motor strategies enhances the knowledge of previously reported reduction in lateral stability control in elderly people, highlighting the contribution of upper body in balance maintenance [33, 44–47].

It has been already reported that elderly people show an impaired APA control in response to an unexpected perturbation [40], nonetheless they maintain the ability to adjust the ongoing stepping process in response to a visual perturbation [48]. In this study, the unvaried APA duration altogether with the reduction in unloading phase duration would indicate an earlier step initiation in older than young adults. In previous research, older individuals have been reported to pre-select a stepping recovery strategy within context of perturbation uncertainty [40] and it has been suggested that the aging process per se (i.e. previous experiences, fear of falling, anxiety) rather than sensorimotor dysfunction may be related to this pre-programmed response [49]. In the present study, analysis of a subset of unperturbed trials within the perturbed condition (EXP condition) showed that both young and older participants modified their APAs (ML CoP and GRF peaks) solely based on the expectation of perturbation occurrence, with young only presenting further differences compared to the actual perturbed condition. A minimal adjustment of trunk frontal motion about (0.5°) was also observed, although it could potentially be attributed to measurement error and/or the accuracy level of motion analysis system [50]. On the other hand, no modification in EMG activity of trunk and neck muscles were evident in the unperturbed trials with perturbation expectancy. This is in line with previous studies showing that fear of falling can alter postural control parameters during quiet stance and gait [51]. Despite similar levels of expectation, results from the perturbed condition showed that young and older individuals performed different postural strategies while responding to the perturbation. Therefore, it might be argued that the age-related alteration in the APA adaptation to perturbation and in the neuromuscular response at trunk level is indicative of a diminished ability to adapt the gait initiation motor program once it has been released by older people.

A limitation of the present study is the small sample size. Although the effect size of all observed significant results was large, a greater number of participants would have reduced the variability of data and increased the likelihood of identifying more subtle age-related differences. Another limitation is the unbalanced number of male and female participants within each age group, which may have hidden gender-related differences and, therefore, should be considered when interpreting the present results.

In conclusion, older participants showed impaired kinematic patterns of the trunk in response to the perturbation, with lower trunk lateral bending towards the stance leg compared to young participants.

In addition, older participants showed asymmetrical and delayed activation of thoracic muscles compared to young participants, suggesting impaired reactive mechanisms that could potentially lead to fall in the early stages of the gait initiation. Given the alteration of the postural response in older adults, it would be relevant to investigate whether the ability to adapt APA while adjusting trunk posture with respect to external perturbation characteristics could be modified through the application of stepping training protocol that involves unexpected ML perturbation during the preparatory phase of gait initiation.

Declarations

Author contributions LR and LL conceived and designed the research. LR conducted the experiments under LL's supervision. LR, GV and LL analysed data and all authors interpreted the experimental results. LR and LL wrote the manuscript. All authors edited, revised and approved the final version of the manuscript.

Compliance with ethical standards

Conflict of interest The authors declare no conflict of interest.

Ethical approval All procedures performed in studies involving human participants were in accordance with the ethical standards of the institutional research committee (Cardiff Metropolitan University ethics committee – 17/4/02R) and with the 1964 Helsinki declaration and its later amendments or comparable ethical standards.

Informed consent Informed consent was obtained from all individual participants included in the study.

References

1. Lepers R, Brenière Y (1995) The role of anticipatory postural adjustments and gravity in gait initiation. Exp Brain Res 107:118–124. https://doi.org/10.1007/BF00228023
2. Crenna P, Frigo C (1991) A motor programme for the initiation of forward-oriented movements in humans. J Physiol 437:635–53. https://doi.org/10.1113/jphysiol.1991.sp018616
3. Jian Y, Winter D, Ishac M, Gilchrist L (1993) Trajectory of the body COG and COP during initiation and termination of gait. Gait Posture 1:9–22. https://doi.org/10.1016/0966-6362(93)90038-3
4. Mille M-L, Simoneau M, Rogers MW (2014) Postural dependence of human locomotion during gait initiation. J Neurophysiol 112:3095–3103. https://doi.org/10.1152/jn.00436.2014
5. Mizusawa H, Jono Y, Iwata Y, et al (2017) Processes of anticipatory postural adjustment and step movement of gait initiation. Hum Mov Sci 52:1–16. https://doi.org/10.1016/j.humov.2017.01.003
6. Mickelborough J, Van Der Linden ML, Tallis RC, Ennos AR (2004) Muscle activity during gait initiation in normal elderly people. Gait Posture 19:50–57. https://doi.org/10.1016/S0966-6362(03)00016-X
7. Maslivec A, Bampouras TM, Dewhurst S, et al (2018) Mechanisms of head stability during gait initiation in young and older women: A neuro-mechanical analysis. J Electromyogr Kinesiol 38:103–110. https://doi.org/10.1016/j.jelekin.2017.11.010
8. Laudani L, Casabona A, Perciavalle V, Macaluso A (2006) Control of head stability during gait initiation in young and older women. J Electromyogr Kinesiol 16:603–610. https://doi.org/10.1016/j.jelekin.2006.08.001
9. Ceccato J-C, de Sèze M, Azevedo C, Cazalets J-R (2009) Comparison of trunk activity during gait initiation and walking in humans. PLoS One 4:e8193.

https://doi.org/10.1371/journal.pone.0008193

10. Summa A, Vannozzi G, Bergamini E, et al (2016) Multilevel upper body movement control during gait in children with cerebral palsy. PLoS One 11:1–13. https://doi.org/10.1371/journal.pone.0151792

11. Aruin AS, Latash ML (1995) The role of motor action in anticipatory postural adjustments studied with self-induced and externally triggered perturbations. Exp brain Res 106:291–300. https://doi.org/10.1007/bf00241125

12. Falla D, Rainoldi A, Merletti R, Jull G (2004) Spatio-temporal evaluation of neck muscle activation during postural perturbations in healthy subjects. J Electromyogr Kinesiol 14:463–474. https://doi.org/10.1016/j.jelekin.2004.03.003

13. Danna-Dos-Santos A, Degani AM, Latash ML (2007) Anticipatory control of head posture. Clin Neurophysiol 118:1802–1814. https://doi.org/10.1016/j.clinph.2007.05.060

14. Rietdyk S, Patla AE, Winter DA, et al (1999) Balance recovery from medio-lateral perturbations of the upper body during standing. J Biomech 32:1149–1158. https://doi.org/10.1016/S0021-9290(99)00116-5

15. Mouchnino L, Robert G, Ruget H, et al (2012) Online control of anticipated postural adjustments in step initiation: Evidence from behavioral and computational approaches. Gait Posture 35:616–620. https://doi.org/10.1016/j.gaitpost.2011.12.009

16. Grabiner MD, Donovan S, Bareither M Lou, et al (2008) Trunk kinematics and fall risk of older adults: Translating biomechanical results to the clinic. J Electromyogr Kinesiol 18:197–204. https://doi.org/10.1016/j.jelekin.2007.06.009

17. Pavol MJ, Owings TM, Foley KT, Grabiner MD (2001) Mechanisms leading to a fall from an induced trip in healthy older adults. J Gerontol A Biol Sci Med Sci 56:M428-37. https://doi.org/10.1093/gerona/56.7.m428

18. Owings TM, Pavol MJ, Grabiner MD (2001) Mechanisms of failed recovery following

postural perturbations on a motorized treadmill mimic those associated with an actual forward trip. Clin Biomech (Bristol, Avon) 16:813–9. https://doi.org/10.1097/00005537-200003002-00002

19. Greig CA, Young A, Skelton DA, et al (1994) Exercise studies with elderly volunteers. Age Ageing 23:185–9. https://doi.org/10.1093/ageing/23.3.185

20. Berg KO, Wood-Dauphinee SL, Williams JI, Maki B (1992) Measuring balance in the elderly: validation of an instrument. Can J Public Health 83 Suppl 2:S7-11. https://doi.org/10.1111/j.1532-950X.2008.00478.x

21. Yardley L, Beyer N, Hauer K, et al (2005) Development and initial validation of the Falls Efficacy Scale-International (FES-I). Age Ageing 34:614–619. https://doi.org/10.1093/ageing/afi196

22. Gutierrez EM, Bartonek Å, Haglund-Åkerlind Y, Saraste H (2003) Centre of mass motion during gait in persons with myelomeningocele. Gait Posture 18:37–46. https://doi.org/10.1016/S0966-6362(02)00192-3

23. Winter DA, Prince F, Frank JS, et al (1996) Unified theory regarding A/P and M/L balance in quiet stance. J Neurophysiol 75:2334–43. https://doi.org/10.1152/jn.1996.75.6.2334

24. MacKinnon CD, Bissig D, Chiusano J, et al (2007) Preparation of Anticipatory Postural Adjustments Prior to Stepping. J Neurophysiol 97:4368–4379. https://doi.org/10.1152/jn.01136.2006

25. Uemura K, Yamada M, Nagai K, et al (2012) Fear of falling is associated with prolonged anticipatory postural adjustment during gait initiation under dual-task conditions in older adults. Gait Posture 35:282–286. https://doi.org/10.1016/j.gaitpost.2011.09.100

26. Drake JDM, Callaghan JP (2006) Elimination of electrocardiogram contamination from electromyogram signals: An evaluation of currently used removal techniques. J Electromyogr Kinesiol 16:175–187. https://doi.org/10.1016/j.jelekin.2005.07.003

27. Timmann D, Horak FB (2001) Perturbed step initiation in cerebellar subjects: 2. Modification of anticipatory postural adjustments. Exp Brain Res 141:110–120. https://doi.org/10.1007/s002210100858

28. Latash ML (2018) Muscle coactivation: definitions, mechanisms, and functions. J Neurophysiol 120:88–104. https://doi.org/10.1152/jn.00084.2018

29. Bland JM, Altman DG (1995) Calculating correlation coefficients with repeated observations: Part 1--Correlation within subjects. BMJ 310:446. https://doi.org/10.1136/bmj.310.6977.446

30. Richardson JTE (2011) Eta squared and partial eta squared as measures of effect size in educational research. Educ Res Rev 6:135–147. https://doi.org/10.1016/j.edurev.2010.12.001

31. Kavanagh JJ, Barrett RS, Morrison S (2004) Upper body accelerations during walking in healthy young and elderly men. Gait Posture 20:291–298. https://doi.org/10.1016/j.gaitpost.2003.10.004

32. Rum L, Laudani L, Macaluso A, Vannozzi G (2017) Upper body accelerations during planned gait termination in young and older women. J Biomech 65:138–144. https://doi.org/10.1016/j.jbiomech.2017.10.019

33. Rogers MW, Mille M-L (2003) Lateral stability and falls in older people. Exerc Sport Sci Rev 31:182–187. https://doi.org/10.1097/00003677-200310000-00005

34. Kanekar N, Aruin AS (2014) The effect of aging on anticipatory postural control. Exp Brain Res 232:1127–1136. https://doi.org/10.1007/s00221-014-3822-3

35. Fujimoto M, Bair WN, Rogers MW (2015) Center of pressure control for balance maintenance during lateral waist-pull perturbations in older adults. J Biomech 48:963–968. https://doi.org/10.1016/j.jbiomech.2015.02.012

36. Hyodo M, Saito M, Ushiba J, et al (2012) Anticipatory postural adjustments contribute to age-related changes in compensatory steps associated with unilateral perturbations. Gait

Posture 36:625–630. https://doi.org/10.1016/j.gaitpost.2012.06.018

37. Khanmohammadi R, Talebian S, Hadian MR, et al (2016) Characteristic muscle activity patterns during gait initiation in the healthy younger and older adults. Gait Posture 43:148–153. https://doi.org/10.1016/j.gaitpost.2015.09.014

38. Polcyn AF, Lipsitz LA, Kerrigan DC, Collins JJ (1998) Age-related changes in the initiation of gait: Degradation of central mechanisms for momentum generation. Arch Phys Med Rehabil 79:1582–1589. https://doi.org/10.1016/S0003-9993(98)90425-7

39. Wang Y, Watanabe K, Asaka T (2017) Aging effect on muscle synergies in stepping forth during a forward perturbation. Eur J Appl Physiol 117:201–211. https://doi.org/10.1007/s00421-016-3514-8

40. Rogers MW, Hedman LD, Johnson ME, et al (2003) Triggering of protective stepping for the control of human balance: Age and contextual dependence. Cogn Brain Res 16:192–198. https://doi.org/10.1016/S0926-6410(02)00273-2

41. Pozzo T, Berthoz A, Lefort L (1990) Head stabilization during various locomotor tasks in humans. I. Normal subjects. Exp brain Res 82:97–106. https://doi.org/10.1007/bf00230842

42. Keshner EA, Chen KJ (1996) Mechanisms Controlling Head Stabilization in the Elderly During Random Rotations in the Vertical Plane. J Mot Behav 28:324–336. https://doi.org/10.1080/00222895.1996.10544602

43. Gerards MHG, McCrum C, Mansfield A, Meijer K (2017) Perturbation-based balance training for falls reduction among older adults: Current evidence and implications for clinical practice. Geriatr Gerontol Int 17:2294–2303. https://doi.org/10.1111/ggi.13082

44. Hilliard MJ, Martinez KM, Janssen I, et al (2008) Lateral Balance Factors Predict Future Falls in Community-Living Older Adults. Arch Phys Med Rehabil 89:1708–1713. https://doi.org/10.1016/j.apmr.2008.01.023

45. Claudino R, dos Santos ECC, Santos MJ (2013) Compensatory but not anticipatory

adjustments are altered in older adults during lateral postural perturbations. Clin Neurophysiol 124:1628–1637. https://doi.org/10.1016/j.clinph.2013.02.111

46. Arvin M, Hoozemans MJM, Pijnappels M, et al (2018) Where to step? Contributions of stance leg muscle spindle afference to planning of mediolateral foot placement for balance control in young and old adults. Front Physiol 9:1–10. https://doi.org/10.3389/fphys.2018.01134

47. Shulman D, Spencer A, Vallis LA (2018) Age-related alterations in reactive stepping following unexpected mediolateral perturbations during gait initiation. Gait Posture 64:130–134. https://doi.org/10.1016/j.gaitpost.2018.05.035

48. Sun R, Cui C, Shea JB (2017) Aging effect on step adjustments and stability control in visually perturbed gait initiation. Gait Posture 58:268–273. https://doi.org/10.1016/j.gaitpost.2017.08.013

49. Mille M-L, Rogers MW, Martinez K, et al (2003) Thresholds for inducing protective stepping responses to external perturbations of human standing. J Neurophysiol 90:666–74. https://doi.org/10.1152/jn.00974.2002

50. Richards JG (1999) The measurement of human motion: A comparison of commercially available systems. Hum Mov Sci 18:589–602. https://doi.org/10.1016/S0167-9457(99)00023-8

51. Young WR, Mark Williams A (2015) How fear of falling can increase fall-risk in older adults: Applying psychological theory to practical observations. Gait Posture 41:7–12. https://doi.org/10.1016/j.gaitpost.2014.09.006

FIGURE CAPTIONS

Fig. 1. Back view of the experimental set-up including subject with the marker set for motion analysis system, three force plates and the waist-pulling device. In the top right panel, the four markers that define the trunk segment for kinematic calculation are displayed in the frontal plane.

Fig. 2. Averaged data from one young and one older representative subject during UP (grey) and P gait initiation (red). All time-series data were synchronized with the CoP onset (vertical dotted line) and the shaded areas represent ±1 SD.

Fig. 3. Frontal trunk angular displacement (mean ± SD) throughout the preparatory phase of gait initiation, i.e. from APA onset to TO, in both groups under unperturbed (UP), expected (EXP) and perturbed (P) conditions. Positive values correspond to lateral bending/tilt towards the direction opposite to the pulling perturbation (Pull). * = Significantly different compared to UP ($p < 0.05$). †= Significantly different compared to EXP ($p < 0.05$).

Fig. 4. RMS (mean ± SD) of EMG activity of left and right CES and TES as a percentage of mean EMG activity before CoP onset during the release phase of gait initiation under unperturbed (UP), expected (EXP) and perturbed (P) conditions. * = Significantly different compared to UP ($p < 0.05$). †= Significantly different compared to EXP ($p < 0.05$).

Fig. 5. RMS (mean ± SD) of EMG activity of left and right CES and TES as a percentage of mean EMG activity before CoP onset during the unloading phase of gait initiation under unperturbed (UP), expected (EXP) and perturbed (P) conditions. * = Significantly different compared to UP ($p < 0.05$). †= Significantly different compared to EXP ($p < 0.05$).

Fig. 6. Coactivation indexes (mean ± SD) of CES (upper panel) and TES (lower panel) during the release (left side) and unloading phase (right side) of unperturbed (UP), expected (EXP) and perturbed (P) conditions in young and older participants. Maximal coactivation corresponds to 50% (dotted horizontal line), while values greater than 50% correspond to a greater activity of the right side with respect to left side. * = Significantly different compared to UP ($p < 0.05$). †= Significantly different compared to EXP ($p < 0.05$).

Figure 1

Click here to access/download;Figure;Fig 1.tif

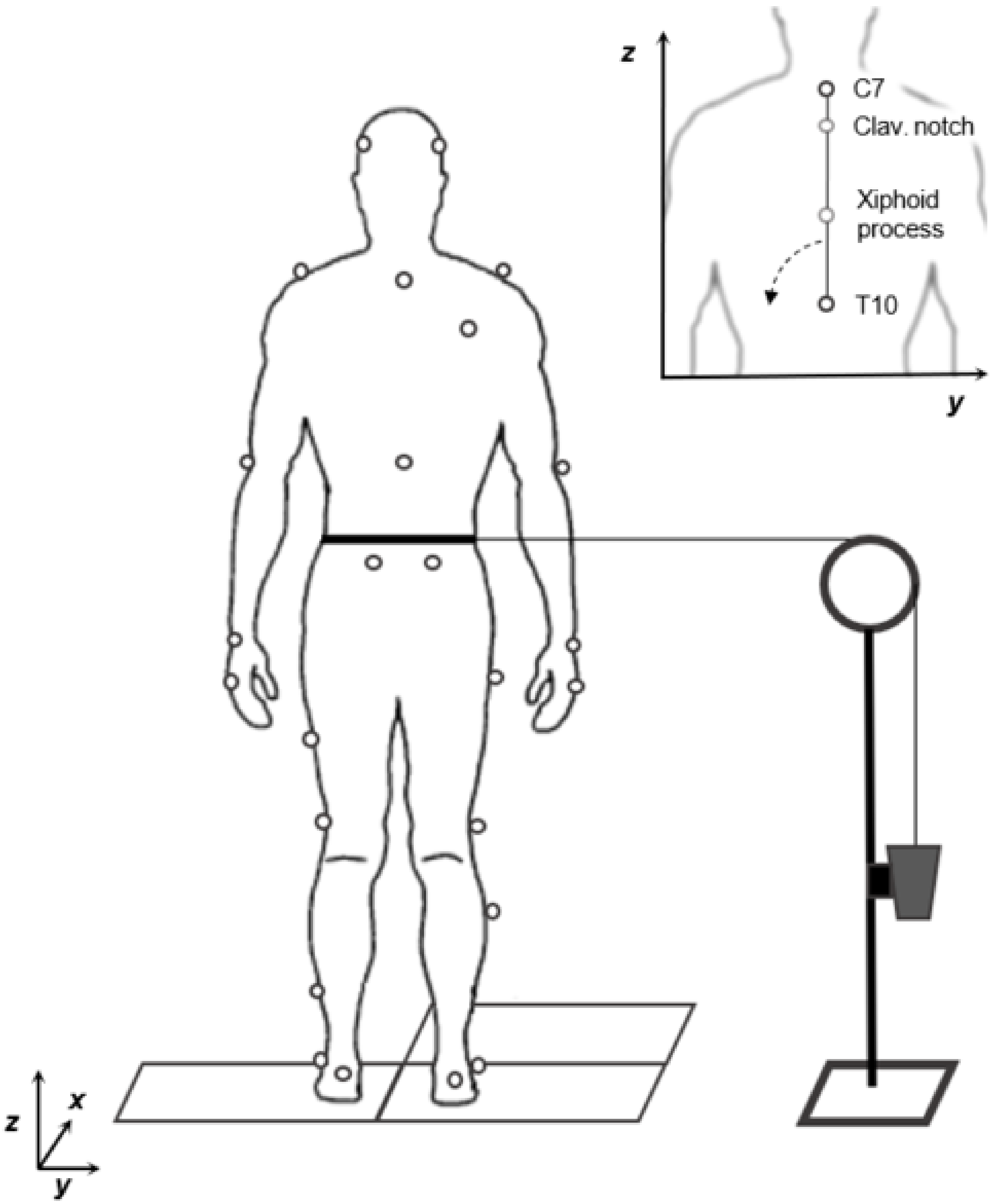


Figure 2

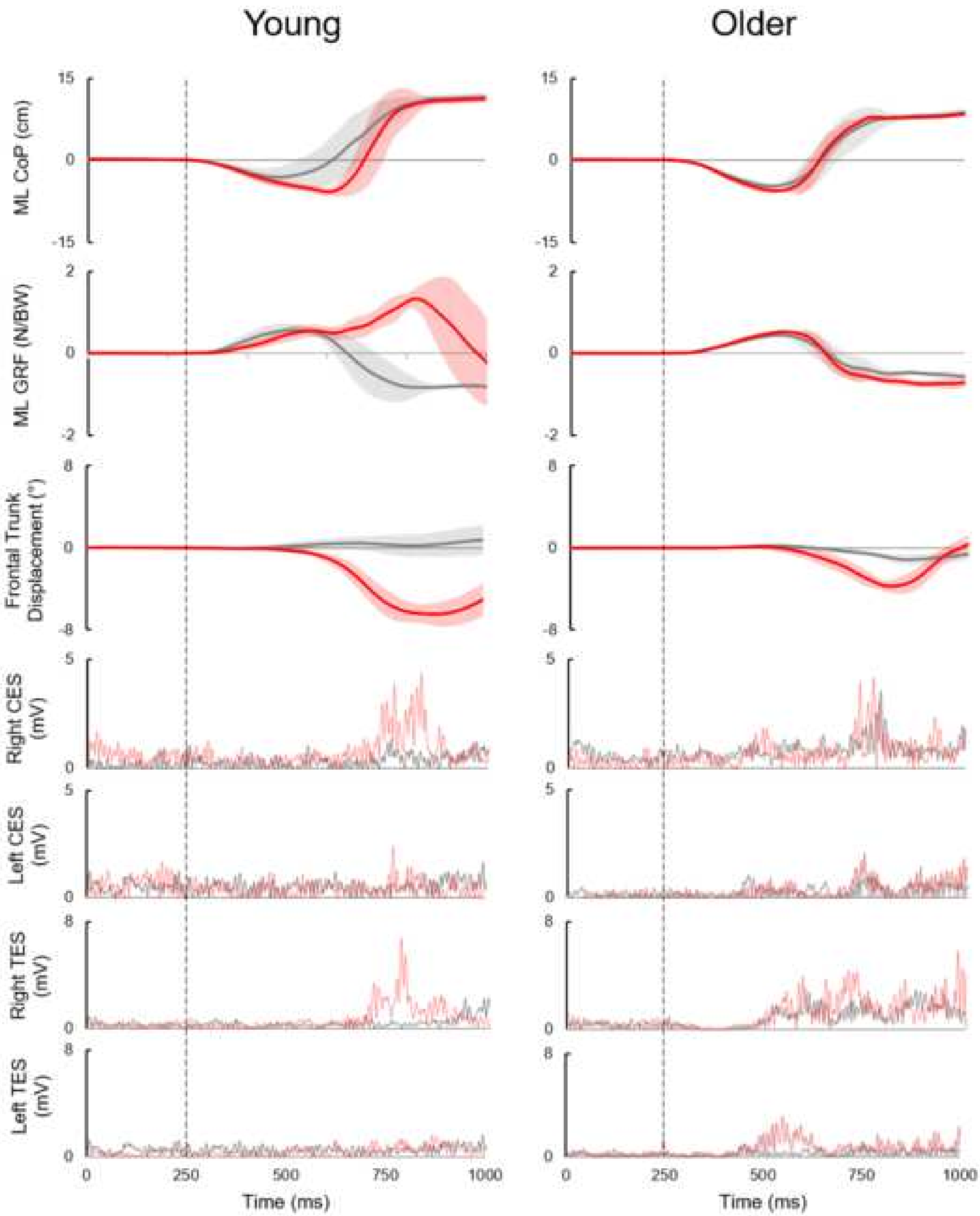
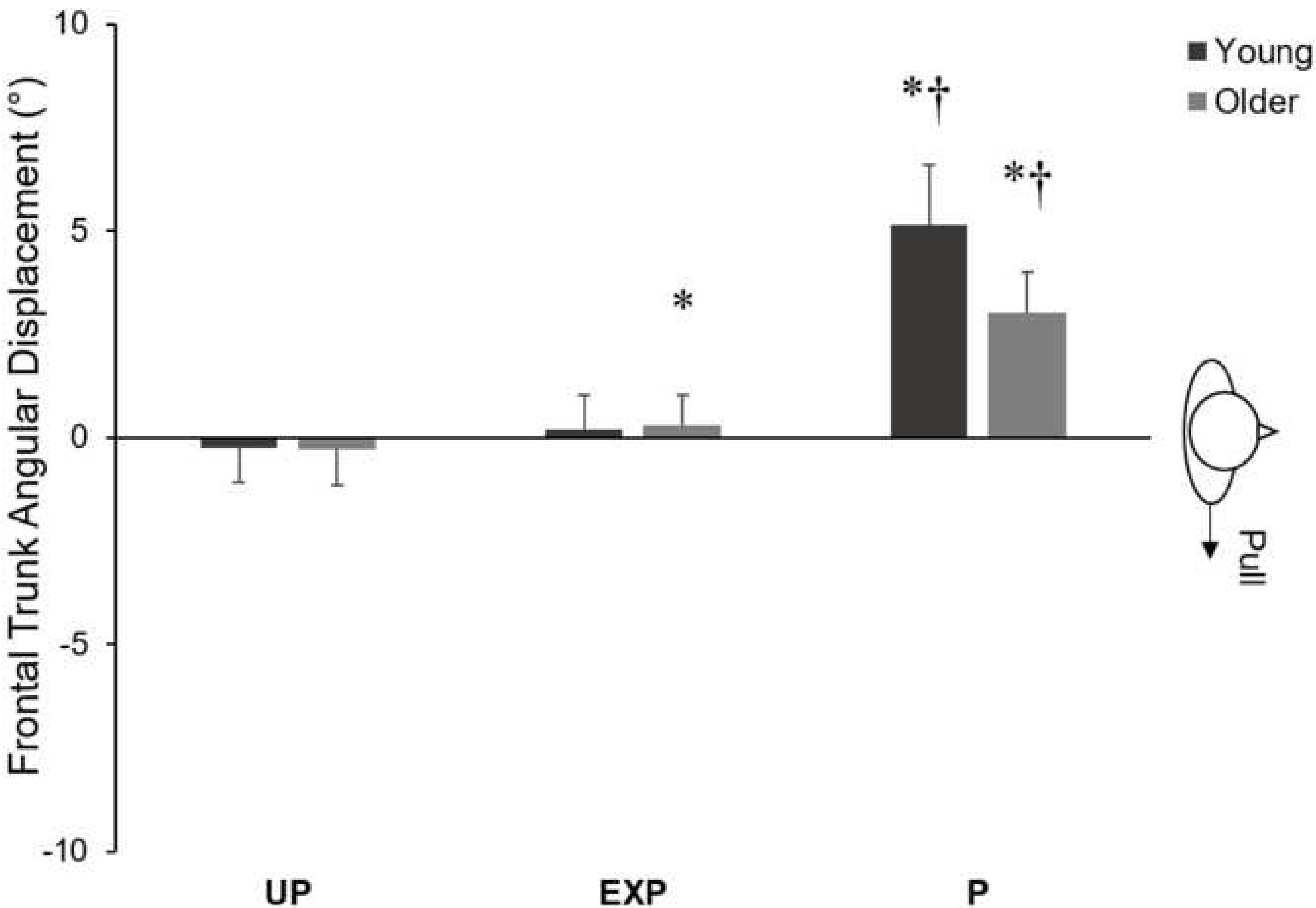

Young
Older
ML CoP (cm)
15
0
-15
ML GRF (N/BW)
2
0
-2
Frontal Trunk Displacement (°)
8
0
-8
Right CES (mV)
5
0
Left CES (mV)
5
0
Right TES (mV)
8
0
Left TES (mV)
8
0
0
250
500
750
1000
Time (ms)

Figure 3



10
5
0
-5
-10
Frontal Trunk Angular Displacement (°)
Young
Older
*†
*†
*
Pull
UP
EXP
P

Figure 4

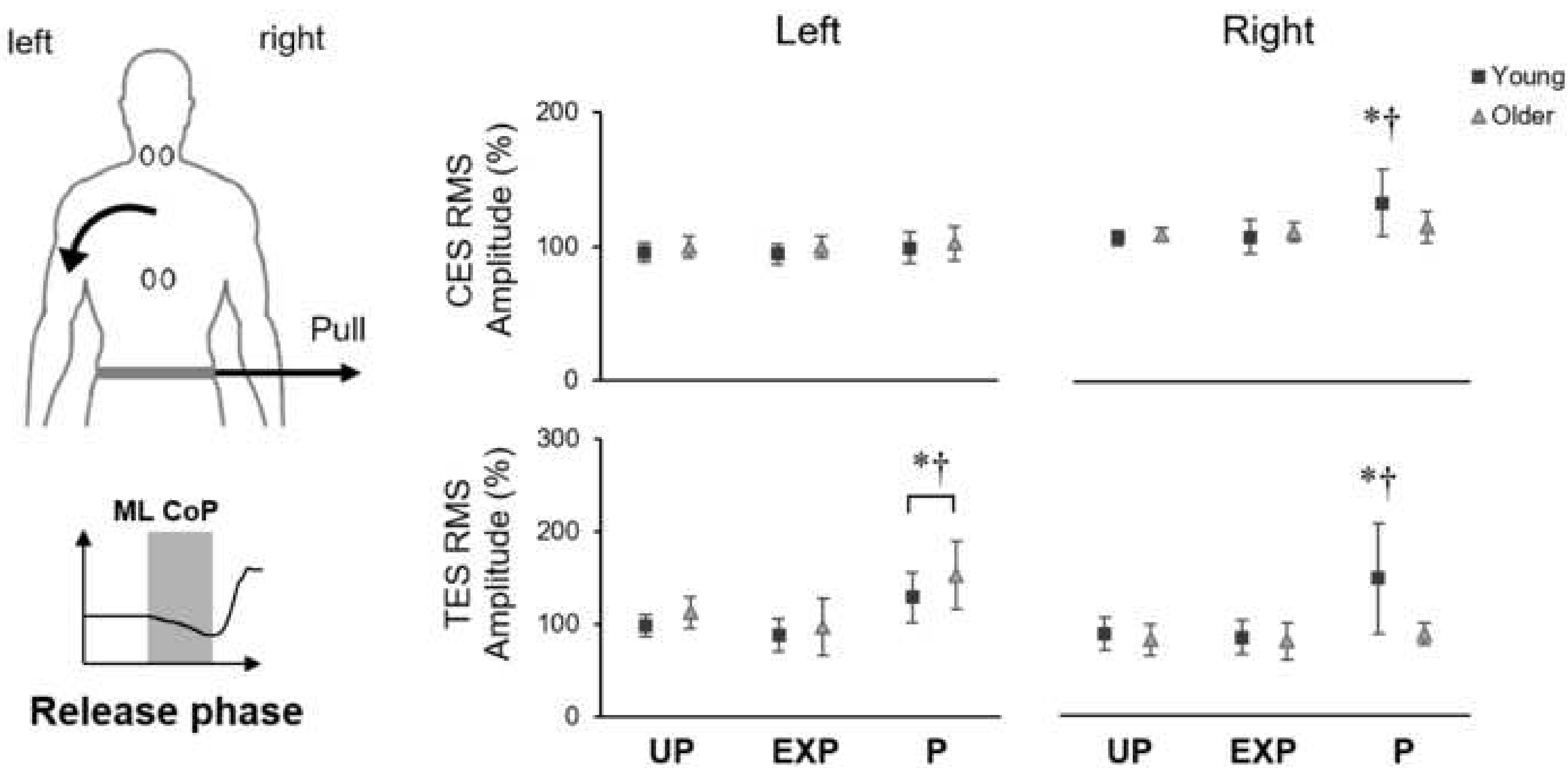
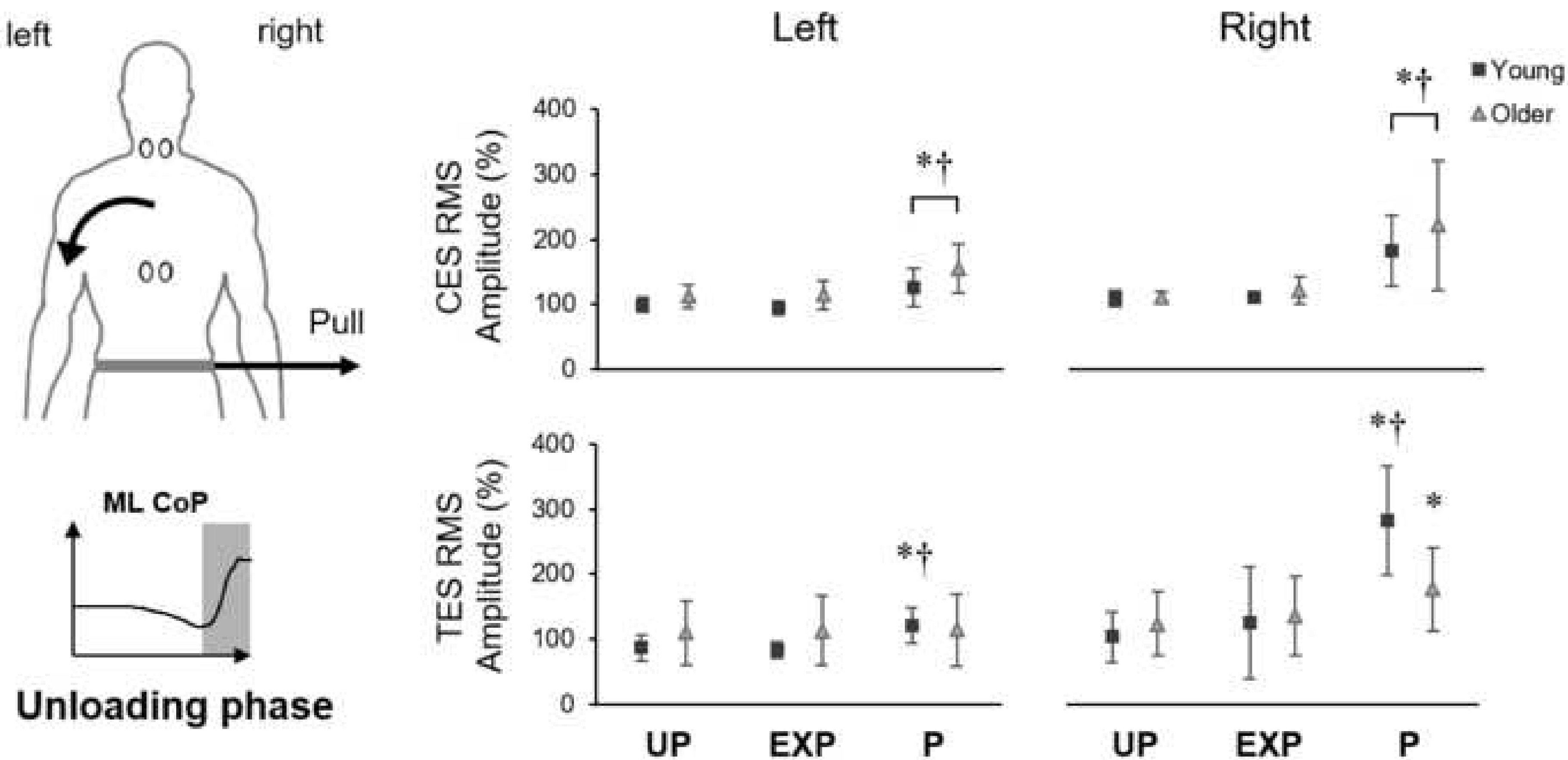

left
right
Pull
ML CoP
Release phase
Left
Right
Young
Older
CES RMS Amplitude (%)
TES RMS Amplitude (%)
200
100
0
300
200
100
0
*†
*†
*†
UP
EXP
P
UP
EXP
P

Figure 5


left
right
Pull
ML CoP
Unloading phase
Left
Right
Young
Older
CES RMS Amplitude (%)
TES RMS Amplitude (%)
400
300
200
100
0
*†
*
UP
EXP
P

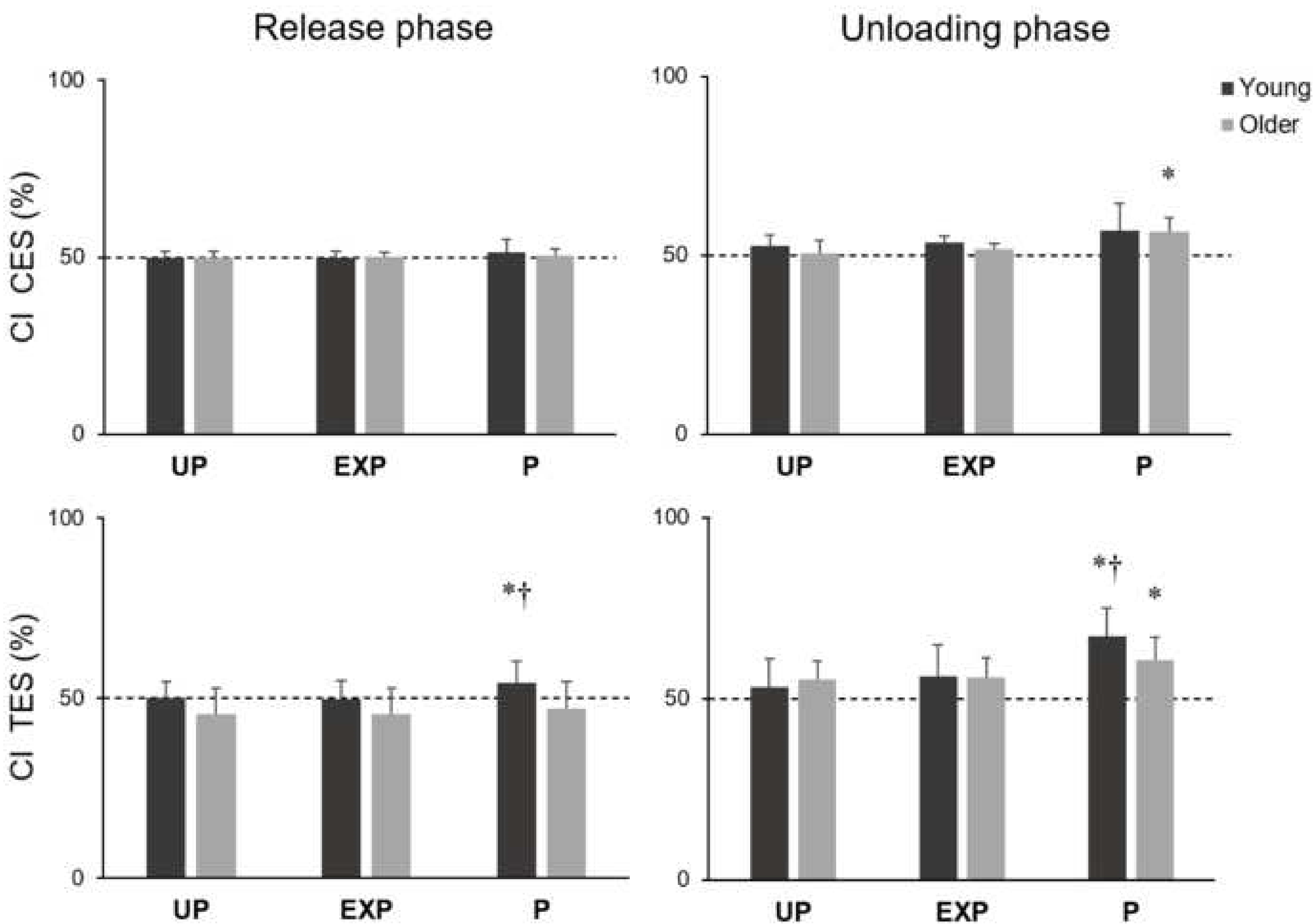

Release phase
Unloading phase
Young
Older
CI CES (%)
CI TES (%)
100
50
0
UP
EXP
P
*
*†
*†
*

Table 1. Mean ± SD of spatiotemporal parameters of CoP displacement and ML GRF normalized by body weight. * = Significantly different compared to UP ($p < 0.05$). †= Significantly different compared to EXP ($p < 0.05$).~~Significant changes in P and EXP conditions compared to UP condition within each group are shown in bold (pairwise comparisons, $p < 0.05$).~~

	Young			**Older**		
	UP	**EXP**	**P**	**UP**	**EXP**	**P**
ML CoP peak (cm)	4.3 ± 0.7	5.6 ± 1.5 *	9.3 ± 1.5 *†	3.0 ± 0.9	4.2 ± 1.0 *	5.4 ± 2.1 *
Release duration (ms)	276 ± 39	296 ± 63	420 ± 103 *†	260 ± 38	263 ± 40	289 ± 80
Unloading duration (ms)	316 ± 54	294 ± 48	245 ± 36 *†	323 ± 89	291 ± 59	228 ± 29 *†
ML GRF peak (N/BW)	0.5 ± 0.1	0.6 ± 0.2 *	1.1 ± 0.3 *†	0.4 ± 0.1	0.5 ± 0.1 *	0.6 ± 0.2 *
RFD (N/BW/s)	1.4 ± 0.4	1.8 ± 0.7	2.4 ± 0.6 *†	1.3 ± 0.3	1.6 ± 0.4	1.8 ± 0.4 *