Experimental Gerontology 181 (2023) 112271

Contents lists available at ScienceDirect

Experimental Gerontology

journal homepage: www.elsevier.com/locate/expgero

Lower back muscle activity during weight-shifting is affected by ageing and dual-tasking

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ARTICLE INFO

Section Editor: Carel Meskers

Keywords:
EMG
Ageing
Dual-task
Postural control
Weight-shifting

ABSTRACT

Purpose: Postural control deteriorates with age, especially under dual-task conditions. It is currently unknown how a challenging virtual reality weight-shifting task affects lower back muscle activity. Hence, this study investigated erector spinae neuromuscular control during mediolateral weight-shifting as part of an exergame during single- (ST) and dual-task (DT) conditions in young and older adults.

Methods: Seventeen young and 17 older adults performed mediolateral weight-shifts while hitting virtual wasps in a virtual environment with and without a serial subtraction task (DT). Center of mass position was recorded in real-time using 3D motion capturing. Electromyography recorded bilateral activation of the lumbar longissimus and iliocostalis muscles.

Results: Weight-shifting ($p < 0.03$) and targeting the wasps ($p < 0.005$) deteriorated with age and DT. Relative muscle activation during both quiet stance and weight-shifting increased with age, while the DT-effect did not differ consistently between age-groups. However, bilateral muscle co-contraction decreased with DT in young adults only. When switching direction and targeting the wasps, variability of muscle activation increased with age and DT and proved related to worse targeting performance. These effects were mainly visible at the non-dominant body side.

Conclusion: Older adults showed a higher erector spinae muscle contribution to perform weight-shifts with increased variability at the end of a shift, whereby muscle activity was modulated less well in older than in young adults in response to DT. Hence, the current findings point to the potential for developing postural training in which older adults learn to fine-tune trunk muscle activity to improve weight-shifting and reduce fall risk.

1. Introduction

With age, postural control deteriorates and the risk of falling increases (Park et al., 2016; Delbaere et al., 2010). Weight-shifting, which refers to the ability to transfer bodyweight without taking a step, is a component of anticipatory postural control (Horak, 2006). For instance, during gait initiation, a shift in bodyweight towards the stance leg is required before taking the first step, which is also known as anticipatory postural adjustments (Brenière and Do, 1991). Particularly, incorrect weight-shifting in the mediolateral direction was shown to be a major contributor to disequilibrium during gait and to falls (Cofré Lizama et al., 2015). The risk of falling increases further when adding cognitive load (Muhaidat et al., 2014; Li et al., 2018), as dual-task (DT) paradigms showed that a higher attentional demand was needed for postural control with ageing (Woollacott and Shumway-Cook, 2002). Exergames and virtual reality training provide potentially motivating interventions for

Abbreviations: ST, single-task; DT, dual-task; CoM, center of mass; EMG, electromyography; MVC, maximum voluntary contraction; CV, coefficient of variance; SPM, statistical parametrical mapping.

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https://doi.org/10.1016/j.exger.2023.112271
Received 9 December 2022; Received in revised form 19 May 2023; Accepted 11 August 2023
Available online 26 August 2023

training postural control (De Vries et al., 2018). Previous research by de Vries et al. (2020) indicated that such games could be used for strength training in older adults, although the intensity and response of the muscles to the game environment varies. Therefore, understanding the mechanism and impact of game-based exercise with and without a DT on the ageing neuromotor system is crucial to fully grasp the potential of such exergames. In a previous study (de Rond et al., 2021), we found that during a weight-shifting game, whereby subjects had to perform mediolateral weight-shifts to hit virtual wasps by center of mass (CoM) displacements, older adults showed different neural resource allocation in both motor and cognitive brain regions than young adults, especially when performing a concurrent cognitive task. These findings raise the question whether the deficits in central neural resources also affect the pattern of muscle recruitment involved in mediolateral weight-shifts.

Muscle strength is an important contributor to postural control (Laughton et al., 2003). Age-related changes in the neuromuscular system may result in an overall reduction in muscle strength (Hepple and Rice, 2016). To compensate for this deficit, older people tend to recruit more of their muscle capacity than young adults for the same motor tasks (Laughton et al., 2003). This may explain why increased relative muscle activation was found in a variety of leg muscles during both balance (Donath et al., 2016; Tokuno et al., 2009) and walking (Li et al., 2011) in older compared to young adults. Similar results were found for antagonist co-contraction ratios (Nagai et al., 2011; Hallal et al., 2013; Acuña et al., 2019), which is considered a feedforward strategy to enhance joint stiffness and compensate for the age-related neuromuscular deficits. So far, four studies looked at neuromuscular activity in the trunk (erector spinae or multifidus) during standing balance tasks (Donath et al., 2016; Tokuno et al., 2009; Shojaei and Bazrgari, 2017; Little and Woollacott, 2014), confirming that muscle activation was increased with age. Trunk muscles are intricately involved in dynamic balance as they not only ensure the erect position (Wilke et al., 1995) and stabilization (Mok et al., 2011) of the spine, but also restrict excessive trunk movements (Thorstensson et al., 1982). In line with these findings, chronic low back pain has been linked to a decreased efficiency of spinal movements as well as compromised postural control (Mok et al., 2011). Moreover, Masaki et al. (2016) found that cross-sectional erector spinae thickness was a determinant of sagittal spinal alignment and affected maximal walking speed in older women. Despite the crucial role of the lower back muscles during balance tasks, the effects of ageing on the lower back muscle activation during dynamic postural control, and specifically during mediolateral weight-shifting, have not been studied so far. Answering this question is important, as so far clinical guidelines on exercise for fall risk tended to emphasize lower limb muscle function, thereby omitting the trunk muscles (Tiedemann et al., 2011; Sherrington et al., 2020) and thus a potentially important contributor to balance problems and falls.

Six studies have investigated the effect of age and DT on neuromuscular activity in various leg muscles, including the hamstrings, quadriceps and/or calf muscles, during standing balance (Little and Woollacott, 2014; Rankin et al., 2000) or walking (Li et al., 2011; Hallal et al., 2013; Acuña et al., 2019; Fraser et al., 2007). One study did not find a difference in activation when adding a cognitive DT in both young and older adults (Acuña et al., 2019). However, three studies observed a higher activation (Li et al., 2011; Hallal et al., 2013; Little and Woollacott, 2014), and two studies showed a lower activation when comparing DT to single-task (ST), with greater muscle activation reductions in older compared to young adults (Rankin et al., 2000; Fraser et al., 2007). These differences may be task- and muscle-specific (Li et al., 2011; Hallal et al., 2013; Acuña et al., 2019; Little and Woollacott, 2014; Rankin et al., 2000; Fraser et al., 2007). The erector spinae was only included in the study of Little and Woollacott (2014) and showed inconsistent DT effects at various time points after anterior-posterior balance perturbations in older adults. Although the lower back muscles play an important role in postural control and may pre-empt stiffening, both directly and through attachments with the thoracolumbar fascia, the combined effects of ageing and cognitive load on the neuromuscular response to a dynamic balance task remains poorly understood.

Therefore, to answer these open questions, this study investigated the neuromuscular activity in the lower back muscles underlying mediolateral weight-shifting during ST and DT conditions in young and older adults during a virtual reality wasp game personalized to the individual's balance capacity. Based on previous research, we expected an overall higher muscle activation and co-contraction with age. However, for the effects of DT our hypothesis was less clear, as limited and conflicting findings were apparent. We also explored the effects of various movement phases on the EMG-patterns between groups and tasks. Finally, the EMG-patterns during weight-shifting were compared with a period of quiet stance.

2. Materials and methods

2.1. Participants

Participants included 17 young and 17 older adults as part of a previous functional Near-Infrared Spectroscopy (fNIRS) study on the effect of DT (de Rond et al., 2021). All participants signed an informed consent before starting the experiment. Inclusion criteria were: 1) 18–30 years old for young adults and 65–85 years old for older adults; and 2) able to stand upright without the use of an aid for at least 5 min. Exclusion criteria were: 1) cognitive impairment (MoCA <26); and 2) self-reported medical history, including neurodegenerative diseases, chronic balance and musculoskeletal impairments (e.g. low back pain), cardiovascular and respiratory conditions, impaired vision (uncorrected), acute pain, and diabetes. A detailed description of the demographics and cognitive and physical test battery can be found elsewhere (de Rond et al., 2021). In brief, we included the Flanker test (inhibition), Set-Shifting test (set-shifting) (Kramer et al., 2020), the Benton Judgement of Line Orientation test (visuospatial function) (Benton et al., 1994), the Mini Balance Evaluation Systems Test (Mini-BEST; postural control) (Franchignoni et al., 2010), the Falls Efficacy Scale-International (FES-I; fear of falling) (Yardley et al., 2005), and the SARC-F (sarcopenia) (Malmstrom et al., n.d.). Ethical approval was given by the Newcastle University Ethics Committee (6761/2018) and the study was performed in accordance with the ethical standards of the Declaration of Helsinki.

2.2. Experimental procedure and tasks

2.2.1. Study design

Participants were asked to perform mediolateral weight-shifts as part of a wasp game, developed in our center (Willaert et al., 2020) (see Fig. S.1A) in both ST and DT conditions. This virtual game required hitting wasps with a water jet, which was manipulated by moving the CoM beyond 80 % of the individual limits of stability. These stability limits, defined as the maximum CoM displacement while maintaining the base of support, were measured in eight directions prior to performing the wasp game. CoM position was determined online within the D-Flow software (Motek Medical BV, Amsterdam, The Netherlands; version 3.26) and captured within Nexus software (Vicon, Oxford Metrics, UK), based on reflective markers at the bilateral acromia, posterior superior iliac spines, lateral epicondyles, and lateral malleoli. Participants' foot position was fixed with a heel separation of 11 % body height and a 14 % angle (Cofré Lizama et al., 2014).

Wasps appeared alternately on the left and right side of a projector screen positioned at approximately three meters distance from the participants. They were instructed to hit as many as possible, by moving their CoM (CoM position was calculated in real time by the game). Meanwhile, patients were instructed to keep their arms crossed over their chest and moving from the ankle joints only to avoid compensation. No priority was assigned to weight-shifting speed or accuracy.

During the DT, a cognitive task was added (see Fig. S.1B) consisting of serial subtractions within a random interval of 2–5 s. This resulted in a minimum of eight and a maximum of 20 subtractions per trial. Young adults were asked to subtract sevens, and older adults were asked to subtract threes to equalize difficulty levels between groups (de Rond et al., 2021). Participants were instructed to only vocalize the final number in order to minimize interference with game performance. Therefore, performance on this serial subtraction task was rated as 'correct' or 'incorrect' at the end of each trial. The researcher assigned no priority to either the weight-shifting or the serial subtraction task. Both ST and DT conditions were performed alternately in five trials of 40 s, separated by a 30 sec period of quiet stance and were randomized within groups.

2.2.2. EMG assessment

For the assessment of the neuromuscular activity, electromyography (EMG) was recorded from the lumbar part of four muscles, split over two bilateral erector spinae muscles: longissimus lumborum and iliocostalis lumborum. We selected these muscles as they are involved in the dynamic control of sagittal and frontal trunk movements, due to their large moment arm on the spine (Lin et al., 2001) and because they are easily palpable and accessible for EMG assessment. First, the electrode placement sites were shaved when needed and cleaned with alcohol wipes. Disposable bipolar pre-gelled surface EMG electrodes (Ambu® Blue-Sensor P Ag/Ag-CI electrodes, Ballerup, Denmark) were placed over the muscle bellies following SENIAM guidelines (Hermens et al., 2000) (see Fig. 1A). EMG signals were then recorded using a wireless telemetric system (ZeroWire®, Aurion, Milan, Italy) with a sampling rate of 1000 Hz. The raw EMG signal (see Fig. 1B) illustrated that these muscles were involved in mediolateral weight-shifting as the left and right muscles were recruited alternatingly and in conjunction with the movement. Prior to weight-shifting, maximum voluntary contractions (MVCs) were measured. Participants were asked to stand upright with their anterior pelvis and abdomen pressed against the wall and perform back extension movements against resistance, three times for 10 s with 30 sec rest in between.

2.3. Data processing

2.3.1. Behavioral CoM data

Behavioral CoM data were split into a weight-shifting and a targeting phase. Processing of the weight-shifting phase was similar to the method described previously (de Rond et al., 2021). Briefly, CoM position was used to define the different phases (see Fig. 2). Weight-shifting started at the last peak of a left shift and finished at the first peak of a right shift and vice versa. At these peaks, the weight-shifting speed was zero. Hence, weight-shifting started after hitting the first wasp, and the number of weight-shifts differed between trials. When the wasp was not hit, a targeting phase was needed, which included corrections in both the mediolateral and anterior-posterior direction. Targeting was defined as the period from the first peak of a left shift to the last peak of a right shift and vice versa. In case the wasp was already hit during weight-shifting, the first peak was also the last peak and no targeting was needed, reflecting successful weight-shifting. Hence, the number of targeting phases differed from the number of weight-shifts within a trial. Outcome measures of the weight-shifting phase included the number of wasps hit, weight-shifting speed and CoM error (accuracy) (de Rond et al., 2021). Outcome measures of the targeting phase included the CoM path (distance travelled), duration, number of peaks and the number of targeting phases needed per trial. The number of peaks represented the number of times that participants changed weight-shifting direction before hitting the wasp. Thus, better targeting was reflected by a shorter path, shorter duration and less peaks before hitting the wasp. In case of a significant task effect without a significant interaction effect, the % DTcost was calculated for the age-groups separately, defined as the decline in performance from ST to DT, relative to ST performance ((ST − DT) / ST * 100) (de Rond et al., 2021; Friedman et al., 1982).

For a functionally meaningful analysis of the EMG signals, weight-shifting was divided into two sub-phases, reflecting that the game required returning to the midpoint (see Fig. 2). We distinguished initiation as the moment from the start of the weight-shift towards the baseline position (standing upright). Termination was defined as starting from the baseline position towards the end of the weight-shift. This was done separately for the right and left. As weight-shifting direction varied during the targeting phase, it was defined in accordance with the targeting side rather than the weight-shifting direction. Finally, the last 10 s of quiet stance in between weight-shifting trials were determined to be able to compare EMG signals during quiet stance with weight-shifting activity. All phases and corresponding outcome measures were extracted from the raw data in an automated fashion, using custom Matlab scripts (Matlab 2018b, Mathworks, MA, USA).

2.3.2. EMG data

Similar to the behavioral analysis, EMG data was analyzed with custom scripts using Matlab 2018b (Mathworks, MA, USA). First, raw data was band-pass filtered between 20 and 450 Hz (butterworth, 2nd order) (De Luca et al., 2010), rectified, and low-pass filtered (Butterworth, 6 Hz, 4th order). Data recorded during weight-shifting was then normalized to the isometric MVC, calculated as the mean of the three maxima determined during the three resisted extension trials. Mean normalized activation was calculated for each weight-shifting and targeting phase separately as well as for quiet stance. To better capture the variable nature of the targeting phase, the coefficient of variance (CV) was calculated as a measure of variability by dividing the standard

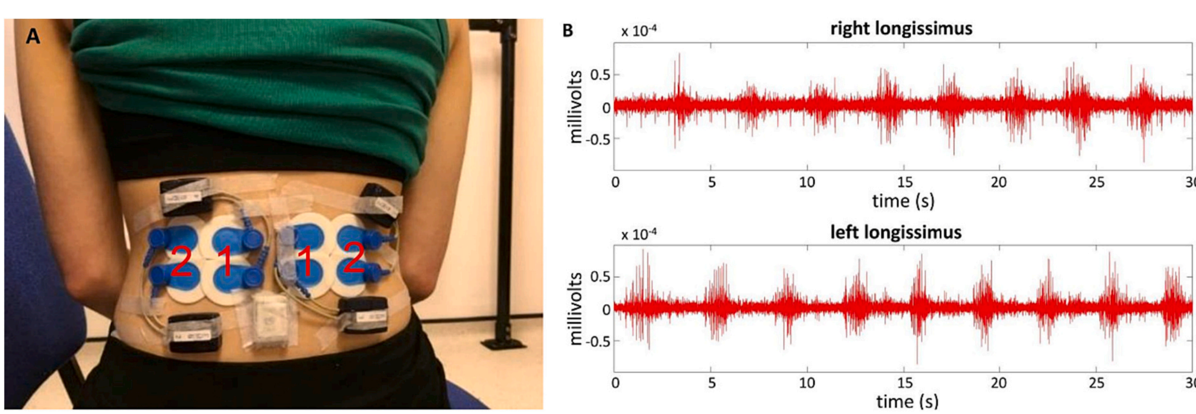

Fig. 1. EMG placement of the bilateral longissimus (1) and iliocostalis (2) of the erector spinae muscle according to SENIAM guidelines (A). Raw EMG signal during weight-shifting for the right and left longissimus in a representative young adult (B). EMG = electromyography.

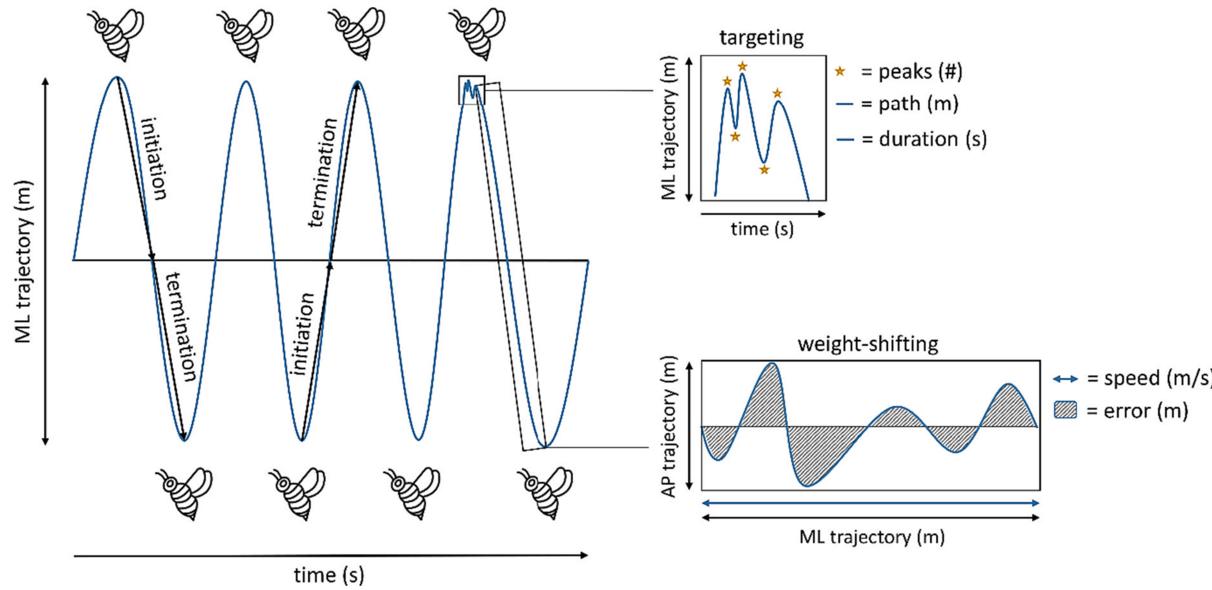

Fig. 2. Visualization of the weight-shifting and targeting phase over time, including corresponding behavioral outcome measures. CoM trajectory is displayed by the blue line. Outcome measures of the weight-shifting phase included speed (m/s) and anterior-posterior error (m), and the number of peaks, path (m) and duration (s) for the targeting phase. The weight-shifting phase was divided into initiation and termination. Initiation was defined as the movement from the last peak to the baseline (standing upright), termination from the baseline to the first peak, and targeting from the first peak to the last peak. Note that targeting was only present when the first and last peak did not overlap. CoM = center of mass, ML = mediolateral, AP = anterior-posterior.

deviation by the mean normalized EMG signal for each trial separately (Green et al., 2019). In addition, co-contraction ratios were calculated for the weight-shifting and targeting phases as the ratio of the least active side divided by the activation of the most active side.

2.4. Statistical analysis

Statistical analysis of participant characteristics and cognitive test scores were conducted with independent samples *t*-tests or Mann-Whitney *U* tests, depending on data normality. Outcome measures during ST and DT task conditions were analyzed in SPSS (IBM SPSS Statistics 28), using a Mixed Model approach to allow for missing values (Boisgontier and Cheval, 2016), as a targeting phase was not present in every trial. Moreover, in 8 out of the total 85 trials (17 participants * 5 trials), older adults did not stand completely still in between weight-shifts within the DT wasp game. Hence, these trials were excluded from analysis. To investigate the effect of age and DT on weight-shifting, EMG co-contraction and EMG variability (CV), we included group (young vs. older) and task (ST vs. DT) as fixed factors, and subject number as random factor. Main effects of group and task as well as a group * task interaction were modelled as effects of interest. For EMG co-contraction, the model was run for both weight-shifting phases (initiation, termination) and the targeting phase separately, as the muscle contributions varied over the different phases. Due to the variable nature of the targeting phase, EMG variability (CV) was investigated for this phase specifically. To consider the different phases while investigating the effect of group and DT on mean EMG, phase (stance, initiation, termination, targeting) was added as a fixed factor. Here, the main effects of phase, and interaction effects of group * phase and group * task * phase were added as effects of interest. For all EMG parameters, the model was run for both weight-shifting directions. In addition, weight-shifting speed was added as a covariate to control for these behavioral effects. Assumptions of homoscedasticity, linearity and normality of residuals were checked prior to running the models. A log-transformation was applied for the co-contraction parameters, as the assumptions were violated here. Bonferroni corrections were applied for post-hoc testing. Alpha was set at 5 % for significance.

As a secondary analysis, Statistical Parametrical Mapping (SPM) was used to compare the EMG patterns during weight-shifting on a continuous time scale (Robinson et al., 2015). For this analysis, a test statistic (SPM{t}) was calculated at each time node separately, as well as a critical significance threshold. For each time node, the calculation and interpretation was equal to scalar *t*-tests. The SPM approach was implemented using the open-source spm1d code (v.M0.3.2, www.spm1d.org) in Matlab (2018b, Mathworks, MA, USA). More specifically, two-tailed independent t-tests and two-tailed paired t-tests were used to compare activation patterns between groups during both ST and DT, and between ST and DT for both young and older adults ($\alpha = 0.05$), respectively. As the SPM analysis was exploratory, no Bonferroni corrections were applied. The targeting phase was not included in this analysis, as SPM requires comparable behavioral patterns, and the CoM targeting trajectory was highly variable both within and between participants.

Finally, a moderated multiple regression analysis was performed to investigate the relation between mean normalized EMG activity and CoM speed for the two weight-shifting sub-phases (initiation and termination) separately, and between variability (CV) in EMG activity and CoM path during targeting. For this analysis, data for ST and DT as well as left and right weight-shifting/targeting were averaged, creating four EMG parameters for ipsilateral iliocostalis, ipsilateral longissimus, contralateral iliocostalis and contralateral longissimus. First, all variables were standardized into z-scores. Second, CoM speed or targeting path was added as dependent variable, and the EMG parameters, group, and the interaction between the two were included as independent variables. As this regression analysis was exploratory, no Bonferroni corrections were applied.

3. Results

3.1. Participant characteristics

Older adults were shorter, and scored lower on the balance performance test (MiniBEST) and executive function (Set-Shifting and Flanker test), as previously published (see Table S.1) (de Rond et al., 2021). No differences between age groups were found for weight, sarcopenia (SARC-F), fear of falling (FES—I), cognitive ability (MoCA), and visuospatial function (Benton Judgement of Line Orientation test). Additionally, right leg dominance was reported by the vast majority of participants (young: 15/17, old: 16/17).

3.2. Task characteristics

We analyzed the impact of the task phases for the ipsi- and contralateral muscles to better understand the muscle function during the weight-shifting task. Active weight-shifting was accompanied by an increase in mean normalized EMG activation in all phases (initiation, termination, targeting) and muscles compared to stance (main phase effect p-values <0.001; post-hoc p-values <0.001; see Fig. S.2). Furthermore, a higher mean normalized EMG activation was seen during initiation compared to termination for the ipsilateral muscles, meaning that when shifting to the right, the right muscles were more active (post-hoc p-values <0.001; see Supplementary Fig. 2A), and when shifting to the left, the left muscles were more active (post-hoc p-values <0.020; see Table S.2). The opposite pattern was found in the contralateral muscles (post-hoc p-values <0.001), meaning that when shifting to the right the left muscles were more active during termination (see Fig. S.2B) and vice versa for the left shift (see Table S.2). Together, the ipsilateral muscles contributed mostly to initiation towards the baseline position and the contralateral muscles towards termination. During the targeting phase, the mean normalized EMG activation was higher for the contralateral muscles compared to initiation (post-hoc p-values <0.001; see Table S.2) and compared to termination for the ipsilateral muscles. The latter result was found for when shifting to the right only (post-hoc p-values <0.05; see Table S.2).

3.3. Between group behavioral results in ST vs. DT

3.3.1. Weight-shifting

Overall, young adults completed 14.04 ± 4.17 weight-shifts per trial during ST and 12.46 ± 3.25 during DT, whereas older adults only completed 9.93 ± 2.64 weight-shift during ST and 8.18 ± 2.53 during DT. As published previously (de Rond et al., 2021), stability limits were smaller in older compared to young adults (see Fig. S.3; all p-values <0.01). During weight-shifting, fewer wasps were hit and weight-shifting speed was lower in the older compared to the young adults (group effects; #wasps: $F_{(1,\ 32)} = 22.05$, $p < 0.001$; speed: $F_{(1,\ 32)} = 21.66$, $p < 0.001$; see Table 1) and in DT compared to ST (task effects; #wasps: $F_{(1,300)} = 125.13$, $p < 0.001$; speed: $F_{(1,300)} = 71.66$, $p < 0.001$; see Table 1). Although no group * task interaction effect was found, there was a higher %DT-cost with age in the number of wasps hit ($t_{(32)} = -2.03$, $p = 0.025$).

3.3.2. Targeting

Results on the targeting phase, not published previously, showed that older adults displayed longer targeting paths (group * task interaction effect: $F_{(1,225)} = 13.71$, $p < 0.001$; see Table 1) and a longer targeting duration (group * task interaction effect: $F_{(1,223)} = 8.51$, $p = 0.004$; see Table 1) in DT compared to ST than their younger counterparts. Post-hoc testing revealed worse targeting with age during DT only (path: $F_{(1,60)} = 11.62$, $p = 0.001$; duration: $F_{(1,57)} = 21.81$, $p < 0.001$). Furthermore, targeting paths increased in DT vs ST in older adults ($F_{(1,221)} = 4.23$, $p = 0.041$), whereas a decrease was found in young adults for both targeting path ($F_{(1,228)} = 9.93$, $p = 0.002$) and duration ($F_{(1,226)} = 2.67$, $p = 0.014$). In contrast, more targeting phases were needed with age during ST ($F_{(1,300)} = 13.46$, $p < 0.001$), while this number significantly decreased during DT for older adults only ($F_{(1,\ 40)}$ =, $p < 0.05$, group * task interaction effect: $F_{(1,300)} = 4.46$, $p = 0.035$; see Table 1). Targeting phases were shown during 50 % of total weight-shifts in young and 60.6 % in older adults during ST, and during 42.4 % in young and 51.8 % in older adults during DT. Moreover, all participants needed at least one targeting phase during ST, while three young adults and one older adult did not need any during DT. No differences were found in the number of peaks during the targeting phases.

3.4. Between group EMG results in ST vs. DT

3.4.1. Mean EMG

No three-way group * task * phase interactions were present for the mean normalized EMG activation. In all phases, mean normalized EMG was consistently higher in the older than in the young adults, including the stance phase prior to weight-shifting (post-hoc p-values <0.003; see Table S.2 and Fig. S.4). These group * phase interactions were significant for all muscles and weight-shifting directions (p-values <0.014). Of note, older adults and to a lesser extend also young adults recruited higher mean normalized EMG levels during all active weight-shifting phases (initiation, termination, targeting) compared to stance (post-hoc p-values <0.018), except for the contralateral muscles from stance to initiation in the young (see Fig. S.4B). The pattern of overall higher relative EMG activity with age was confirmed by our SPM analysis, as a higher activation was found in older compared to young adults during almost all time points of the normalized weight-shifting pattern (see Fig. 3 for weight-shifting to the right and Fig. S.5 for weight-shifting to the left). In short, older adults exhibited higher normalized EMG activity during both stance and active weight-shifting phases.

When comparing ST to DT, almost no group * task effects were found on mean normalized EMG activation, except for the ipsilateral longissimus during a shift to the right. Here, mean normalized EMG activation was lower in DT compared to ST for older adults specifically

Table 1
Weight-shifting outcomes for young and older adults during wasp game ST and DT.

	Young adults (N = 17)		Older adults (N = 17)		Group	Task	Group * task
	ST	DT	ST	DT	Main	Main	Interaction
Weight-shifting							
#wasps hit	14.56 ± 4.00	12.84 ± 3.11	10.12 ± 3.08	7.88 ± 2.96	**<0.001**	**<0.001**	0.155
speed (m/s)	0.110 ± 0.034	0.101 ± 0.029	0.067 ± 0.027	0.058 ± 0.020	**<0.001**	**<0.001**	0.628
error (m)	0.005 ± 0.002	0.006 ± 0.002	0.006 ± 0.002	0.006 ± 0.002	0.780	0.051	0.639
Targeting							
duration (s)	**1.349 ± 1.126**†	**0.729 ± 0.634***,†	**1.629 ± 1.255**	**2.187 ± 2.396***	**0.007**	0.488	**0.004**
path (m)	**0.061 ± 0.102**†	**0.013 ± 0.207***,†	**0.044 ± 0.061**†	**0.072 ± 0.120***,†	0.260	0.367	**<0.001**
#peaks[a]	1.347 ± 0.591	1.267 ± 0.439	1.433 ± 0.772	1.664 ± 1.348	0.056	0.519	0.182
#phases[b]	**1.459 ± 1.259**†	1.271 ± 1.322	**2.188 ± 1.749***,†	**1.494 ± 1.297***	0.168	**<0.001**	**0.035**

Note. Data are displayed as mean ± SD. Bold values indicate significant group main, task man and group * task interaction effects (p < 0.05). ST = single-task; DT = dual-task; CoM = center of mass.

* Significant post-hoc group-effect in ST and/or DT.
† Significant post-hoc task effect in young and/or older adults.
[a] Average within each targeting phase.
[b] Average within each trial.

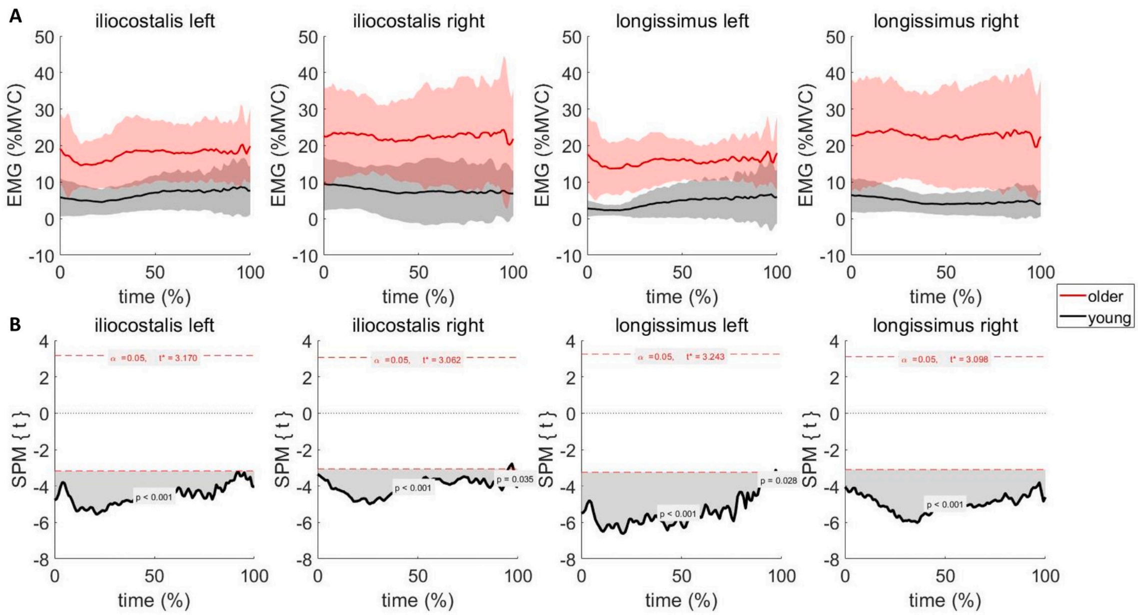

Fig. 3. Continuous EMG data for young and older adults during ST weight-shifting to the right used for SPM analysis (A), and the difference between groups based on SPM for each time point (B). EMG (%MVC) activation was higher in older vs young adults during the entire weight-shift (0–100 %, $p < 0.001$) for the iliocostalis left, and longissimus right. For the iliocostalis right, EMG (%MVC) activation was higher in older vs young adults during 0–95 % ($p < 0.001$) and 98–100 % ($p = 0.035$), and for the longissimus left during 0–96 % ($p < 0.001$) and 98–100 % ($p = 0.028$). MVC = maximum voluntary contraction.

(group * task interaction: $p = 0.028$; post-hoc: $p = 0.013$; see Table S.2). In contradiction, mean normalized EMG activation in the contralateral muscle was higher in DT compared to ST (task effects: p-values <0.029; see Table S.2), though only when shifting to the left. No consistent task-differences for either young or older adults were found in the SPM analysis.

3.4.2. EMG variability

During DT, a higher CV was found in older compared to young adults for the ipsilateral muscles while left targeting (group * task interaction p-values <0.038; post-hoc p-values <0.038; see Fig. 4A). Additionally, a higher CV in older compared to young adults was present in the contralateral iliocostalis while right targeting, independent of task load (group effect: $p = 0.002$; see Table S.3). These results indicate that variability increased with age, and this mainly for the muscles on the left

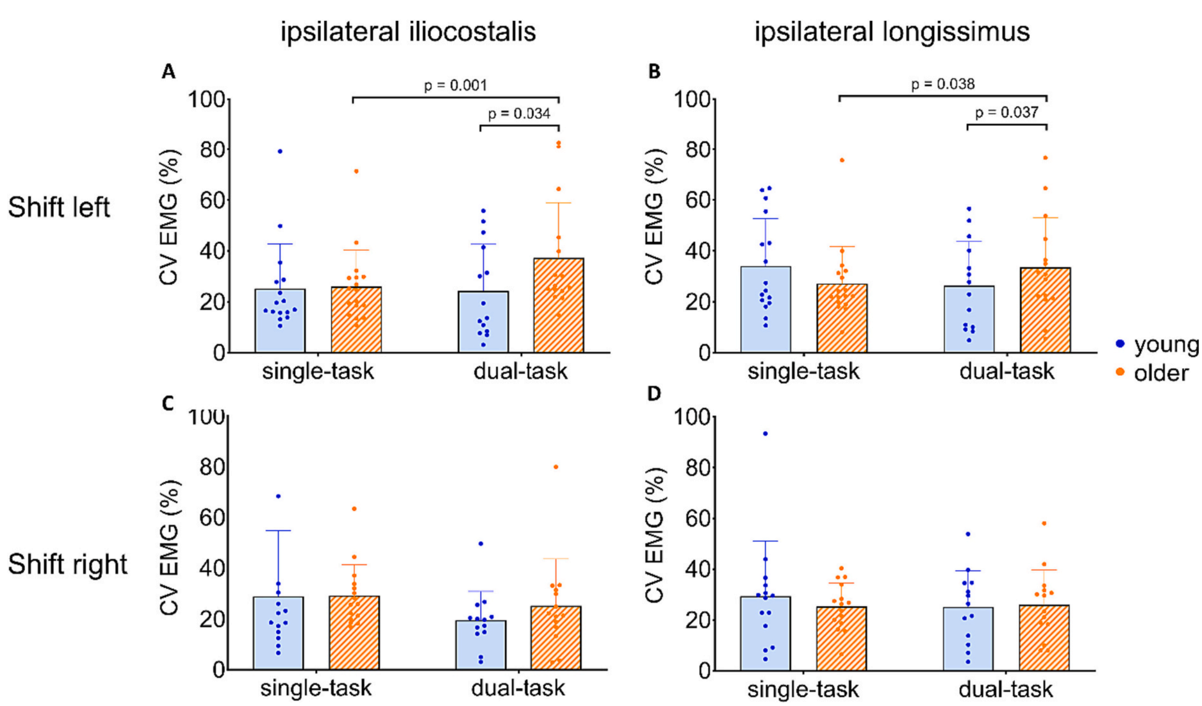

Fig. 4. Coefficient of variance (CV) EMG group * task effects for the ipsilateral iliocostalis and longissimus muscles during targeting on the left (A, B) and right (C, D) side. EMG data is normalized to MVC and displayed as mean ± SD. Individual data points represent EMG muscle activation data for each participant. Significant post-hoc effects are displayed with their corresponding p-values. MVC = maximum voluntary contraction.

side during DT. Furthermore, when comparing DT to ST, a higher CV was seen in older adults for the ipsilateral muscles during left targeting (post-hoc p-values <0.039; see Fig. 4A and B), whereas no such difference was found in young adults. These results were not found for the ipsilateral muscles during a right shift (see Fig. 4C and D).

3.4.3. EMG co-contraction

Co-contraction ratios were calculated as the ratio of activations of the least active side divided by the most active one. This resulted in a co-contraction ratio for the contralateral/ipsilateral muscle for the initiation phase and for the ipsilateral/contralateral muscles for the termination and targeting phases. No co-contraction ratio was calculated during stance, as here there is no clear distinction between the least and most active side. Co-contraction was lower in DT versus ST during the initiation and termination phase for both muscles and weight-shifting directions (task effects: p-values <0.015; see Table S.4) and for the longissimus during left targeting (task effect: p-value <0.001; see Table S.4). Interestingly, these task effects were mainly visible in young adults while leaning on the left leg; during initiation of a right shift, termination of a left shift and left targeting (group * task interactions: p-values <0.02; post-hoc: p-values <0.013; see Fig. 5A, B and C). During left termination, this task effect was also seen in older adults (post-hoc: p-values <0.001; see Fig. 5B). In addition, more longissimus co-contraction was found in older compared to young adults during DT while left targeting (group * task interaction: $p < 0.001$; post-hoc: $p = 0.012$; see Fig. 5C). Taken together, the effect of task and age was mostly present when leaning on the non-dominant left leg, whereas no group * task interaction effects were found when leaning on the dominant right leg (see Fig. 5D, E and F).

3.5. Relation between weight-shifting, targeting and muscle activation

To better understand the before mentioned results, we investigated the relation between behavioral and EMG parameters. This regression analysis showed that a higher CV for the bilateral iliocostalis muscles were related to a longer CoM path during targeting across age-groups (ipsilateral: $\beta = 0.58$, CI = 0.29–0.86, $p < 0.001$; contralateral: $\beta = 0.64$, CI = 0.34–0.94, $p < 0.001$; see Fig. 6). A similar patterns was found for the longissimus (ipsilateral: $\beta = 0.44$, CI = 0.10–0.77, $p = 0.012$; contralateral: $\beta = 0.46$, CI = 0.13–0.78, $p = 0.008$). Mean normalized EMG showed no significant relation with weight-shifting speed.

4. Discussion

4.1. Main study findings

This study is the first to look at the neuromuscular activity of the erector spinae muscles underlying mediolateral weight-shifting when young and older adults are engaged in a targeting game by moving their CoM sideways with and without a DT. We expected an overall higher muscle activation and co-contraction with age. Indeed, results showed that relative lower back muscle activation was increased during stance and all weight-shifting phases with age, though it was rather unaffected by DT. However, in the young the amount of muscle co-contraction decreased with DT and not in older subjects. The most striking age-differences were found in the targeting phase of weight-shifting with a DT, during which older adults revealed performance deficits. Here, EMG-variability was also increased during left DT-targeting in older adults, whereas this was not the case in young adults. Importantly, increased variability was found to be related to decreased targeting performance. Overall, this study adds to the literature that the modulation of lower back muscular activity to higher task demands is affected by ageing against a background of increased lower back muscle activity compared to the young during stance and weight-shifting.

4.2. Behavioral results

Despite the fact that we investigated a personalized weight-shifting task, that was personalized to individual stability limits, combined with an age-corrected cognitive subtraction task, we found age- and DT-related differences, especially for the number of wasps hit. We speculate that older adults may have been more afraid to move towards their stability limits compared to young adults (Scheffer et al., 2008). Specifically in the targeting phase, when the virtual wasps were aimed for, the targeting path and duration increased with age, indicating reduced efficiency in this more demanding task component. Adding the cognitive

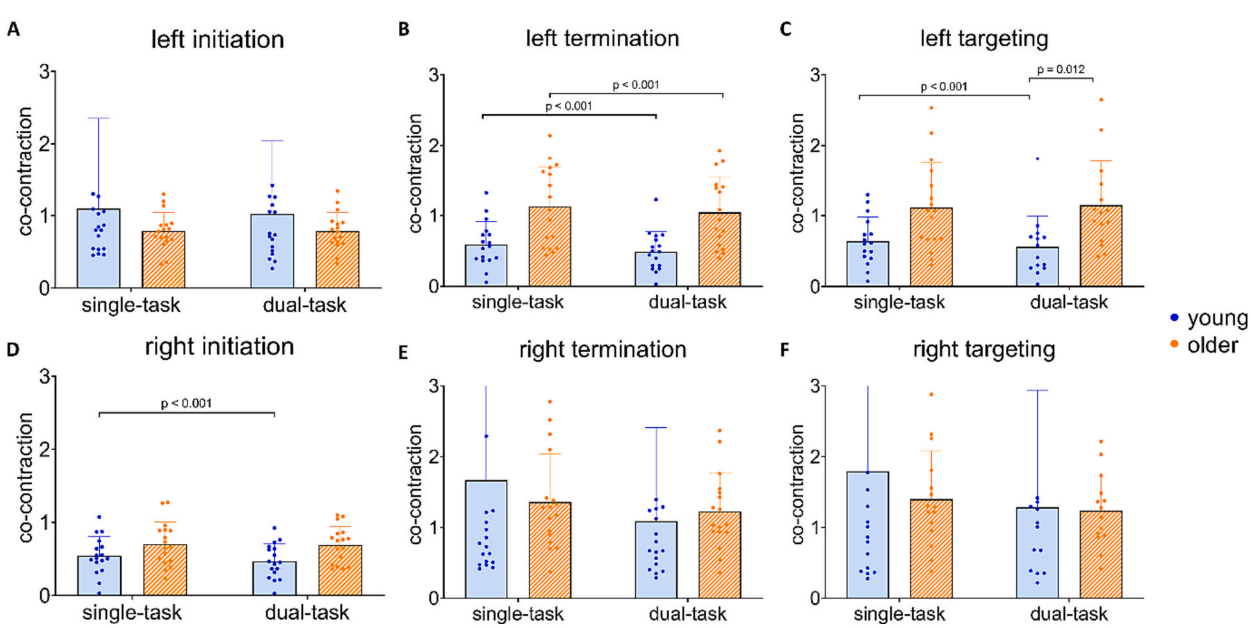

Fig. 5. EMG group * task effects for longissimus co-contraction during initiation (A), termination (B), and targeting (C) of a left shift, and initiation (D), termination (E), and targeting (F) of a right shift. EMG data is displayed as a ratio between the activation of the least active side divided by the activation of the most active side, and is displayed as mean ± SD. Individual data points represent EMG co-contraction data for each participant. Significant post-hoc effects are displayed with their corresponding p-values, statistics based on log-transformed data.

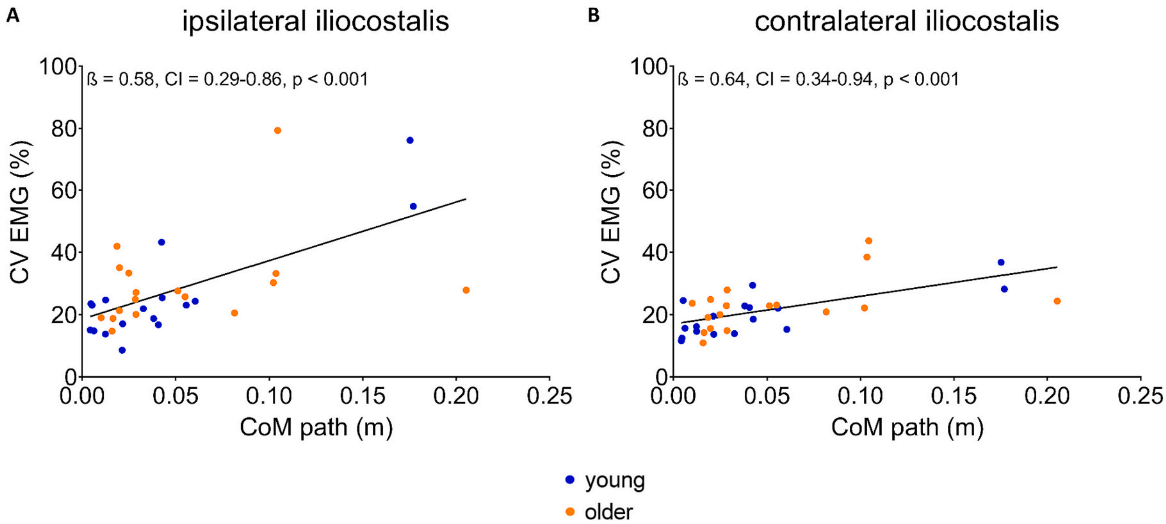

Fig. 6. Age-corrected regression between CoM pathway (m) and EMG variability (%), as defined by the CV, for the ipsilateral (A) and contralateral iliocostalis (B). Data for single- and dual-task as well as left and right targeting were averaged. CoM = center of mass. MVC = maximum voluntary contraction. CV = coefficient of variance.

DT seemed to positively influence young adults' performance, underscoring the intact ability to flexibly allocate attention to wasp aiming when needed (Scheffer et al., 2008).

4.3. EMG results

4.3.1. Muscle activation during weight-shifting

The findings from this study confirmed our hypothesis in that there was an overall higher relative muscle activation in the bilateral iliocostalis and longissimus muscles with age, both during weight-shifting and quiet stance. This is in accordance with results from previous studies showing higher activation in leg (Donath et al., 2016; Tokuno et al., 2009; Li et al., 2011) and trunk (Donath et al., 2016; Tokuno et al., 2009; Shojaei and Bazrgari, 2017) (erector spinae or multifidus) muscles during different balance and walking conditions. The general increase in relative muscle recruitment during quiet stance, before undertaking the wasp game, could be explained by age-related changes in the neuromuscular system. With age, there is a deterioration in muscle function, characterized by altered presynaptic inhibitory mechanisms (Laughton et al., 2003), a decrease in the number of motor units recruited, the proportion of type II muscle fibers, and an increase in fatty tissue (Crawford et al., 2017; Hunter et al., 2016). Other explanations for the age-related increase in relative muscle activation include that it could represent a way to enhance proprioception to strengthen the postural feedback loop (Laughton et al., 2003), or create a stiffening response during upright stance (Donath et al., 2016). However, we also found preserved alternating activity of ipsi- and contralateral muscles similar to that of young adults during weight-shifting.

Adding the cognitive DT did not consistently affect mean muscle activation differently between age-groups. Only two age-differences were found which were contradictory; mean normalized EMG increased in the contralateral muscle during a left shift and decreased in the ipsilateral longissimus during a right shift in older adults only. Previous research on leg and trunk muscle activation during balance and gait tasks also showed mixed results (Li et al., 2011; Hallal et al., 2013; Acuña et al., 2019; Little and Woollacott, 2014; Rankin et al., 2000; Fraser et al., 2007). These discrepancies may be attributed to the varied nature of the motor and cognitive tasks between studies as well as to the muscles investigated. Only one earlier study included erector spinae muscles during a postural perturbation task combined with a cognitive change-detection task (Little and Woollacott, 2014). The same study revealed no DT-effect in early perturbation and an increase of relative activity in the late perturbation during DT in older adults. As this involved platform perturbations and tested reactive postural control, these findings are difficult to compare with our mediolateral weight-shifting task testing anticipatory postural control.

4.3.2. Muscle activation and variability during targeting

Although previous studies addressed the importance of variability in muscle activation in the context of reliability (Green et al., 2019; Biviá-Roig et al., 2019), the meaning of variability in postural muscles during balance and weight shifting is less clear. During targeting, variability in muscle activation increased with age, especially during DT. As increased variability was related to worse targeting performance, we interpret this as an expression of compromised automatic motor control, which in line with a study on variability of brain activity (Maidan et al., 2022). The current results also showed higher variability from ST to DT in older adults, although not for all muscles, whereas no difference was found in the young. We suggest that the young have an intact ability to focus on a specific goal while maintaining weight-shifting. In contrast, older adults lack the motor and cognitive ability, to flexibly allocate neural resource during DT (Leone et al., 2017), expressed as increased variability of EMG in the trunk muscles. Interestingly, these DT effects were only seen during left targeting, which may be explained by the left leg being the non-dominant leg in almost all participants.

4.3.3. Co-contraction during weight-shifting

Greater bilateral longissimus co-contraction was found with age during DT left targeting specifically. Thus, while leaning towards the non-dominant left side. This age-related increase in co-contraction is in accordance with prior studies (Nagai et al., 2011; Hallal et al., 2013; Acuña et al., 2019). As regional longissimus activation contributes to stabilization of the spine, it might explain why this age-effect was found for this muscle specifically (Abboud et al., 2020). When comparing DT versus ST, longissimus and iliocostalis co-contraction decreased during initiation and termination, and during left targeting for the longissimus. However, there was no such impact found earlier during self-paced treadmill walking (Hallal et al., 2013; Acuña et al., 2019). Possibly, our postural task was more challenging and therefore able to show a DT effect. Interestingly, the decrease in co-contraction with DT was mainly visible in young adults while leaning on the left leg, signifying a more flexible recruitment in function of task demands (Leone et al., 2017). The fact that both age- and DT-related effects were most prominent while leaning towards the left side, may again point to the left leg being the non-dominant body side in almost all participants.

4.4. Task characteristics

Comparing the different weight-shifting phases revealed that the ipsilateral muscles are mostly active during initiation of the weight-shift and the contralateral muscles during termination. This is likely attributable to the role of the ipsilateral muscles as (co-)actuators at the onset of movement, while the contralateral muscles decelerate movement past the neutral position to maintain equilibrium. This pattern of concentric versus eccentric contraction might explain the fact that the majority of results were found on the ipsilateral side. Also, this pattern of results may be comparable to the triphasic agonist-antagonist-agonist muscle activation pattern during arm reaching, which is suggested to correspond to acceleration, deceleration, and stabilization, respectively (Ueyama, 2021). Furthermore, targeting appeared to elicit more contralateral muscle activation compared to initiation, and more ipsilateral muscle activation compared to the termination of the weight-shift. As direction switches are necessary during the targeting phase, it is reasonable to assume that both the ipsilateral and contralateral muscles were activated, albeit to different degrees.

4.5. Limitations and strengths

This study was based on a convenience sample, and due to the limited capacity of the available EMG system, it could only focus on two bilateral lower erector spinae muscles (iliocostalis and longissimus). We selected these muscles due to their importance for postural control and the limited evidence, so far, on their involvement during mediolateral weight-shifting. Nevertheless, the insights from this study can build a framework for future studies, including leg muscles and assessing muscle synergies (Baggen et al., 2020) to further investigate underlying differences in neuromuscular control during weight-shifting. EMG signals were normalized to isometric MVCs during stance, which could have been submaximal (Biviá-Roig et al., 2019), especially in older adults. Hence, increased (co-)activation could have been overestimated. As this study was part of a previous fNIRS study (de Rond et al., 2021), the weight and cables of this system could have affected weight-shifting performance and subsequent muscle activation patterns. However, both groups and task conditions suffered from this limitation. Outcome measures were averaged over all weight-shifts within a trial, which differed between subjects and age groups, possibly affecting the results found. As weight-shifting speed was not standardized between groups, we took this potential influential factor into account in our statistical model as a covariate.

4.6. Clinical implications

As the popularity of virtual reality training for rehabilitation is increasing, this study holds some important clinical implications. Results showed that the lower back muscles played an important role in weight-shifting control during gaming. The increase in relative muscle recruitment with age during weight shifting was already heralded by a higher relative activation in stance but importantly was modulated less than in young adults during increased task complexity. These results warrant future studies investigating whether specific training of the ability to adapt trunk muscle recruitment in older adults may improve postural control. Furthermore, leg dominance seemed to influence muscle activation patterns, particularly when adding cognitive distraction, warranting attention for this aspect in the future design of exergames. For instance, by measuring performance at the non-dominant and dominant sides separately and adjusting the game demands accordingly.

5. Conclusion

This study provided more insight into the contribution of the lower back muscles during a mediolateral weight-shifting game with and without extra cognitive distraction. Weight-shifting proved compromised with age and DT, even when scaling the task to individual stability limits. Older adults showed higher relative activation of the erector spinae muscles when standing quietly and performing weight-shifts, with a higher variability during multiple direction switches when dual tasking. In addition, weight-shifting direction and leg dominance played a role in the variability of muscle activation and the co-contraction patterns with age and DT.

Funding

This work was supported by the European Union's Horizon 2020 research and innovation programme under the Marie Sklodowska-Curie grant agreement (grant number 721577), and the Research Foundation Flanders (FWO) (grant number G0A5619N).

CRediT authorship contribution statement

Veerle de Rond: Methodology, Software, Formal analysis, Investigation, Writing – original draft, Writing – review & editing, Visualization. **Femke Hulzinga:** Software, Writing – review & editing. **Remco Johan Baggen:** Software, Writing – review & editing. **Aijse de Vries:** Software, Writing – review & editing. **Jean-Jacques Orban de Xivry:** Conceptualization, Methodology, Writing – review & editing, Supervision. **Annette Pantall:** Conceptualization, Methodology, Writing – review & editing, Supervision, Resources. **Alice Nieuwboer:** Conceptualization, Methodology, Writing – review & editing, Supervision, Funding acquisition.

Declaration of competing interest

None.

Data availability

The datasets generated during and/or analyzed during the current study are available from the corresponding author on reasonable request.

Acknowledgements

This work was supported by the European Union's Horizon 2020 research and innovation programme under the Marie Sklodowska-Curie grant agreement (grant number 721577), and the Research Foundation Flanders (FWO) (grant number G0A5619N). The authors would like to thank Esther Matupi and Leanne Kapa for their help with data collection, and Sabine Verschueren for supervising the wasp game development. Most importantly, the authors would like to thank all the participants for their essential contribution to this study.

Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.exger.2023.112271.

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