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Trunk kinematics and muscle activation patterns during stand-to-sit movement and the relationship with postural stability in aging

Woohyoung Jeon [a,b], Jill Whitall [b], Lisa Griffin [a], Kelly P. Westlake [c,*]

[a] Department of Kinesiology and Health Education, University of Texas at Austin, Austin, TX, United States
[b] University of Maryland School of Medicine, Department of Physical Therapy & Rehabilitation Science, Baltimore, MD, United States
[c] University of Maryland School of Medicine, Department of Physical Therapy & Rehabilitation Science, Baltimore, MD, 21201, United States

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ABSTRACT

Background: Stand-to-sit (StandTS) movement is an important functional activity that can be challenging for older adults due to age-related changes in neuromotor control. Although trunk flexion, eccentric contraction of the rectus femoris (RF), and coordination of RF and biceps femoris (BF) muscles are important to the StandTS task, the effects of aging on these and related outcomes are not well studied.
Research question: What are the age-related differences in trunk flexion, lower extremity muscle activation patterns, and postural stability during a StandTS task and what is the relationship between these variables?
Methods: Ten younger and ten older healthy adults performed three StandTS trials at self-selected speeds. Outcomes included peak amplitude, peak timing, burst duration, and onset latency of electromyography (EMG) activity of the RF and BF muscles, trunk flexion angle and angular velocity, whole body center of mass (CoM) displacement, center of pressure (CoP) velocity, and ground reaction force (GRF).
Results: There were no age-related differences in weight-bearing symmetry, StandTS and trunk flexion angular velocity, or BF activity. In both groups, EMG peak timing of RF was preceded by BF. Compared to younger adults, older adults demonstrated shorter RF EMG burst duration, reduced trunk flexion, and reduced stability as indicated by the longer duration in which CoM was maintained beyond the posterior limit of base of support (BoS), greater mean anterior-posterior CoP velocity and larger standard deviation of CoM vertical acceleration during StandTS with smaller vertical GRF immediately prior to StandTS termination. Trunk flexion angle and RF EMG burst duration correlated with stability as measured by the duration in which the CoM stayed within the BoS.
Significance: Decreased trunk flexion and impaired eccentric control of the RF are associated with StandTS instability in aging and suggest the importance of including StandTS training as a part of a comprehensive balance intervention.

1. Introduction

Sit-to-stand (STS) and stand-to-sit (StandTS) movements are important functional daily activities that are commonly used in clinical settings to assess mobility [1,2]. Though StandTS initially appears to be a reverse movement of STS, muscle activation patterns and force contributions during the descending phase of StandTS are different from STS [3,4]. During STS, the initiation of trunk flexion occurs prior to the onset of knee extension, whereas with StandTS, there is little difference in the timing of trunk flexion and knee flexion, requiring almost simultaneous control of the anterior-posterior (A-P) and vertical displacement of whole body center of mass (CoM) [5]. Therefore, control of trunk flexion and eccentric contraction of the rectus femoris (RF) play important roles during StandTS in order to precisely move CoM downward and backward while also controlling descent against gravity [6,7]. In relation to postural stability, control of trunk flexion is important for controlling the vertical downward acceleration phase of StandTS [6] and eccentric contraction of the RF plays a major role in decreasing the downward velocity to enable a stable and safe landing [8]. In addition, co-contraction of RF and biceps femoris (BF) provides postural stability through efficient control of CoM direction in the frontal and sagittal planes [9].

* Corresponding author.
E-mail addresses: wjeon@som.umaryland.edu (W. Jeon), jwhitall@som.umaryland.edu (J. Whitall), l.griffin@austin.utexas.edu (L. Griffin), kwestlake@som.umaryland.edu (K.P. Westlake).

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Considering the motor control demands of the StandTS task and known age-related changes in muscle composition and neuromotor mechanisms, StandTS task becomes more challenging for older adults. For example, loss of muscle mass and change in muscle fiber type with age contribute to a decline in lower extremity muscle strength [10]. In addition, age-related changes in neuromuscular, and central and peripheral sensorimotor systems result in a gradual decrease in motor control [11,12]. Thus, alterations in neuromuscular control and insufficient lower limb muscle strength can lead to delays and inefficiencies in motor performance, thereby challenging the ability to maintain postural stability during StandTS [13,14]. Indeed, frail older adults demonstrate both a decrease in trunk movement and a loss of StandTS smoothness with longer StandTS transition times compared to healthy controls [15]. Moreover, compared to younger adults, older adults demonstrated smaller trunk flexion angle during a normal speed StandTS and slower angular velocity of trunk flexion during a fast StandTS, which may be a compensatory strategy to prevent disequilibrium [3,14].

Although a lack of trunk movement and altered task timing have been demonstrated during StandTS in older adults, an investigation of age-related changes in trunk movement along with other age-related changes in neuromuscular activation patterns of the lower extremity musculature, and the relationship of these changes to postural stability remain unknown. Thus, the purpose of this study was to investigate age-related differences in trunk flexion, lower extremity muscle activity, and postural stability during a StandTS task. Secondly, we aimed to determine the relationship between trunk flexion angle and eccentric RF muscle activation with postural stability. During StandTS, we hypothesized reduced trunk flexion and decreased EMG peak amplitude and burst duration of the RF and BF in older adults compared to younger adults, which would be associated with postural instability. Specifically, we expected that insufficient trunk flexion and reduced eccentric RF muscle contraction would be related to postural stability as indicated by the duration in which CoM remained inside the base of support (BoS) prior to landing.

2. Methods

2.1. Participants

Ten healthy younger adults (5 men, 5 women, 20 ± 1 years) and ten healthy older adults (5 males and 5 females, 77 ± 7 years) participated. All procedures were approved by the Institutional Review Board at the University of Texas at Austin and were in accordance with the Helsinki Declaration of 1975.

2.2. Data collection

A Bagnoli electromyography (EMG) System (Delsys, Inc) was used for acquisition of muscle activity signals. Adhesive pre-gelled Ag/AgCl surface EMG electrodes (Delsys Inc., Boston, MA) were placed bilaterally on the RF and long head of biceps femoris (BF). The positioning of the electrodes was in accordance with Rainoldi et al. [16]. Participants performed three maximal voluntary isometric contractions (MVIC) for normalization of EMG data (% EMG max). For RF, participants seated on a chair (hip and knee joints at 90°) with a cuff around their ankle and MVIC knee extension was measured. For BF, participants lay on a massage table in prone position and performed MVIC knee flexion at 45° knee flexion against manual resistance applied to ankle [17]. Thirty-nine reflective markers were placed on the body according to a full body Plug-In Gait. A 10-camera motion capture system (VICON Motion Systems Ltd, UK) was used to record body kinematics.

At the start of StandTS, participants maintained an upright standing position with the feet shoulder-width apart and placed their feet in a self-selected parallel foot position with each foot on a force plate (Bertec Corporation, Columbus, OH, USA). Participants performed three StandTS trials after a visual light cue was turned on with instructions to "sit down at your comfortable speed". There was a rest period of two minutes between trials.

Participants crossed their arms over their chest during the StandTS task and were instructed to sit down on an armless, backless height-adjustable bench with the back of the knees not touching the bench. Seat height was adjusted as the distance from the center of the knee joint to the floor for each individual. A pressure-sensitive pad (Microgate, NY, USA) was placed on the seat to determine seat-on timing.

2.3. Data processing

EMG, force plate, and pressure pad data were sampled at 1,200 Hz and kinematic data was sampled at 120 Hz. All data were analyzed in Matlab 9.3 (Matworks Inc., Natick, MA, USA).

2.3.1. Kinematics and kinetics

2.3.1.1. Kinematics. Body CoM trajectory and trunk angle were calculated using Vicon Nexus 1.8.5 Software (Vicon, Oxford Metrics, UK). To calculate StandTS speed, the movement distance of the CoM in the sagittal plane was divided by the elapsed time from the onset of CoM movement to StandTS termination. Standard deviation of the CoM acceleration (variation of CoMAccel) in the vertical direction during a StandTS was used to quantify the vertical body oscillation [18]. Trunk flexion angle was defined as the angle between the thorax and the laboratory coordinate system (Plug-in Gait Model). Bilateral heel markers were used to define the posterior border of the BoS.

StandTS initiation was defined as the instant when A-P GRF exceeded 2.5 % of the peak-to-peak value during the task. StandTS termination was defined as the instant when A-P GRF variations were within 1% of its steady state value as calculated during quiet sitting [6]. The descending phase was defined as the time between StandTS initiation and the start of the stabilization phase. The start of the stabilization phase was defined as the instant when the downward velocity of the vertical CoM reached its minimum value and the end of stabilization phase occurred at the time of the StandTS termination.

2.3.1.2. Kinetics. Center of pressure (CoP) and GRFs were recorded from the two force plates. Peak GRFs in the A-P and vertical directions during a StandTS were calculated and normalized to body mass (kg). The mean velocity of CoP in the A-P and medio-lateral (M-L) directions during a StandTS was calculated to quantify the A-P and M-L postural sway [18]. GRF data were filtered with 4^{th} order low-pass Butterworth filter with a cut-off frequency of 25 Hz [19]. CoP path data (the A-P and M-L time series) were filtered with a 4^{th} order low-pass Butterworth filter with a cut-off frequency of 6 Hz [20].

2.3.1.3. Weight-bearing symmetry. Weight-bearing symmetry during a StandTS was calculated from the vertical GRFs immediately prior to the stabilization phase using the following equation [21]:

$$\text{Symmetry }(\%) = 100 - \frac{|Dominant\text{max} - \ Non\ dominant\ max| \cdot 100}{Dominant\text{max} + \ Non\ dominant\ max}$$

2.3.2. EMG

Data acquired from surface EMG electrodes were filtered with a 5–500 Hz band-pass filter. A root mean square with a 100 ms window was applied as a digital smoothing algorithm [22]. The EMG signals were then normalized to the MVIC EMG. Each muscle's onset time was determined as the time at which the EMG exceeded three standard deviations (SD) of the initial mean baseline [23].

2.4. Statistical analysis

The mean of three trials was used for all EMG, kinematic, and kinetic analyses. A one-way multivariate analysis of variance (MANOVA) was used to determine group differences (older vs. younger) in EMG activity of the RF and BF. A one-way repeated measures ANOVA was used to compare the EMG activity between the RF and BF. A one-way multivariate analysis of covariance (MANCOVA) was used to compare kinematics and kinetic parameters between the two groups with weight and height as covariates [24]. Spearman's correlation (ρ) was run to estimate the correlation between trunk flexion angle and duration in which the CoM remained within the BoS. Tukey's test was used for post hoc analyses. SPSS (Chicago, IL) was used for all statistical analyses with a-priori an alpha level of 0.05.

3. Results

All data are presented as mean ± SD in the text. No difference was observed in weight-bearing symmetry across all participants. The descending phase was divided into two phases based on the CoM vertical displacement velocity. The vertical downward velocity of the CoM increased during the first phase and decreased during the second phase. The phase divisions are represented in Fig. 1, which displays exemplars of younger and older adult.

3.1. Kinematics

3.1.1. StandTS velocity

There was no difference in StandTS velocity between the two groups (older: 254.76 ± 61 mm/s younger: 306.66 ± 53.50 mm/s, $p = 0.06$).

3.1.2. Trunk flexion and CoP displacement during StandTS

Compared to younger adults, older adults demonstrated reduced trunk flexion angles (older: 36.39 ± 3.39°, younger: 41.83 ± 6.64°, $p = 0.03$) whereas there was no difference in maximal angular velocity of trunk flexion (older: 77.46 ± 22.38°/s, younger: 82.03 ± 12.53°/s, $p = 0.58$). Older adults showed reduced anterior CoP displacement (older: 75.10 ± 17.44 mm, younger: 90.90 ± 10.40 mm, $p = 0.04$). In the M-L direction, there was no difference in trunk angle (older: 2.97 ± 1.45°,

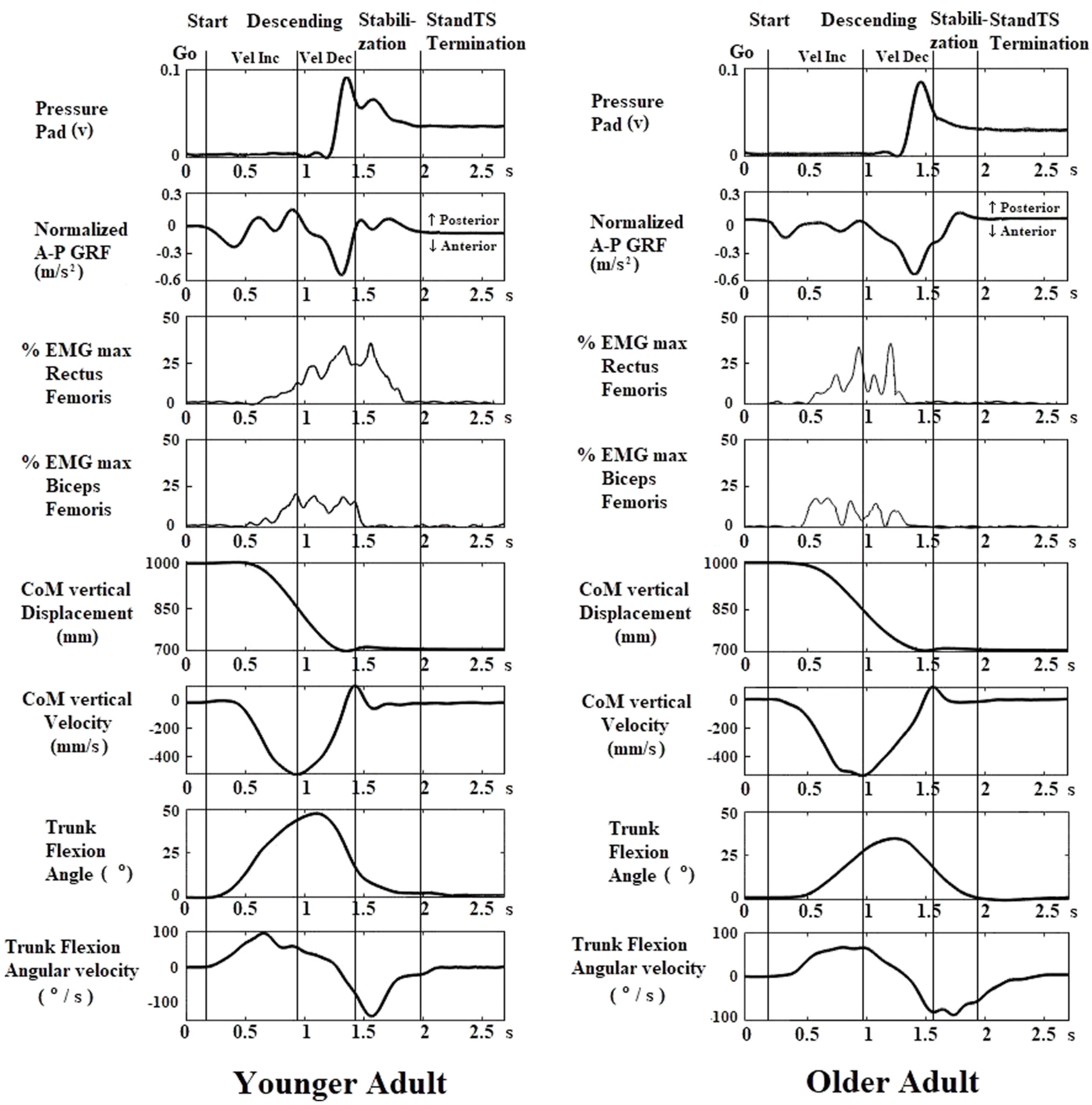

Fig. 1. Trunk and whole body CoM kinematics and normalized EMG activity of the rectus femoris and biceps femoris muscles from a representative younger and older adult during StandTS. Horizontal axes display time, and "0 s" represents "Go" signal. Vertical lines depict the time points of StandTS start, transition between increasing and decreasing CoM vertical velocity during the descending phase, start of the stabilization phase, and start of the StandTS termination. Vel Inc: Velocity Increment, Vel Dec: Velocity Decrement. v: voltage, s: second.

younger: 2.49 ± 1.12°, $p = 0.42$) and CoP displacement (older: 22.03 ± 7.99 mm, younger: 19.69 ± 3.13 mm, $p = 0.39$) between age groups.

3.1.3. Timing of whole body CoM excursion outside the BoS

The elapsed time from the instant when CoM moves beyond the posterior limit of the BoS immediately prior to StandTS termination was longer for older compared to younger adults (older: 0.87 ± 0.22 s, younger: 0.65 ± 0.15 s, $p = 0.02$).

3.2. Kinetics

3.2.1. Ground reaction force

Maximum vertical GRF during StandTS was reduced in older compared to younger adults (older: 31.53 ± 8.24 %, younger: 42.75 ± 11.14 %, $p = 0.03$).

3.2.2. StandTS stability

The mean velocity of CoP in the A-P direction was greater for older adults compared to younger adults ($p < 0.01$) whereas there was no difference between age groups in the M-L direction ($p = 0.35$). In the vertical direction, variation of CoMAccel was greater for older adults compared to younger adults ($p = 0.03$) (Fig. 2).

3.3. EMG

Compared to younger adults, the EMG burst duration of the RF during the StandTS was reduced in older ($p = 0.023$), whereas there was no difference in the BF. The RF peak timing of younger adults was closer to StandTS termination compared to older adults. The time difference between the first RF peak timing and StandTS termination was longer in older adults (1.01 ± 0.22 s, $p < 0.01$) compared to younger adults (0.66 ± 0.19 s). There was no difference in EMG peak amplitude of the RF and BF between age groups (Table 1).

In both younger and older adults, the EMG peak amplitude of the RF was greater than the BF ($p < 0.01$) with no difference in EMG onset between the RF and BF muscles. However, the EMG peak timing of the BF was earlier than the RF (the time difference between the first peak of the BF and RF: 0.37 ± 0.24 s, $p < 0.01$) (See Fig. 1).

3.4. Correlation of trunk flexion and RF activity with CoM position in relation to the BoS

There was a significant positive correlation between trunk flexion angle and duration in which the CoM remained within the BoS ($r_s(18) = 0.77$, $p < 0.001$). In other words, reduced trunk flexion angle correlated

Table 1
EMG activity of the rectus femoris and biceps femoris during stand-to-sit between age groups.

	EMG Onset Latency (sec)		EMG Peak Amplitude (% EMG max)		EMG Burst Duration (sec)	
	Rectus Femoris	Biceps Femoris	Rectus Femoris	Biceps Femoris	Rectus Femoris	Biceps Femoris
Young	**0.28 ± 0.25**	**0.34 ± 0.15**	**32.80 ± 9.09**	**7.92 ± 2.29**	**1.41 ± 0.35**	**0.96 ± 0.22**
Old	**0.59 ± 0.14**	**0.58 ± 0.15**	**33.82 ± 5.89**	**8.01 ± 2.79**	**1.08 ± 0.21***	**0.94 ± 0.24**
Pooled	**0.44 ± 0.10**	**0.44 ± 0.12**	**33.31 ± 7.47 ǂ**	**7.96 ± 2.49**	**1.24 ± 0.33**	**0.95 ± 0.22**

Data represent the mean ± standard deviation. EMG, electromyography; Young, younger adults; Old, older adults.
* Significantly different from younger adults ($p < 0.05$).
ǂ Significantly different from the biceps femoris ($p < 0.01$).

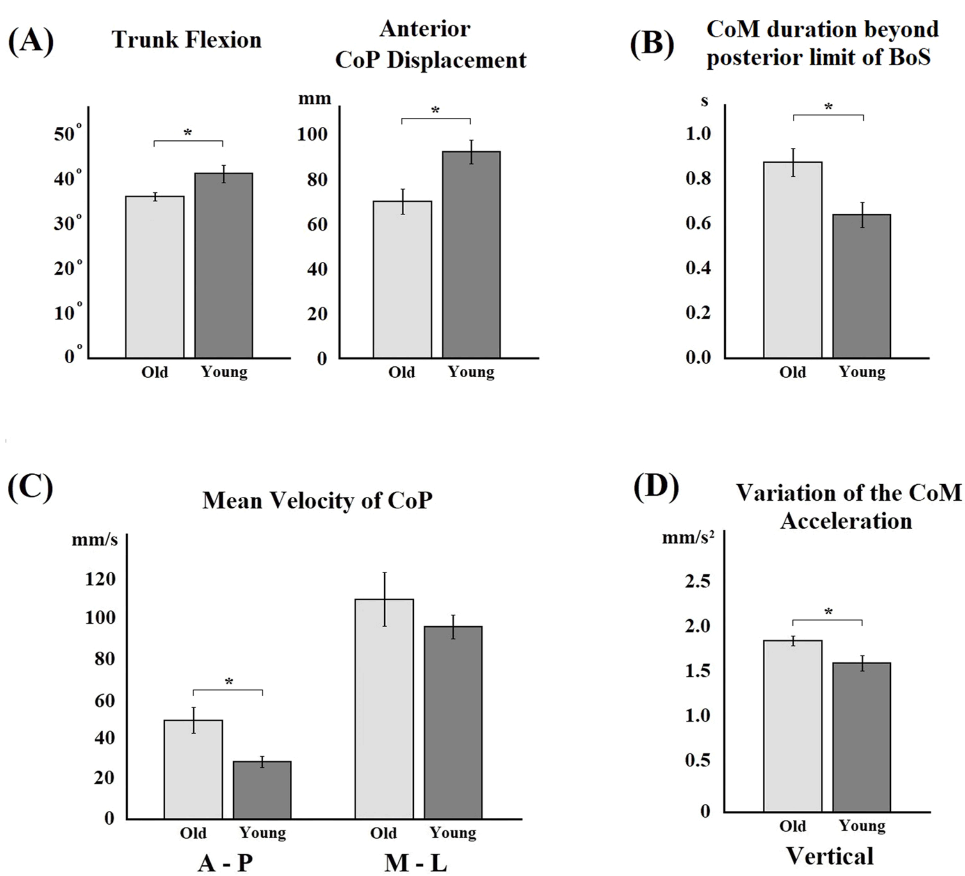

Fig. 2. Age-related differences in trunk flexion angle and stability measures during StandTS. Mean ± SE of (A) trunk flexion angle and anterior CoP displacement, (B) CoM duration beyond posterior limit of BoS, (C) velocity of CoP in the anterior-posterior (A-P) and medio-lateral (M-L) directions, and (D) variation of the CoM acceleration in the vertical direction in both younger and older adults. * $p < 0.05$, CoP: center of pressure, CoM: whole body center of mass. BoS: base of support s: second.

with a reduced time in which the CoM remained inside the BoS during StandTS (Fig. 3).

The EMG burst duration of the eccentric RF muscle contraction was correlated with duration in which the CoM remained within the BoS ($r_s(18) = 0.65$, $p < 0.01$) (Fig. 4).

4. Discussion

The purpose of this study was to investigate differences in trunk kinematics and lower extremity muscle activity and measures of postural stability between older and younger adults and determine the relationship between trunk flexion angle and eccentric RF muscle activation with postural stability during a StandTS task. We found that older adults had smaller trunk flexion angles with shorter RF EMG burst duration, and greater A-P and vertical postural sway compared to younger adults. Additionally, the reduced trunk flexion and shorter RF burst duration correlated with a shorter duration in which the CoM remained within the BoS during a StandTS.

4.1. Relationship between trunk flexion and StandTS stability

The point at which the CoM crossed the posterior limit of the BoS during StandTS occurred earlier in older adults compared to younger adults. As a result, the duration in which the CoM remained beyond the BoS was longer in older adults, thereby further supporting a prolonged state of instability in older adults.

To maintain postural stability during StandTS, the sagittal extrapolated position of the CoM should be regulated within the BoS [25]. Considering that the angular velocity of trunk flexion is not different between age groups, one reason for the earlier crossing of the CoM over the posterior limit of the BoS is the reduced trunk flexion angle in older adults during StandTS compared to younger adults. Indeed, the positive correlation between trunk flexion and duration in which CoM stayed in the BoS during StandTS supports this suggestion. We found that younger adults demonstrated greater anterior CoP displacement and generated greater vertical GRF during the descending phase of StandTS. Thus, greater trunk flexion led to increased weight-bearing on the lower extremities. The increased trunk flexion also led to increased stability in younger adults. These findings are supported by others, who have reported postural instability during StandTS as a result of reduced lumbar spine and trunk flexion in individuals with chronic non-specific low back pain [26]. Greater trunk flexion moves the CoM towards the anterior border of the BoS at the start of StandTS and this enables the CoM to retain its position longer in the BoS area throughout the StandTS, which appears to have contributed, at least in part, to the improved A-P balance in younger adults compared to older adults.

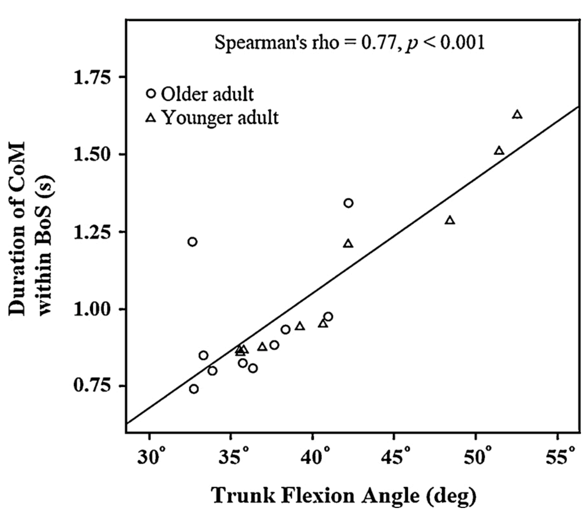

Fig. 3. Correlation between maximum trunk flexion angle and duration in which center of mass remained within the BoS during StandTS across younger and older adults. CoM: whole body center of mass. s: second.

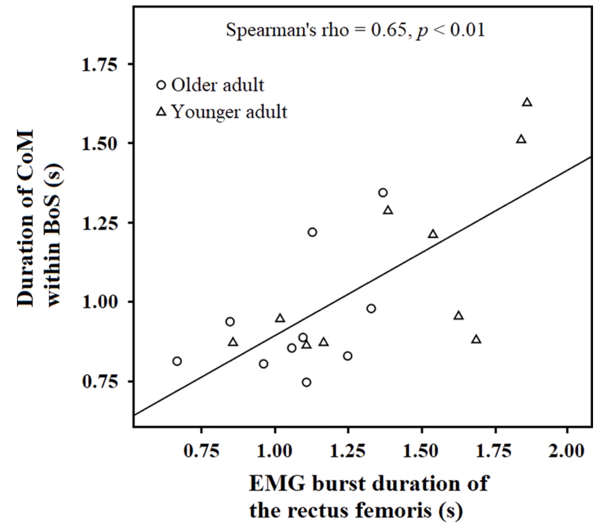

Fig. 4. Correlation between EMG burst duration of the rectus femoris and duration in which center of mass remained within the BoS during StandTS across younger and older adults. CoM: whole body center of mass. s: second.

4.2. RF and BF muscles activation and StandTS stability

Although there was no difference in BF activity between age groups, we observed that the peak EMG of the RF was preceded by the first peak EMG of the BF across age groups. The activation of the BF increased quickly around the maximum angular velocity of the trunk flexion before reaching the maximum trunk flexion. Given that the long-head of the BF is a bi-articular muscle, the initial activity of the BF muscle was likely due to an eccentric lengthening triggered by active trunk flexion. Eccentric contraction of the BF, therefore, contributed to controlling the forward velocity of the CoM during trunk flexion, which served to stabilize the pelvis and trunk during StandTS initiation [17,27,28]. Following the BF peak onset, RF activation onset occurred at the instant when the increasing downward velocity of the CoM converted to decreased velocity.

Decreasing downward velocity of the CoM is mainly due to eccentric contraction of the quadriceps which provides knee extension torque to resist body weight loading for a stable and safe landing [8]. We hypothesized that older adults would demonstrate decreased EMG peak amplitude and burst duration of the RF. Despite a lack of age-related differences in the peak amplitude of the eccentric contraction of the RF, older adults showed a shorter EMG burst duration of the RF compared to younger adults. This resulted in a longer time interval between the peak timing of the RF and StandTS termination in older adults. In contrast, peak timing of the RF in younger adults was relatively close to the StandTS termination.

Eccentric contraction of the quadriceps controls downward displacement of the CoM by decelerating the CoM in the vertical direction [29] and provides postural stability during unexpected perturbation responding instantly to stretch imposed by loads [30]. Thus, reduced eccentric RF burst duration in older adults leads to a lack of descending control and, in combination with the decreased trunk flexion mentioned above, reduced postural stability during StandTS. Previous reports of age-related decreases in eccentric control and/or strength of the para-spinal musculature, decreased eccentric control and/or strength of the quadriceps musculature, and increased stiffness of the lumbar spine with perception of instability provide supportive evidence for our findings [11,31,32].

Given that older adults showed smaller vertical GRF compared to younger adults, short anterior displacement of the CoM by reduced trunk flexion may result in relatively low weight bearing on the lower limb at the beginning of StandTS and this may lead to smaller eccentric contraction of the quadriceps in older adults. It is thus plausible that a decrease in trunk flexion in older adults resulted in short duration of the RF activation and this might have contributed to reduced postural control. Indeed, our findings demonstrated that short RF EMG burst duration correlated with a reduced time in which the CoM remained inside the BoS during a StandTS task. Furthermore, findings of greater mean velocity of CoP in the A-P and vertical variation of CoMAccel in older compared to younger adults supports a state of postural instability. Given that StandTS velocity was not different between age groups, a shorter activation of the eccentric contraction of the RF in older adults appears to be related to an increased body oscillation in the A-P and vertical directions.

4.3. Clinical significance

Although there are no StandTS clinical tests per se, the STS test is often used as a clinical assessment of functional lower extremity strength in older adults. Tests are typically timed and can include the number of STS completions over 30 s or the time it takes to complete five STS repetitions. However, the focus on speed is at the expense of controlled stability during the StandTS phase that is needed to achieve an accurate and safe landing on the chair. Our findings demonstrated that eccentric lower extremity strength and trunk flexion control are important factors to improve stability throughout a StandTS task and prevent high impact forces during seat contact that would otherwise lead to increased impact to the spine, and/or skin damage, and falls [33,34]. Thus, consideration of controlled StandTS transition should be included in STS functional tests in older adults or as a separate test altogether in which stability is emphasized over speed. In addition, eccentric resistance training with kinematic cues regarding trunk flexion could be added to functional mobility and balance training protocols in order to improve dynamic balance during StandTS in older adults.

4.4. Limitations

There are a few study limitations that are worth mentioning. The observed difference in the RF muscle activation pattern may not be the only factor that affects postural stability during a StandTS. The vastus lateralis and medialis are also important muscles for the eccentric contraction of the quadriceps during this task. However, the quadriceps muscle group in healthy adults work together such that they demonstrate similar activation patterns during eccentric contractions [35]. Future studies should investigate whether there are age-related changes in the contribution of these individual muscles during StandTS. In addition, although lower back and abdominal muscles control the core stability [36], they were not measured in this study. Finally, given that the experimental protocol required a controlled arm position, future research should evaluate the influence of arm assistance on trunk kinematics and lower extremity muscle activity in relation to balance control.

5. Conclusions

Reduced trunk flexion and shorter duration of RF eccentric contraction during StandTS was associated with instability in older adults as defined by reduced duration in which CoM is maintained within BoS. These findings support a need to develop StandTS rehabilitation assessments and interventions focused on improving trunk flexion and maintaining eccentric RF control as means to improve postural stability in older adults in this common task of daily living.

Declaration of Competing Interest

The authors have no conflict of interest.

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