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Original Article

Association between ankle muscle strength and postural sway in older adults

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Abstract

Our study aimed to investigate the relationship between the strength of ankle muscles and postural sway in older adults during upright standing tasks of varying difficulty. Forty-three elderly participants (20 women) performed maximal isometric contractions and three distinct postural tasks: normal quiet stance (60 s), tandem stance (20 s), and one-legged stance (10 s) on a pressure platform. Postural sway was comprehensively assessed, considering center of pressure velocity, peak-to-peak amplitude, and standard deviation of center of pressure displacement in both anterior/posterior and medio/lateral directions. Additionally, weight distribution asymmetry in both directions served as a quantifiable index. Surface electromyographic (EMG) activity of the Tibialis Anterior, Medial Gastrocnemius, Rectus Femoris, and Semitendinosus was recorded during the postural tasks. Throughout the balance tasks, isometric plantar and dorsiflexion torque exhibited a consistent decrease as posture variables increased ($p < 0.001$). Narrowing the base of support led to significant increases in all postural variables and EMG activity ($p < 0.01$), with a more pronounced effect on ankle muscles compared to hip muscles ($p < 0.0005$). In the normal quiet stance ($r = .683$) and tandem stance ($r = .641$), greater center of pressure velocity correlated with a more symmetrical weight distribution between the legs. These results suggest that ankle muscle strength serves as a reliable predictor of static balance control, particularly in postures involving a narrow stance. Therefore, the enhancement of plantar flexors and dorsiflexors, rather than an exclusive focus on symmetrical weight distribution, is more likely to contribute to improved static balance control in older adults.

Keywords: aging; posture; strength; plantar flexors; dorsiflexors; EMG.

Introduction

When we stand upright, our bodies are not stationary but sway back and forth and side to side although we are usually not aware of these displacements. This so-called postural sway is often characterized by measuring the shifts in the center of pressure under the feet during tests of standing balance. When the body's center of mass moves in either the forward-backward or side-to-side directions, leg muscles are activated to keep the center of pressure within the base of support (Héroux et al., 2014; Toth et al., 2021). When standing in an upright relaxed position, the muscles involved in maintaining balance are largely those that span the ankle joint, although muscles that act across the knee and hip joints can become involved when standing in more challenging balance conditions (Amiridis et al., 2003).

Older adults exhibit greater and more variable center-of-pressure (CoP) displacements and associated electromyographic (EMG) activity in leg muscles during short-duration postural tasks than young active adults (Amiridis et al., 2003; Watanabe et al., 2018). Moreover, both the differences between age groups (Amiridis et al., 2003; Tsabalaki et al., 2023) and the amount of EMG activity (Sozzi et al., 2013) increases with the difficulty of the postural task. However, 4-weeks of training the tibialis anterior muscle by superimposing electrical stimulation on voluntary isometric contractions increases maximal voluntary contraction (MVC) torque of the dorsiflexors and reduces postural sway in older adults (Amiridis et al., 2005). Although, it is not clear whether it was the training intervention or muscle strength *per se* that influenced the quality of balance control (Billot et al., 2010; Horlings et al., 2008).

The relative amplitude of the center-of-pressure (CoP) displacements in the forward-backward (F/B) and side-to-side (S/S) directions depends on the size and orientation of the base of support. When standing upright in a relaxed position, for example, the CoP displacements are greater in the F/B direction than the S/S direction in

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both young and older adults (Amiridis et al., 2003). When standing on one foot, however, there is an increase in the CoP displacements in the S/S direction relative to normal quiet standing (NQS) (Amiridis et al., 2003; Watanabe et al., 2018) and they are much greater when the eyes are closed than when they are open (Sozzi et al., 2013). Moreover, the proportion of body weight supported by each foot varies across postural conditions (Jonsson et al., 2007).

The purpose of our study was to examine the association between the strength of the dorsi- and plantar flexor muscles and postural sway in the forward-backward and side-to-side directions when older adults stood upright during tasks of varying difficulty. The relative roles of the dorsiflexor and plantar flexor muscles in the control of postural sway appears to differ during upright standing (Di Giulio et al., 2009), but it is the net torque capacity of the ankle muscles that is negatively correlated with CoP displacement across postural conditions (Billot et al., 2010; Cattagni et al., 2014). We hypothesized that the maximal voluntary contraction (MVC) torque generated by the plantar flexor muscle would exhibit the strongest correlation with postural sway across conditions.

Methods

Participants

A total of 43 elderly adults (20 women) volunteered to participate in this study (age: 70 ± 4 years; height: 167 ± 10 cm; mass: 74 ± 6 kg). Thirty-eight of the 43 participants reported right-leg dominance and the agreement between self-reported and observed leg dominance for mobilizing and stabilizing tasks was 100% (van Melick et al., 2017). All participants were physically active with no chronic pain in the lower extremities (e.g., chronic ankle instability) and no surgery on the lower extremities in the last 12 months. The exclusion criteria were acute and chronic infections, neurological disorders, visual and vestibular impairments, and limited walking ability.

Prior to being enrolled in the study, participants were informed of the benefits and risks of the investigation and signed an institutionally approved informed-consent document. The standardized Mini Mental Health Examination was administered to confirm language competency and adequate cognition, with the requirement that all subjects score at least 25/30. Approval for the experiment was obtained from University Ethics Committee on Human Research in accordance with the Declaration of Helsinki.

Study Design

To investigate the correlation of ankle strength with postural control, a group of 23 elderly men and 20 elderly women was asked to perform maximal isometric contractions with the plantar flexors and dorsiflexors in an isokinetic dynamometer and three postural tasks of increasing difficulty: normal quiet stance (NQS), tandem stance (TS), and one-leg stance (OLS). After being familiarized with the postural tasks and strength measurements, MVC torque was measured at ankle angles of neutral (0^o) and $\pm 10^o$. Static balance was evaluated by calculating several CoP parameters in each of the three stance conditions: CoP velocity, total CoP path, sway area, sway ellipse, peak-to-peak amplitude, and standard deviation of CoP displacement in the forward-backward and side-to-side direction (Amiridis et al., 2003). The activation of ankle and knee muscles during the postural tasks was also measured using electromyography (EMG).

Procedures

Strength measurements

The torque capacity of the dorsiflexors and plantar flexors during isometric contractions was evaluated with an isokinetic dynamometer (Humac Norm, CSMI, MA). Participants were familiarized with the apparatus and the standardized testing procedure during three 30-min sessions in 1 week. Special attention was given to the stabilization of the participants with Velcro straps, isolation of the muscle group being tested, and alignment of the limb with the machine axes of rotation. Participants were placed in a seated position with the back reclined at approximately 110^o, the knee fixed to 0° (full extension), and the arms placed across the chest and the hands grasping the straps. After a standardized warm-up, they performed three maximal contractions (MVCs) with the dorsiflexors and plantar flexors in a random order. The duration of each MVC was 5 s, with a 2-3 min rest between trials. Visual feedback of the applied torque was provided on the computer screen. The trial in which the peak torque was achieved from the three trials at each ankle angle was selected for further analysis.

Balance measurements

All balance tasks were performed on a pressure platform (Comex, LorAn engineering, Bologna, IT). This pressure platform (70 x 50 cm) comprises 2304 quartz piezoelectric sensors to record pressure distribution at 50 Hz. Static balance was assessed during three tasks:

1. Normal Quiet Stance. Participants were asked to stand on the pressure platform in a natural position with approximately 10-15 cm between the malleoli for 60 s. They were instructed to stand as still as possible with the arms freely hanging along the sides of the body.
2. Tandem Stance. The instruction was to stand on the platform with the heel of the non-dominant foot in front of the toes of the dominant foot leaving no space between the feet. The arms were placed on the hips with the elbows flexed at 100^o approximately and the position was maintained for 20 s.

3. One-Legged Stance. Participants stood on the platform with one foot pointing straight ahead in a sagittal plane. The other leg was raised above the ground with the hip flexed to ~45º (0º = hip angle at full standing position) and the knee at ~45º (0º = full extension). The position was maintained for 10 s.

Data recording started once the participant was stationary in the required posture. Participants were instructed to look straight ahead at a marker (3 cm diameter) positioned at eye level at 1.5 m in front of them. They performed three trials of each task with 3 min of rest between trials. The balance tasks were performed at the same time of the day and performance order was counterbalanced across tasks.

The Footchecker software (Loran Engineering, IT) automatically calculated all the variables in each task: the average CoP velocity (CoP_{vel}, mm/s), peak-to-peak amplitude (CoP_{max}, mm), standard deviation (CoP_{sd}, mm) of the CoP displacement, and relative body-weight distribution in the forward-backward and side-to-side directions. An asymmetry index of 0.47 in the forward-backward direction indicated that 47% of the body weight was supported by the front foot and 53% by the rear foot. Similarly, an index of 0.47 in the side-to-side direction indicated that 47% of the body weight was supported by the left leg and 53% by the right leg. The asymmetry index in the side-to-side direction was reversed for the five left-dominant participants. The initial and final 2 s of each trial were discarded from the analysis. The middle 56 s, 16 s, and 6 s of the NQS, TS, and OLS recordings, respectively, were analyzed.

Electromyography

EMG signals were recorded with a TEL100D (Biopac Systems, Goleta, CA) system comprising shielded electrode-lead assemblies (bipolar silver/silver chloride electrodes) that were interfaced with a portable amplifier (CMRR > 110 db at 50/60 Hz, bandwidth = 10–500 Hz). Recordings were obtained for tibialis anterior (TA), soleus (SOL), rectus femoris (RF), and semitendinosus (ST) muscles during postural tasks. Low impedance ($Z < 1$ kΩ) at the skin-electrode interface was obtained by shaving, abrading the skin, and cleaning it with alcohol.

Electrodes were placed according to SENIAM recommendations (Hermens et al., 2000): those for TA were placed at 1/3 along the line from the tip of the fibula and the tip of the medial malleolus, those for SOL were placed over the lateral-posterior compartment, those for RF were placed at 50% along the line from the anterior superior iliac spine to the superior patella, and those for ST were placed at 50% along the line between the ischial tuberosity and the medial epicondyle of the tibia. The EMG signals were band-pass filtered (cut-off: 10-330 Hz), sampled at 1000 Hz, and low-pass filtered with a 4th-order, zero-lag Butterworth that had a cut-off of 100 Hz. The signal was then full-wave rectified and the root mean square (RMS) values were calculated.

Statistical Analysis

MVC torque was compared with a 2 (action: plantar flexion and dorsiflexion) × 3 (angle: -10°, 0° and 10°) repeated-measures ANOVA. The balance measurements were examined with a one-way ANOVA to compare the influence of the balance task on the average CoP velocity (CoP_{vel}), total CoP path, sway area, and sway ellipse. Furthermore, a 2 (direction: Anterior/Posterior and Medio/Lateral) × 3 (balance task: NQS, RS and OLS) repeated-measures ANOVA was used to investigate the influence of the direction and the task on the CoP_{max} and CoP_{sd}.

We also investigated the association between balance and strength measures using multiple correlation and linear regression analyses. For both muscle groups the neutral (0°) angle was chosen for the correlation analysis because it represents the anatomical position of the foot during the postural tasks. MVC torque for the dorsiflexors and plantar flexors was included in a simple bivariate correlation (Pearson product moment) with the postural stability measures for the three balance tasks. We then developed a multiple linear regression model to identify those variables that could predict postural stability. A correlation analysis was also performed between interlimb asymmetry of weight distribution and average CoP_{vel} during NQS, as well as between weight distribution on front/rear foot and CoP_{vel} during TS.

Moreover, a 4 (muscles: TA, SOL, RF and ST) × 3 (balance tasks: NQS, TS and OLS) ANOVA was used to investigate the influence of muscle group and the task on CoP variables. Significant interactions were analyzed employing a post hoc Tukey test. Level of significance was set at $P < 0.05$ for all tests.

Results

Strength measurements

Figure 1 presents the torque–angle relation for the plantar flexors and dorsiflexors. The intraclass correlation coefficient (ICC) was 0.927. The two-way repeated measures ANOVA revealed significant main effects of muscle group ($F_{(1,42)} = 188.871$, $p < 0.0005$, $\eta p^2 = 0.818$) and angle ($F_{(2,84)} = 4.756$, $p = 0.011$, $\eta p^2 = 0.102$) on MVC torque. Also, there was a significant group × angle interaction ($F_{(2,84)} = 97.260$, $p < 0.0005$, $\eta p^2 = 0.698$), which indicated that MVC torque was greatest for each muscle group at its longest length (-10º for the plantar flexors and 10º for the dorsiflexors; Figure 1).

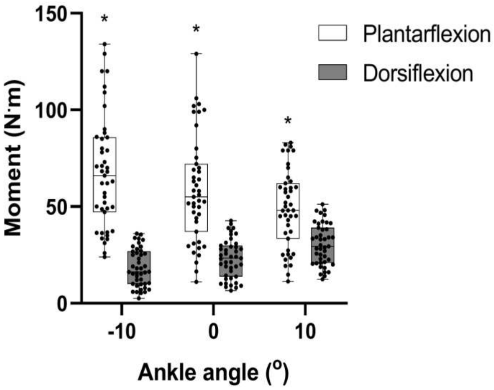

Figure 1. Maximal isometric plantar flexion (black) and dorsiflexion (grey) torque (N.m), at three ankle angles (-10º, 0º, and 10º) for all participants (n = 43). Data are shown as box graphs with "whiskers" displaying the 5-number summary of the data: the maximum limit (upper whisker), the 3rd quartile (75th percentile), the median value (line inside the box), the first quartile (25th percentile) and the minimum limit (lower whisker) *: significantly greater than dorsiflexion at the same angle, $p < 0.0005$.

Balance measurements

Center of Pressure (CoP)

Figure 2 presents the CoP displacement for a representative participant during the three postural tasks. There was a significant difference for average CoP_{vel} (ICC = 0.684) across the three postural tasks ($F_{(2,126)} = 175.271$, $p < 0.0005$).

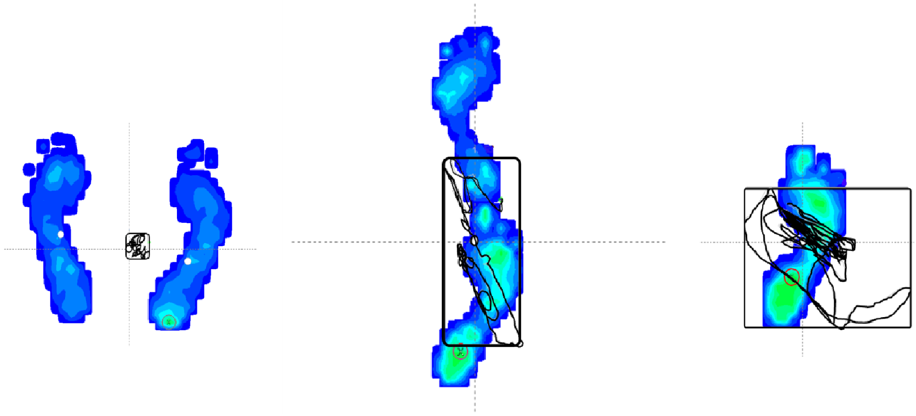

Figure 2: Center of Pressure (CoP) displacement (black line) for a representative participant during Normal Quiet Stance (NQS) on the left, Tandem Stance (TS) in the middle, and One Leg Stance (OLS) on the right.

Tukey post hoc tests revealed that average CoP_{vel} was significantly ($p < 0.0005$) greater during OLS (11.3 ±3.8 mm/s) relative to TS (8.3 ±1.9 mm/s) and the TS value was greater ($p < 0.0005$) compared with NQS (1.5 ±0.60 mm/s) (Table 1). A two-way, repeated-measures ANOVA (direction × task) found a marginally significant main effect of direction on CoP_{max} (ICC = 0.860, $F_{(1,42)} = 4.087$, $p = 0.05$, $\eta p^2 = 0.089$). Also, there was a significant main effect of task ($F_{(2,84)} = 391.762$, $p < 0.0005$, $\eta p^2 = 0.903$) and a significant direction × task interaction ($F_{(2,84)} = 12.133$, $p < 0.0005$, $\eta p^2 = 0.224$) on CoP_{max}. The maximal displacement in the forward-backward direction (17.7 ± 3.8 mm) during the OLS task was greater than the side-to-side direction (15.3 ± 3.7 mm) (Table 1).

The analysis found no significant influence of direction (ICC = 0.878, $F_{(1,42)} = 0.443$, $p = 0.509$, $\eta p^2 = 0.010$) on CoP_{sd}, but a significant main effect of task ($F_{(2,84)} = 87.702$, $p < 0.0005$, $\eta p^2 = 0.751$) and a significant direction × task interaction ($F_{(2,84)} = 18.502$, $p < 0.0005$, $\eta p^2 = 0.306$) on CoP_{sd}. The variance of the CoP displacement (CoP_{sd}) in the side-to-side direction (3.5 ±1.5 mm) during the OLS task was greater than that in the forward-backward direction (2.9 ±1.2 mm) (Table 1).

Table 1. Postural variables during normal quiet stance (NQS), tandem stance (TS), and one-leg stance (OLS) for all participants (n=43). Vales are mean ± SD. * P<0.05 relative NQS; ** P<0.05 relative to TS; † P<0.05 relative to the same variable in the other direction. Values are mean (± SD). F/B = forward-backward, S/S = side-to-side.

	NQS	TS	OLS
CoP_{vel} (mm/s)	1.50 ± 0.59	8.32 ± 1.92*	11.26 ± 3.8**
CoP_{max} F/B (mm)	6.2 ± 1.9	12.2 ± 2.7*	17.6 ± 3.8** †
CoP_{max} S/S (mm)	6.7 ± 2.9	11.8 ± 4.5*	15.3 ± 3.7**
CoP_{sd} F/B (mm)	1.33 ± 0.60	2.92 ± 0.99*	2.89 ± 1.15*
CoP_{sd} S/S (mm)	1.22 ± 0.74	2.55 ± 1.10*	3.54 ± 1.48** †

Electromyography

A 4 (muscles) × 3 (postural tasks) ANOVA revealed significant main effects of muscle (ICC = 0.891, $F_{(3,126)} = 38.215$, $p < 0.0005$, $\eta p^2 = 0.476$) and task ($F_{(2,84)} = 55.439$, $p < 0.0005$, $\eta p^2 = 0.569$) on EMG_{RMS}, indicating that the quantity of muscle activation increased as a function of task difficulty (Figure 3).

There was a significant muscle × task interaction ($F_{(6,252)} = 27.207$, $p < 0.0005$, $\eta p^2 = 0.393$) indicating that the increase in EMG_{RMS} with task difficulty varied across muscles. Particularly, the increase in EMG_{RMS} for TA and SOL was significantly greater than that for RF and ST (Figure 3). Post hoc Tukey test revealed that the EMG_{RMS} for TA (37 ± 23 au) and SOL (28 ± 15 au) was significantly ($p < 0.0005$) greater than that for RF (15 ± 7 au) and ST (15 ± 8 au) during the TS task. Similarly, EMG_{RMS} for TA (57 ± 41 au) and SOL (49 ± 36 au) was significantly ($p < 0.0005$) greater than that for RF (16 ± 7 au) and ST (19 ±12 au) during the OLS task (Figure 3).

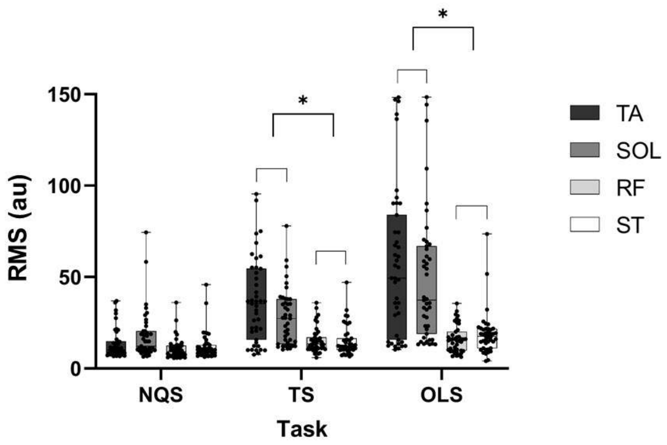

Figure 3: Integrated EMG activity of Tibialis Anterior (TA), Soleus (SOL), Rectus Femoris (RF), and Semitendinosus (ST) muscles during the 3 postural tasks: Normal Quiet Stance (1), Tandem Stance (2) and One-

Leg Stance (3). Data are shown as box graphs with "whiskers" displaying the 5-number summary of the data: the maximum limit (upper whisker), the 3rd quartile (75th percentile), the median value (line inside the box), the first quartile (25th percentile) and the minimum limit (lower whisker) *: significantly different from RF and ST muscle, $p < 0.0005$.

Correlation and regression analysis between strength and balance measurements

The MVC torques for the plantar flexors (PF_0) and dorsiflexors (DF_0) were significantly correlated with all postural variables during the three tasks (Table 2). The Pearson correlation coefficient between PF_0 and all postural variables ranged from -0.636 to -0.835 and that between DF_0 and all postural variables varied from -0.634 to -0.758 (Table 2). A negative correlation indicates that each CoP variable was smaller for stronger individuals.

Table 2. Pearson correlation coefficients between MVC torque for the plantar flexors (PF) and dorsiflexors (DF) at the neutral angle and the postural variables derived from the center of pressure during the three postural tasks. NQS: Normal Quiet Stance, ST: Stance Tandem, OLS: One Legged Stance. CoP_{vel}: mean velocity of CoP displacement; CoP_{max} A/P: peak-to-peak amplitude in the anterior-posterior direction; CoP_{max} M/L: peak-to-peak amplitude of the CoP displacement in the medio-lateral direction; CoP_{sd} F/B: standard deviation of the CoP displacement forward and backward; CoP_{sd} S/S: standard deviation of the CoP displacement side-to-side. All correlations were significant at $p < 0.05$.

		Plantar flexion	Dorsiflexion
CoP_{vel}	NQS	-0.69	-0.72
	ST	-0.76	-0.71
	OLS	-0.84	-0.76
CoP_{max} F/B	NQS	-0.65	-0.63
	ST	-0.70	-0.72
	OLS	-0.77	-0.70
CoP_{max} S/S	NQS	-0.64	-0.65
	ST	-0.69	-0.64
	OLS	-0.71	-0.73
CoP_{sd} F/B	NQS	-0.70	-0.64
	ST	-0.75	-0.63
	OLS	-0.72	-0.71
CoP_{sd} S/S	NQS	-066	-0.65
	ST	-0.67	-0.69
	OLS	-0.73	-0.70

Multiple linear regression models were developed to determine the relative influence of MVC torque of the plantar flexors and dorsiflexors on the variance in the postural variables (Table 3). As the models for all postural variables were similar, the associations for CoP_{vel} are used as an example of the findings. The regression model ($CoP_{vel} = 2.510 – 0.027DF – 0.007PF$) indicated that 55.5% ($R^2 = 0.555$) of the variance in CoP_{vel} during NQS could be explained by the MVC torques ($F_{(2, 40)} = 24.915$, $p < 0.0005$). The dorsiflexor (DF) torque was a unique predictor of CoP_{vel} (partial r = -0.72), whereas plantar flexor (PF) torque was not (partial r = -0.25). In contrast, the model ($CoP_{vel} = 11.711 – 0.054DF – 0.037PF$) explained 59.9% of the variance in CoP_{vel} ($R^2 = 0.599$) during TS ($F_{(2,40)} = 29.890$, $p < 0.0005$) with plantar flexor torque as a significant predictor (p = 0.005) of CoP_{vel}, but not dorsiflexor torque (p = 0.115). Similarly, the model explained 71.3% ($R^2 = 0.713$) of the variance in CoP_{vel} during OLS ($F_{(2,40)} = 49.773$, $p < 0.0005$; $CoP_{vel} = 18.466 – 0.084DF – 0.090PF$) with plantar flexor torque as a significant predictor ($p < 0.0005$) of CoP_{vel}, but not dorsiflexor torque (p = 0.139) (Table 3). However, the CoP position variables exhibit large differences between them and also in comparison to CoP_{vel}. For example, during NQS, the CoP_{sd} in both directions F/B and S/S could explained by the MVC torques (50% and 47%), however, the PF torque was a unique predictor of CoP_{sd} (partial r = -0.70 and -0.66), whereas the DF torque was not. On the contrary, during TS, the DF torque was a unique predictor, but the PF torque (Table 3).

Table 3. F values and coefficients of determination (R^2) for the multiple linear regression models in which the MVC torques for the dorsiflexors and plantar flexors were used to predict the CoP variables during normal quiet stance (NQS), tandem stance (TS), and one-legged stance (OLS). All F values were significant at $p < 0.0005$.

	NQS			TS			OLS		
	R^2	PF	DF	R^2	PF	DF	R^2	PF	DF
Total CoPvel	0.555		-0.72*	0.599	-0.76*		0.713	-0.83*	
CoP_{max} F/B	0.450	-0.64*		0.556		$-0.36^{\#}$	0.604	-0.76*	
CoP_{max} S/S	0.451		-0.65*	0.492	-0.69*		0.568		$-0.37^{\#}$
CoP_{sd} F/B	0.500	-0.70*		0.561		-0.75*	0.560	$-0.33^{\#}$	$-0.30^{\#}$
CoP_{sd} S/S	0.472	-0.66*		0.515		-0.69*	0.560	$-0.37^{\#}$	

* $p < 0.001$, # $p < 0.05$

Correlation between weight distribution and CoP velocity

The weight-distribution asymmetry index and the CoP_{vel} were positively correlated during NQS (r =0.683, $p < 0.0005$), indicating that CoP_{vel} was greater when the weight was distributed more evenly between the two feet (Figure 4, left). Moreover, the weight distribution in the forward-backward direction and the CoP_{vel} were positively correlated during the TS task, (r = 0.641, $p < 0.0005$), indicating that CoP_{vel} was greater when more weight was placed on the front foot (Figure 4, right).

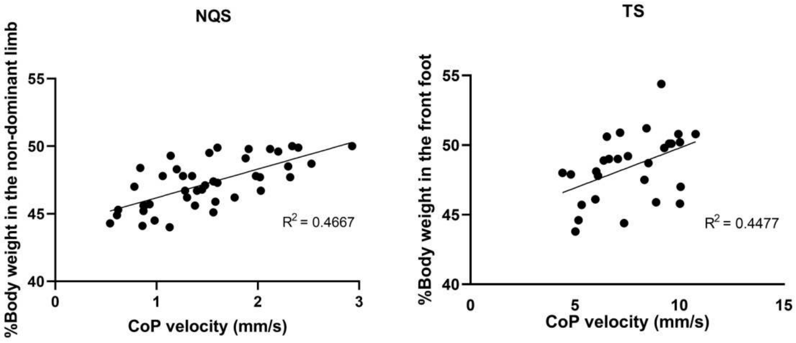

Figure 4: Correlation plots between weight distribution and average velocity of the Center of Pressure (CoP_{vel}). a) % weight on the non-dominant limb and CoP_{vel} during Normal Quiet Stance (NQS); the higher the symmetry, the greater the CoP_{vel} ($R^2 = 0.4667$) b) % weight in the front foot and CoP_{vel} during the Tandem Stance (TS); the higher weight is placed on the front foot, the greater the CoP_{vel} ($R^2 = 0.4177$).

Discussion

The main findings of our study were the strong correlations between the strength of the plantar flexor and dorsiflexor muscles with the variance in all the metrics used to characterize postural sway in older adults. EMG recordings indicated that the amount of activation observed for the ankle muscles (TA and SOL) was modulated more across postural conditions than was that for the thigh muscles (RF and ST). A regression analysis found that MVC torque of the dorsiflexor muscles was the more significant explanatory variable for the CoP velocity during the NQS condition, whereas it was the correlation with MVC torque for the plantar flexors that was significant during both the TS and OLS conditions. Moreover, CoP velocity was influenced by the symmetry of the weight distribution across the two feet.

Strength and balance control during upright standing

Healthy aging is accompanied by declines in postural control and walking ability. The broad range of factors (vestibular, proprioceptive, visual) that can influence the control of static and dynamic balance challenges our understanding of the adaptations responsible for reductions in postural control. Despite the multi-factorial nature of these adaptations, our findings underscore a critical role for decreases in the strength of ankle muscles. We found a strong negative correlation between the strength (MVC torque) of the dorsiflexor muscles and postural sway during NQS, but between plantar flexor MVC and postural sway during TS and OLS.

These results are consistent with previous studies that also reported a strong correlation between MVC torque of the dorsiflexors and plantar flexors during isometric contractions and CoP displacement in older adults who were categorized as fallers and non-fallers (Cattagni et al., 2014). Moreover, Billot et al. (2010) reported that the MVC torques of the ankle muscles was negatively correlated with CoP displacement when performing a number of different postural tasks. Importantly, they observed that the strength of the correlation between MVC torque and the length of the CoP path increased with the difficulty of the postural task being least in NQS ($R^2 = 0.48$), greatest in OLS ($R^2 = 0.91$), and intermediate in TS ($R^2 = 0.62$). In our study, the strength of the correlations between the CoP measures and the MVC torques was weaker, albeit statistically significant for all comparisons, and did not differ systematically across postural tasks. Likely explanations for the differences between the two studies were that the participants in Billot et al. (2010) were older (average age: 80 yrs vs 70 yrs). Nonetheless, these findings indicate that declines in the strength of the ankle muscles have a significant negative influence on the amount of postural sway exhibited by older adults, especially during conditions that involve reductions in the base of support.

Dorsiflexors vs plantar flexors

In NQS, the body's center of mass is located forward of the point of application of the center of pressure, which imposes a forward toppling moment about the base of support (Héroux et al., 2014). This destabilizing torque is counteracted by the mechanical properties of the plantar flexor muscles (Di Giulio et al., 2009). Indeed, it seems that the passive stiffness of the plantar flexors is able to oppose the modest displacements that occur in postural sway during NQS (Hasson et al., 2014; Loram et al., 2007), even though the plantar-flexor muscles exhibit low levels of intermittent EMG activity during this condition (Amiridis et al., 2005; Cohen et al., 2020; Di Giulio et al., 2009). Nonetheless, the results of our study indicate that it is the strength of the dorsiflexor muscles that is more critical in constraining the postural sway of older adults during normal upright standing.

In contrast, it is the strength of the plantar flexor muscles that is more important in accommodating challenges to balance in less stable conditions, especially for older adults. For example, MVC torque of the plantar flexors is able to distinguish between fallers and non-fallers (Cattagni et al., 2014). Similarly, the soleus muscle is continuously active during OLS, whereas tibialis anterior and peroneus longus exhibit alternating EMG activity that controls postural sway in the side-to-side direction (Sozzi et al., 2013). In a previous study, we reported a similar increase in EMG amplitude for the ankle and hip muscles with increasing task difficulty in 19 older adults (Amiridis et al., 2003). The normalized EMG amplitude for both the ankle and hip muscles increased by 1.5x during the TS task and 1.7x during the OLS task. In the current study, we observed an average increase on EMG amplitude of 2.5x during the TS task and 4x during the OLS task. The larger increases in the current study are likely attributable to differences in the durations of the tasks: 5 s for all tasks in the previous study compared with 60 s (NQS), 20 s (TS), and 10 s (OLS) in the current study. Also, only men were enrolled in the previous study, whereas in the current study 20/43 participants were women and their increases in EMG amplitude for the two tasks (3x and 5x, respectively) were greater than those for the men (2x and 3x, respectively).

Postural tasks and asymmetries in weight distribution

It seems that the distribution of body weight under the feet is not always symmetrical during upright standing (Day & MacNeilage, 1996; Hesse et al., 1996), probably because one of the two limbs provides the necessary postural support while the other executes voluntary action as the complementary role action (Amiridis et al., 2007). We found that most of the participants placed slightly more of their weight on the foot of the dominant leg (range: 0-7% body weight), and that the variance in this measure of asymmetry was significantly correlated with CoP velocity (Figure 4, left). Similarly, most participants supported more of body weight on the back foot during TS and the variance in this asymmetry was significantly with the much greater CoP velocities during this condition (Figure 4, right). The asymmetrical weight distribution presumably enables an individual to initiate a rapid step response to a perturbation when it is necessary and determines the stability and flexibility of inter-leg coordination dynamics in postural control (Blaszczyk et al., 2000; Wang & Newell, 2012).

Conclusion

The force capacity of the ankle muscles contributes significantly to the control of postural sway during upright standing. The correlations with MVC torque were strongest for the dorsiflexors during NQS, but for the plantar flexors during the two more challenging conditions (TS and OLS). Even though we focused on the ankle and did not examine the correlations for the knee or hip muscles – a limitation of our study – the observed associations were substantial and underscore a key role for the ankle muscles of older adults in maintaining balance during upright standing. Moreover, the improvement of static balance control in older adults may not require the symmetrical weight distribution. Additional studies are needed to explain the variance in the postural sway metrics that were not accounted for by the MVC torques. Nonetheless, the findings inform clinicians about valid therapeutic targets.

Conflict of interest

Authors declare no conflict of interest or any disclosure of professional relationships with companies or manufacturers who will benefit from the results of the present study.

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