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The involvement of ankle muscles in maintaining balance in the upright posture is higher in elderly fallers.

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Abstract:

The purpose of this study was to determine whether the mechanical contribution of ankle muscles in the upright stance differed among young adults (YA) (n = 10, age: ~24.3), elderly non-fallers (ENF) (n = 12, age: ~77.3) and elderly fallers (EF) (n = 20, age: ~ 80.7). Torque and electromyographic (EMG) activity were recorded on the triceps surae and tibialis anterior during maximum and submaximum contractions in the seated position. EMG activity was also recorded in subjects standing still. Plantar flexor (PF) and dorsal flexor (DF) torques generated in the upright posture were estimated from the torque–EMG relationship obtained during submaximum contractions in the seated position. Center of pressure (CoP) displacement was measured to quantify postural stability. Results showed that, in upright standing, EF generated greater ankle muscle relative torque (i.e. PF+DF torque in the upright stance/PF+DF during maximum isometric torque) than non-fallers (i.e. ENF, YA). The greater involvement of ankle muscles in EF was associated with higher CoP displacement. PF+DF torque in the upright stance was no different among the groups, but PF+DF torque during maximum effort was impaired in older groups compared with YA and was lower in EF than ENF. These results suggest that the postural stability impairment observed with aging is highly related to ankle muscle weakness.

1. Introduction

Falls are a leading cause of morbidity and mortality in older people and concern approximately 30% of the population aged ≥ 65 years (Campbell et al., 1989; Rubenstein and Josephson, 2002). Several authors have reported that the deterioration in postural stability is a key factor in falls. Indeed, it is common to observe that postural sway during the quiet stance is larger in older adults than in younger individuals and even more pronounced in elderly adults with recent history of falls (Cattagni et al., 2014; Fernie et al., 1982; Maki et al., 1994; Mignardot et al., 2014; Simoneau et al., 2008). In an earlier investigation, we found that the center of pressure (CoP) displacement in fallers and non-fallers correlated negatively with the maximum isometric torque (MIT) of ankle muscles (Cattagni et al., 2014). Moreover, we observed that EF were weaker and that the slope of the regression line for EF was steeper than that for non-fallers implying that below a certain level of muscle force the functional impact of the decline in ankle muscle strength is greater. This result suggests that muscular weakness is a major cause of postural instability and may increase the risk of falling. Considering the inverted pendulum model, which is generally used to describe the upright stance in humans, the equilibrium equation for the foot shows that a variation in muscle torque developed at the ankle joint is directly and proportionally translated into a variation in CoP position (Morasso and Schieppati, 1999). This was confirmed by authors who reported that increased CoP displacement was related to increased calf muscle electromyographic (EMG) activity (Amiridis et al., 2003; Billot et al., 2010; Loram et al., 2004). In 2010, Billot et al. found a strong correlation between ankle muscle torque generated in the upright posture (expressed as a percentage of the torque developed during maximum voluntary contraction) and CoP displacement in young and elderly adults (Billot et al., 2010). They also reported that this relative torque was greater in elderly non-fallers (ENF) than in young adults (YA), thus indicating that aging is associated with an increased implication of ankle muscles in maintaining balance in the upright stance. As far as

we know, no investigation has measured the relative contribution of ankle muscles in the upright stance in elderly people with a history of falls.

Therefore, the first purpose of this study was to determine whether the mechanical contribution of ankle muscles in the upright stance differed among young adults, elderly non-fallers and elderly fallers. As EF are weaker and more unstable (Cattagni et al., 2014), we hypothesized that the relative torque needed to maintain the upright stance would be greater in EF than in ENF.

Several authors found that the MIT of plantar flexors (PF) decreased during aging whereas the MIT of dorsal flexors (DF) seemed less sensitive to the aging process (Billot et al., 2010; Cattagni et al., 2014; Simoneau et al., 2009). This observation may be extended to EF since we recently found that PF MIT was 33% lower in EF than in ENF while DF MIT was only 24% lower in EF (Cattagni et al., 2014). Therefore, the second aim was to determine, the relative contribution of both the antagonist ankle muscle groups in maintaining postural balance for the three populations (i.e. YA, ENF and EF). We supposed that the PF relative torque required to maintain the upright posture would be greater in ENF and even more so in EF than in YA while the DF relative torque would be less influenced by aging and by the history of falling. A greater mechanical contribution of PF, especially for EF, could imply a change in the muscle activation strategy adopted to maintain the upright posture.

2. Methods

2.1. Participants

Forty-two volunteers aged between 18 and 89 years old participated in this experiment. The population included three groups according to age and the history of falls: 10 healthy YA (age: 24.3 ± 6.3 years), 12 healthy ENF (age: 77.3 ± 5.2 years) and 20 EF (age: 80.7 ± 6.4 years). For subjects aged between 60 and 90 years old, the history of falls in the previous 6 months was recorded by interview. People who had fallen unexpectedly at least once in the previous six months and with- out musculoskeletal impairment were included in the EF group. The YA were students at the University of Burgundy. Exclusion criteria for all subjects were myopathy, neurological disorders (cerebral vascular accident, multiple sclerosis, Parkinson's disease), serious visual impairment, body mass index (BMI) ≥ 35 and impaired cognitive status (score of less than 23 on the Mini Mental State Examination). Informed written consent was obtained from all participants after they had been fully informed of all potential risks, discomfort and benefits of the study, which was approved by the local ethics committee.

The protocol of the current investigation was approved by the French National Drugs and Health Administration and by the National Ethics Committee section Dijon Est I and was carried out in accordance with the Declaration of Helsinki.

2.2. Data recording

2.2.1. Maximum isometric torque

Participants were examined in the seated position, their trunk strapped to a chair and inclined backwards at 40° to the vertical in order to minimize body motion, the knee joint angle at 180°

and the ankle joint angle at 90°. Torque was measured with the foot secured by two straps to the footplate of a custom-made ergometer developed by the mechanical workshop of a local engineering school (I.U.T. Génie Mécanique, Dijon, France). The center of rotation of the ergometer shaft was aligned with the anatomical ankle flexion–extension axis. The footplate was connected to a strain-gauge transducer (Société Doerler-Vandoeuvre, France) and an amplifier (PM Instrumentation, model 1300B, amplification: 8 V = 200 Nm, France). Torque was sampled at a frequency of 2 kHz and processed by a multichannel analogue-digital converter (Biopac Systems Inc., USA).

2.2.2. EMG recordings

EMG activity from the soleus, gastrocnemius medialis, gastrocnemius lateralis and tibialis anterior muscles was recorded during the postural task, and submaximum and maximum contractions. First, the subjects' skin was carefully prepared by shaving and cleaning with alcohol. Then, for soleus measurements, pairs of Ag/AgCl bipolar surface electrodes (Ambu Blue sensor M, Ambu A/S, Denmark) of 15 mm diameter with an interelectrode distance (centre-to-centre) of 2 cm were placed along the mid-dorsal line of the leg, approximately 5 cm below the insertion of the two heads of the gastrocnemii on the Achilles tendon (De Luca, 1997). Medial and lateral gastrocnemii recording electrodes were fixed lengthwise over the middle of the muscle belly. For the tibialis anterior, which represents the dorsal flexor group, the electrodes were positioned at 1/3 on the line between the fibula and the tip of the medial malleolus (Duclay et al., 2009; Hermens et al., 2000). The reference electrode was placed in a central position between the two gastrocnemii bellies on the right leg. The EMG signal was amplified with a bandwidth frequency ranging from 10 Hz to 5 kHz (Gain=1000). The EMG and mechanical signals were sampled at 2 kHz with the Biopac acquisition system and stored with commercially available software (Acqknowledge, MP 150, USA) for off-line analysis.

2.2.3. Postural stability

CoP displacement during the static postural task was assessed via a force-platform (Stabilotest, TechnoConcept, Cereste, France). This platform was composed of three force sensors which instantaneously measured the coordinates of the point of application of the resultant of the ground reaction forces or CoP. Data from the force-platform were sampled at a frequency of 40 Hz and were synchronised by triggering with the Biopac acquisition system.

2.3. Experimental procedure

All measurements were made in one experimental session lasting approximately 2h. All subjects first carried out the orthostatic postural task and then the maximum voluntary contraction (MVC) task to avoid interference due to fatigue during the postural task.

2.3.1. Postural task

During the postural task, subjects stood barefoot on the force-platform, with the two anteroposterior feet axes forming an angle of 30° (distance between the heels = 2 cm) and with their arms alongside their body. The subjects were asked to remain as still as possible, looking straight ahead at a target point located at eye height and placed at a distance of 150 cm in front of them. This postural task was repeated three times, and a rest period of 1 min was given between trials. During each 30 s trial, CoP displacement and EMG activity of soleus, gastrocnemius medialis, gastrocnemius lateralis and tibialis anterior were recorded.

2.3.2. Maximum isometric torque

The MIT for PF and DF was obtained during maximum voluntary isometric contraction tests performed for each leg separately. A three-minute warm-up, which included several submaximum plantar and dorsal flexions, was carried out. Subjects then performed two maximum contractions lasting 5 s at random for legs and ankle movements (i.e. Plantar flexion and dorsal flexion). A one-minute rest period was systematically given between trials to avoid any fatiguing effect on measurements. If there was a variation of more than 5% between the first and the second MVC, participants were asked to perform an additional MVC. Standardized verbal encouragement was given during attempts to produce the maximum effort. EMG activity of the soleus, gastrocnemius medialis, gastrocnemius lateralis and tibialis anterior was recorded during each MVC.

2.3.3. Submaximum contractions

To estimate PF and DF torque values during postural task for each leg, subjects were asked to perform in seated position 10-15 submaximum contractions, 4 s long, from 1% to 30% of PF and DF MIT. EMG activity of the soleus, gastrocnemius medialis, gastrocnemius lateralis and tibialis anterior was recorded during each submaximum contraction.

This method allowed us to plot for each subject the relationship between torque and EMG.

2.4. Data analysis

2.4.1. CoP displacement recording

During the postural task, the total length of CoP displacement was recorded for 30 s. This parameter was expressed as an average of three measurements. Then CoP displacement was normalized by the height (CoP/Height).

2.4.2. Maximum isometric torque

Only the highest PF and DF maximum voluntary contractions were considered for analysis. For each contraction type, maximum torques achieved by both legs were summed to obtain the overall PF or DF MIT. A sum of PF and DF MIT (PF+DF MIT) was calculated to obtain the overall force production capacity at the ankle joint. The MIT was then normalized by the body mass (MIT/Mass).

2.4.3. EMG activity

The EMG activities of the DF and PF were quantified as root mean square (RMS) values measured over 30 s during the postural task and 0.5 s during the submaximum and MVC effort tests. For each leg, the sum of the EMG-RMS of the three muscles of the triceps surae represented the activity of PF while the EMG-RMS of the tibialis anterior represented the activity of DF.

The relative muscle activity of PF and DF of each leg corresponded to the EMG-RMS value obtained during the postural task normalized to the EMG-RMS value obtained during the MVC of the corresponding muscle group and leg.

2.4.4. Antagonist and agonist isometric and postural torques

For each leg, the torque-EMG relationship for submaximum isometric contractions was plotted. The antagonist isometric torque and postural torque were determined from the equation of the torque-EMG relationship (Fig. 1). For each subject, r -values (i.e. Pearson correlation coefficient) of regression lines were calculated for the two legs and muscle actions (i.e. PF and DF).

From the inverted pendulum model, torque at the ankle joint depends on the body mass and the distance between the ankle joint and CoP position. Torque generated in the upright posture was

therefore divided by mass to obtain the torque per unit of mass (torque/Mass). During muscle effort, the net (recorded) moment does not reflect the EMG activity (or specific torque) of the ag/antagonist muscles (Simoneau et al., 2009). Furthermore it was previously reported an age-related difference in coactivation during maximal isometric effort (Billot et al., 2010; Simoneau et al., 2009). The agonist maximum isometric torque (AMIT) was therefore calculated for both PF and DF MVC. It corresponds to calculate the sum of the antagonist torque and the MIT (Billot et al., 2010; Simoneau et al., 2009). For example, to calculate PF AMIT, PF MIT was summed with the DF antagonist torque obtained from the torque-EMG relationship. The torque generated in the upright posture was also expressed as a percentage of the AMIT to determine the relative torque. As previously seen for MIT, AMIT of both legs were summed to obtain the overall PF or DF AMIT. A sum of PF and DF AMIT (PF+DF AMIT) was calculated in order to obtain the overall agonist isometric torque production capacity at the ankle joint. The AMIT was then normalized by the body mass (AMIT/Mass).

The AMIT was used to normalize torque during the postural task. The normalized value was regarded as the relative torque. In order to quantify the CoP displacement per unit of torque generated in the upright posture, CoP/Height was divided by PF+DF torque/Mass (CoP/torque).

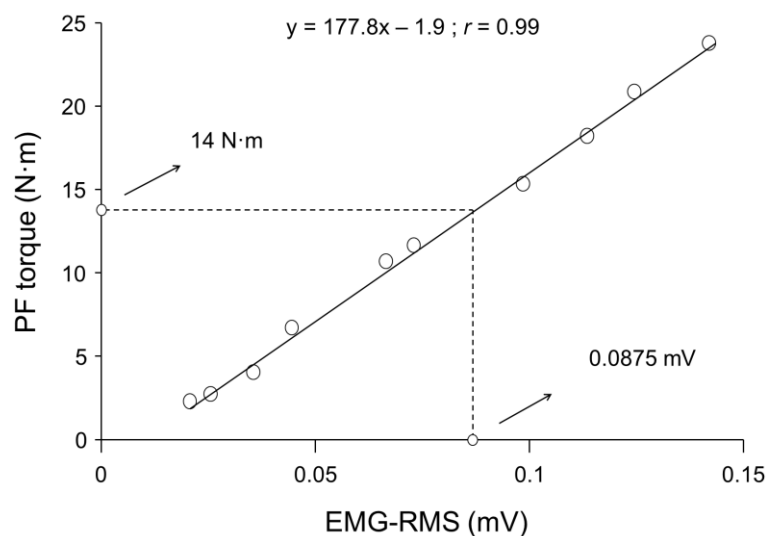


Figure 1. Torque-EMG relationship for plantar flexors (PF) in a representative elderly subject. PF torque recorded during submaximum contraction is plotted against the corresponding EMG-RMS value. Each circle corresponds to submaximum contraction. The linear best-fit regression line and the correlation coefficient (r) from Pearson's analysis are presented. The equation of the linear regression was used to determine the torque produced in the standing position from the ankle muscle EMG activity recorded in the upright posture and the antagonist isometric torque, recorded during the dorsal flexor isometric MVC. For example, a PF EMG-RMS value of 0.0875 mV, recorded in the upright posture, corresponds to a torque estimated at 14 N·m.

2.5. Statistical analysis

All statistical tests were performed with Statistica 10 (Statsoft, Tulsa, OK, USA). Data are presented as means \pm standard deviations (SD). The critical level for statistical significance was set at 5%. Normality criteria were tested using the Kolmogorov-Smirnov test. One-way ANOVA was performed on age, body mass, height, BMI, CoP, CoP/Height and CoP/torque to determine the differences between the three groups (YA, ENF and EF). Two-factor ANOVAs with repeated measures [Group (YA, ENF and EF) \times muscle group (PF and DF)] were performed on the MIT, MIT/Mass, AMIT, AMIT/Mass, torque/Mass and relative torque generated in the upright posture. Three-factor ANOVAs with repeated measures [Group (YA, ENF and EF) \times muscle group (PF and DF) \times leg (right and left)] were performed on relative EMG-RMS, postural torque and r -values determined from the equation of the torque-EMG relationship. Effect size for each statistical test was calculated as partial eta squared (η_p^2). When a main effect was found, Tukey's post-hoc analysis was performed. The best degree of association between PF+DF MIT/Mass and CoP/Height was determined by applying a log-linear correlation test to the whole sample and a linear correlation test to the faller and non-faller groups, separately (Cattagni et al. 2014). The best degree of association between CoP/Height and torque/Mass or relative torque was determined by a linear correlation test (Pearson product-moment correlation).

3. Results

3.1. Subjects' characteristics

The main characteristics of participants with statistical analysis (including values of partial eta squared) are described in table 1. Body mass and height were lower in both senior groups than in YA ($P<0.01$). Despite differences in height and mass, BMI was similar across the three groups.

Table 1. Subjects' characteristics, strength and balance performances.

	Young adults n = 10 range = 18-34 years	Elderly non-fallers n = 12 range = 65-83 years	Elderly fallers n = 20 range = 64-89 years	η_p^2
Age, years	24.3 ± 6.3	77.3 ± 5.2*	80.7 ± 6.4*	0.94
Body mass, kg	72.6 ± 8.2	60.8 ± 12.0*	60.3 ± 11.4*	0.19
Height, m	1.78 ± 0.02	1.64 ± 0.09*	1.58 ± 0.08*	0.56
Body mass index, kg/m ²	22.8 ± 2.7	22.5 ± 2.7	24.1 ± 4.0	0.05
PF MIT, N·m	315.8 ± 50.1	163.8 ± 69.2*	109.2 ± 35.8*\$	0.74
PF MIT/Mass, N·m·kg ⁻¹	4.4 ± 0.6	2.6 ± 0.7*	1.8 ± 0.5*\$	0.76
DF MIT, N·m	88.8 ± 9.5	51.9 ± 21.4	42.9 ± 10.8*	0.64
DF MIT/Mass, N·m·kg ⁻¹	1.2 ± 0.1	0.8 ± 0.2	0.7 ± 0.2*	0.57
PF+DF MIT, N·m	404.6 ± 54.6	215.6 ± 86.2*	155.4 ± 40.9*\$	0.75
PF+DF MIT/Mass, N·m·kg ⁻¹	5.6 ± 0.6	3.4 ± 0.8*	2.6 ± 0.6*\$	0.79
PF AMIT, N·m	342.8 ± 48.3	170.4 ± 72.0*	118.4 ± 34.8*\$	0.77
PF AMIT/Mass, N·m·kg ⁻¹	4.7 ± 0.6	2.7 ± 0.7*	2.0 ± 0.5*\$	0.79
DF AMIT, N·m	152.1 ± 32.9	74.1 ± 23.0*	64.2 ± 19.7*	0.70
DF AMIT/Mass, N·m·kg ⁻¹	2.1 ± 0.4	1.2 ± 0.3*	1.1 ± 0.3*	0.64
PF+DF AMIT, N·m	494.8 ± 73.2	244.5 ± 88.7*	182.5 ± 48.9*\$	0.79
PF+DF AMIT/Mass, N·m·kg ⁻¹	6.8 ± 0.9	3.9 ± 0.8*	3.0 ± 0.7*\$	0.81
CoP displacement, mm	312.7 ± 77.6	433.0 ± 149.7	648.9 ± 385.4*\$	0.21
CoP/Height	1.8 ± 0.4	2.7 ± 0.9	4.1 ± 2.4*\$	0.26

MIT: Maximum isometric torque; AMIT: Agonist maximum isometric torque; PF: Plantar flexors; DF: Dorsal flexors; CoP: Centre of pressure displacement; η_p^2 : Partial eta squared; Value are means ± SD. * Significant

difference from young subjects, $P<0.05$. \$ Significant difference from elderly non-fallers, $P<0.05$. † Significant difference from MIT, $P<0.05$.

3.2. CoP displacement and maximum isometric torque

CoP displacement and maximum isometric torque measurements are described in table 1 with statistical analysis (including values of partial eta squared). Since the statistical analyses showed similar changes for CoP and CoP/Height, MIT and MIT/Mass and AMIT and AMIT/Mass, we have only presented the results for the normalized value.

In the non-faller groups (i.e. YA, ENF), The CoP/Height was not affected by age since it was equivalent between YA and ENF (Table 1), but it was significantly higher in EF than in the non-faller groups ($P<0.01$).

As shown in table 1, PF MIT/Mass, PF AMIT/Mass, PF+DF MIT/Mass and PF+DF AMIT/Mass were significantly lower in older groups ($P<0.001$) than in YA and even lower in EF ($P<0.01$) than in ENF. The DF MIT/Mass was significantly ($P<0.05$) lower in EF than in YA but the value for ENF was similar to that of the other groups. DF AMIT/Mass was significantly ($P<0.001$) lower in the older groups than in YA, but similar in ENF and EF.

As shown in Figure 2, a significant relationship defined by a curvilinear function was found between CoP/Height and PF+DF MIT/Mass ($r=-0.53$, $P<0.001$). A significant linear correlation between these two parameters was observed in EF ($r=-0.46$, $P<0.05$) while no significant association was found in the non-faller group (i.e. YA + ENF) ($r=-0.31$, $P=0.15$)

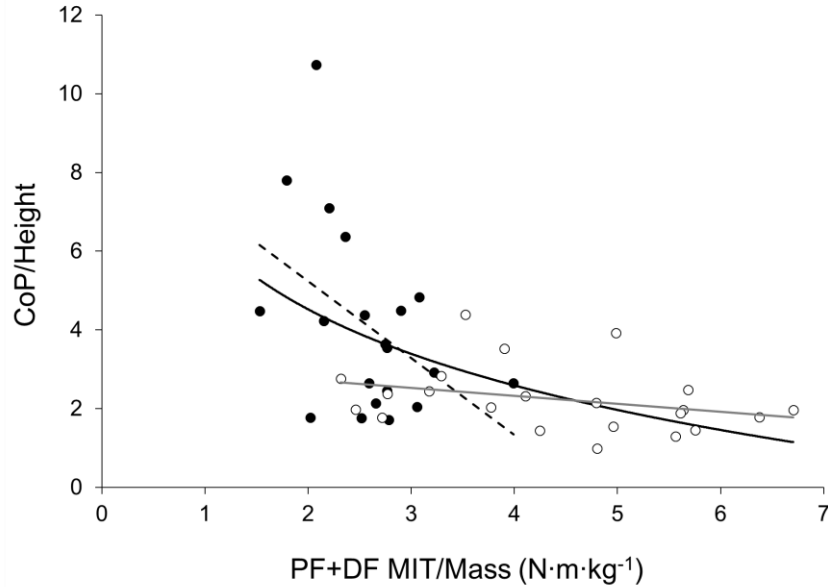


Figure 2: Relationship between CoP displacement normalized for height (CoP/Height) and the sum of plantar (PF) and dorsal (DF) flexor maximum isometric torque normalized for body mass (PF+DF MIT/Mass). Black circles represent elderly fallers and white circles represent young or elderly non-fallers. The continue black line represents the logarithmic relationship between CoP/Height and PF+DF MIT/Mass for the whole population; the equation is: $CoP/Height = -6.43 \cdot \log_{10}(PF+DF MIT/Mass) + 6.45$, ($r = -0.53$, $P < 0.001$). The dashed black line represents the linear relationship between CoP/Height and PF+DF MIT/Mass in elderly fallers; the equation is: $CoP/Height = -1.95 \cdot (PF+DF MIT/Mass) + 9.94$, ($r = -0.46$, $P < 0.05$). The gray line represents the linear relationship between CoP/Height and PF+DF MIT/Mass in the non-faller group (i.e. young adults and elderly non-fallers); the equation is: $CoP/Height = -0.20 \cdot (PF+DF MIT/Mass) + 3.14$, ($r = -0.31$, $P = 0.15$).

3.3. EMG activity and ankle muscle torque during upright posture

For the relative muscle activity (i.e. EMG-RMS activity in upright standing expressed as a percentage of EMG-RMS activity during MVC), the ANOVA analysis showed a group effect ($P < 0.001$; $\eta_p^2 = 0.46$) and a muscle effect ($P < 0.05$; $\eta_p^2 = 0.10$) with no interaction between the two ($P = 0.69$; $\eta_p^2 = 0.03$). The global relative EMG-RMS of ankle muscles was significantly ($P < 0.05$) higher in both older groups than in YA ($3.6 \pm 2.7\%$) and even higher in EF ($21.3 \pm 15.0\%$) than in ENF ($12.8 \pm 10.2\%$). It was also greater for PF ($16.8 \pm 11.6\%$) than for DF ($12.5 \pm 15.4\%$) whatever the group ($P < 0.05$).

The PF and DF torques in upright posture were estimated from the equation of the torque-EMG relationship obtained in the sitting posture. The statistical analysis showed no difference among

the three groups of subjects and no leg and muscle action (i.e. PF and DF) effects on r -values. All r -values were ranged from 0.90 to 1. The statistical analysis revealed that in the upright posture, DF torque (6.8 ± 5.4 N·m) was lower than PF torque (18.9 ± 7.5 N·m; $P < 0.001$; $\eta_p^2 = 0.46$) and no difference was observed among the three groups ($P = 0.45$; $\eta_p^2 = 0.04$). When the PF and DF torques generated in the standing position were normalized to the body mass (Fig. 3A), the statistical analysis revealed an interaction between group and muscle ($P < 0.05$; $\eta_p^2 = 0.15$). PF torque/Mass were higher in EF than in YA ($P < 0.05$) while no difference was observed between ENF and the other groups. Concerning DF torque/Mass and PF+DF torque/Mass, no difference was observed among the three groups. PF torque/Mass was significantly ($P < 0.001$) higher than DF torque/Mass in the older groups but not in YA.

When ankle muscle torques generated in the upright posture were normalized to the corresponding AMIT (Fig. 3B), the statistical analysis revealed an interaction between group and muscle ($P < 0.05$; $\eta_p^2 = 0.16$). PF and PF+DF torque/AMIT were higher in EF than in both non-faller groups ($P < 0.05$) while no difference was observed between YA and ENF. Concerning DF torque/AMIT, no difference was observed among the three groups. PF torque/AMIT was higher ($P < 0.01$) than DF torque/AMIT in EF, but no difference was observed in YA and ENF.

As shown in Figure 4, positive correlations were observed between: i) PF+DF torque/AMIT and CoP/Height ($r = 0.77$, $P < 0.001$, Fig. 4A), ii) PF torque/AMIT and CoP/Height ($r = 0.63$, $P < 0.001$, Fig. 4B) and iii) DF torque/AMIT and CoP/Height ($r = 0.65$, $P < 0.001$, Fig. 4C).

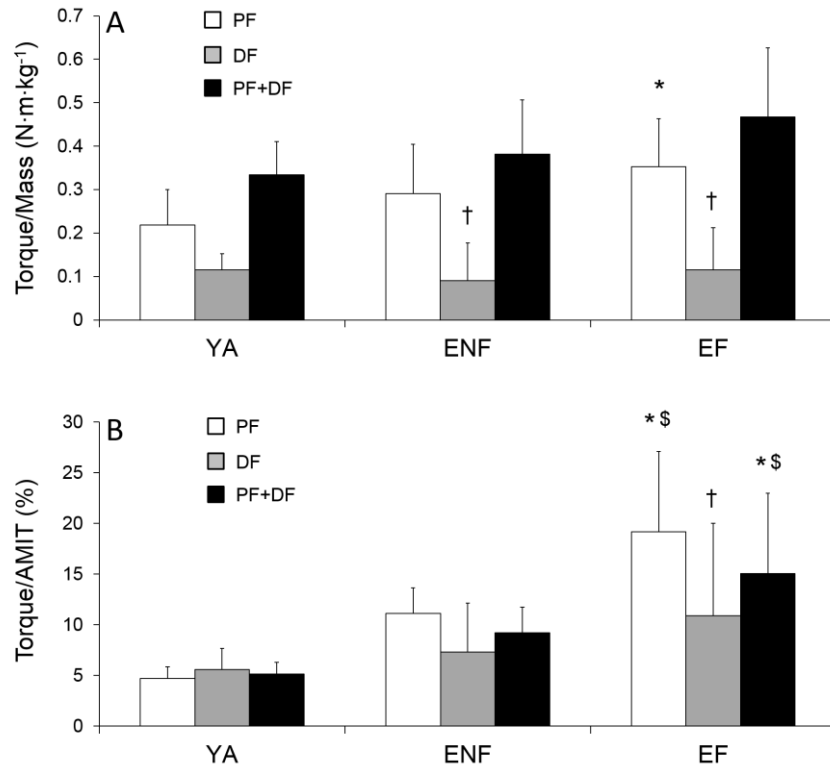


Figure 3. Ankle muscle torque generated in the upright posture in young adults (YA), elderly non-fallers (ENF) and elderly fallers (EF). (A) represents the plantar flexor (PF), dorsal flexor (DF) and PF+DF torques generated in the upright posture normalized to body mass. (B) represents PF, DF and PF+DF torques generated in the upright posture normalized to the respective maximum isometric torque. * Significant difference from YA, $P < 0.05$. \$ Significant difference from ENF, $P < 0.05$. † Significant difference from PF, $P < 0.05$.

To compare, among the three groups, body sway according the PF+DF torque generated for maintaining balance, the ratio between CoP/Height and PF+DF torque/Mass (CoP/torque) was calculated. The statistical analysis showed that EF generated a higher CoP/torque than the YA ($P < 0.01$) while ENF had the same torque/CoP as the YA and EF groups (Fig. 5).

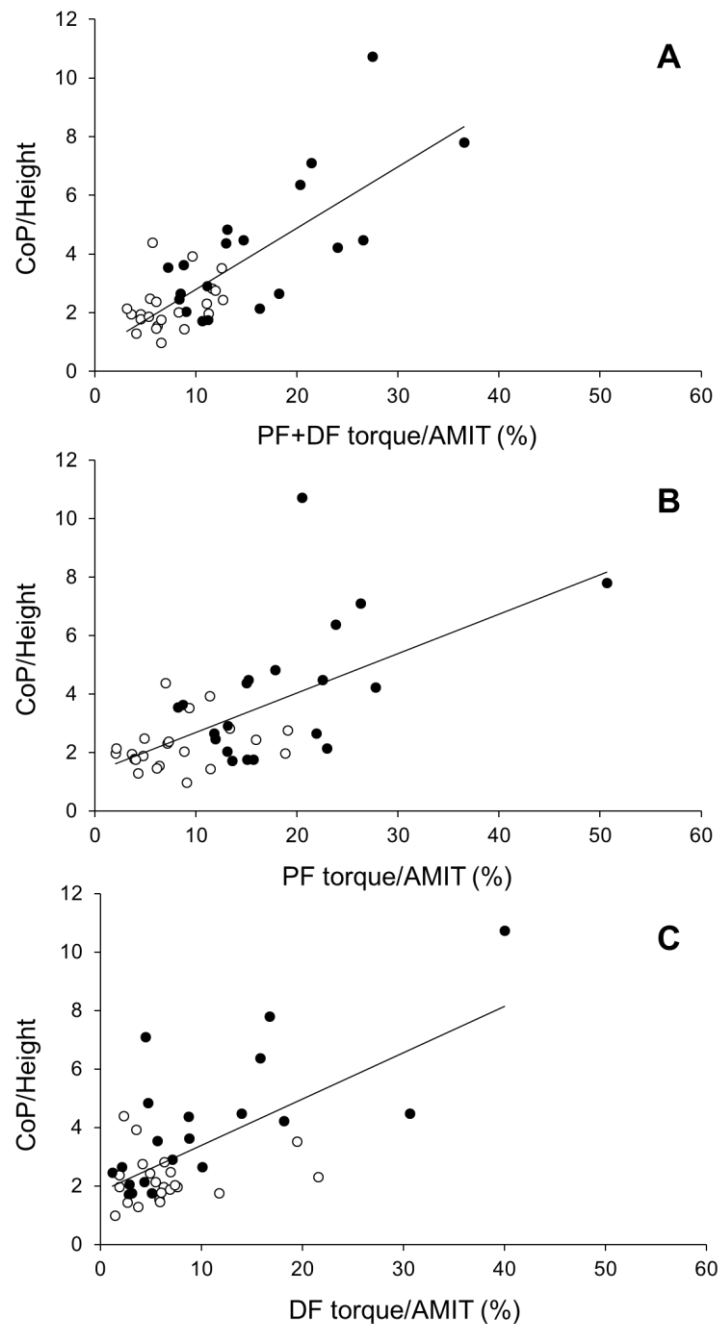


Figure 4. Relationship between the centre of pressure displacement (CoP) and ankle muscle relative torque generated in the upright posture for the whole population. Black circles represent elderly fallers and white circles represent young or elderly non-fallers. (A) represents the linear relationship between CoP/height and the sum of plantar flexor (PF) and dorsal flexor (DF) (PF+DF) torques generated in the upright posture normalized and expressed as percentage of the PF+DF maximum agonist isometric torque (PF+DF torque/AMIT); the equation is: $CoP/Height=0.21 \cdot (PF+DF \text{ torque}/AMIT)+0.73$ ($r=0.77$, $P<0.001$). (B) represents the linear relationship between CoP/Height and the PF torque normalized and expressed as percentage of the PF maximum agonist isometric torque (PF torque/AMIT); the equation is: $CoP/Height=0.14 \cdot (PF \text{ torque}/AMIT)+1.28$ ($r=0.63$, $P<0.001$). (C) represents the linear relationship between CoP/Height and the DF torque normalized and expressed as percentage of the DF maximum agonist isometric torque (DF torque/AMIT); the equation is: $CoP/Height=0.16 \cdot (DF \text{ torque}/AMIT)+1.74$ ($r=0.65$, $P<0.001$).

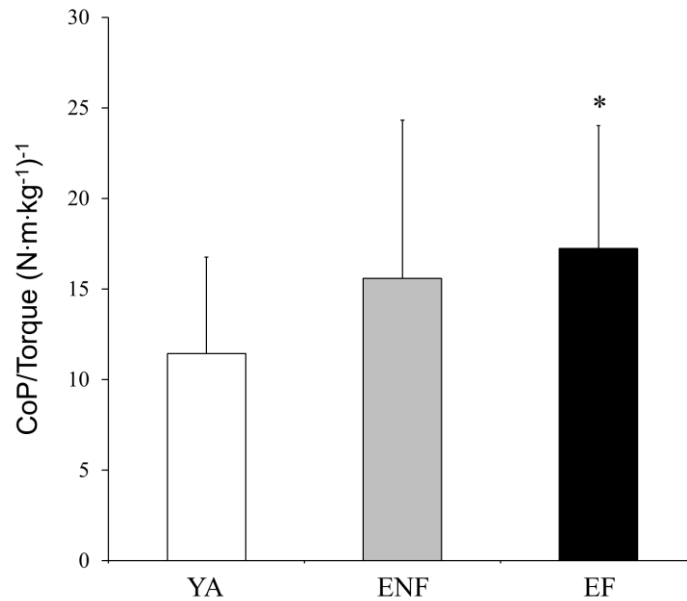


Figure 5. Centre of pressure displacement (CoP) normalized to the torque of ankle muscles generated in the upright posture in young adults (YA), elderly non-fallers (ENF) and elderly fallers (EF). Columns represent the CoP normalized to body height and divided by the sum of torques generated by plantar flexors and dorsal flexors in the upright posture, normalized to body mass. Data are mean \pm standard error. * Significant difference from YA, $P < 0.01$.

4. Discussion

The present results showed that EF exerted a higher relative torque in the upright stance than did non-fallers (i.e. YA and ENF). This higher relative torque was associated with increased CoP displacement, suggesting that postural instability was related to increased involvement of ankle muscles in maintaining balance. Because submaximum PF+DF torque/Mass generated by ankle muscles when standing was no different among the three groups, but the PF+DF MIT/Mass was lower in ENF and even more so in EF, we can deduce that the increase in postural sway with aging is related to the decline in maximum ankle muscle performance.

In addition, in the standing position, the relative contribution of PF was greater than that of DF in both groups of seniors (with and without a history of falls), while it was no different in YA. These results suggest that in the standing position impaired postural stability could be due in part to an excessive contribution of the PF muscles.

4.1. MIT and CoP displacement

The higher PF+DF AMIT/Mass in older adults (ENF and EF) than in YA indicated an age-related decrease in maximum capacity of torque production of ankle muscles. Our results showed that both PF and DF AMIT/Mass were affected by aging, as both were lower for ENF than for YA. Although the majority of authors have reported the same trend for PF muscles (Billot et al., 2010; Cattagni et al., 2014; Davies et al., 1986; Simoneau et al., 2006, 2005, 2009; Vandervoort and McComas, 1986; Winegard et al., 1996), there is no consensus in the literature concerning DF muscles. Indeed, in agreement with our results, some authors observed a decrease in maximum performance of DF with aging (Lanza et al., 2003; Vandervoort and McComas, 1986) while other found preservation (Billot et al., 2010; Cattagni et al., 2014; Lanza et al., 2004; Simoneau et al., 2006, 2005, 2009). By comparing ENF with EF, we observed that the PF+DF AMIT/Mass in the EF group was lower than that in the ENF group. The more pronounced weakness of EF seems to be related to a particular impairment of PF muscles. Indeed, PF AMIT/Mass was lower in EF than in ENF while DF AMIT/Mass was similar in both groups. This observation is in keeping with previous studies, which showed that in elderly fallers muscle weakness particularly concerns PF muscles (LaRoche et al., 2010; Perry et al., 2007; Sieri and Beretta, 2004).

The CoP displacement was no different in young adults and elderly non-fallers (Billot et al., 2010; Cattagni et al., 2014; Shumway-Cook et al., 1997; Simoneau et al., 2008). However, the lack of a difference could be due to the higher variability observed in ENF than in YA (coefficient of variation: 34% and 25% for ENF and YA respectively). We observed that the CoP/Height displacement was higher in individuals with a history of falls than in both non-faller groups. The higher body sway in EF associated with reduced muscle strength suggests

that postural instability and the risk of falling could be the consequence of a deficit in maximum capacity of torque production at the ankle joint. In agreement with our previous findings (Cattagni et al., 2014), the present results confirmed a negative curvilinear relationship between CoP/Height and PF+DF MIT/Mass, indicating that the greater the instability of an individual the more his balance depends on ankle muscle strength. A significant linear correlation was found in elderly fallers but not in the non-faller group (i.e. YA + ENF). We assume that the lack of significant correlation between CoP/Height and PF+DF MIT/Mass in non-fallers was mainly due to the size of the sample. Indeed, in a previous study where we included 60 non-faller subjects (instead of the 22 in the present investigation), we found an association between CoP/Height and PF+DF MIT/Mass (Cattagni et al., 2014)

4.2. Postural torque and CoP displacement

We observed that the relative EMG-RMS (i.e. the ratio between EMG-RMS of ankle muscles recorded in the upright stance and EMG-RMS recorded during MVC) of ankle muscles was higher in ENF than in YA and even more so in EF. This finding is consistent with previous works (Baudry and Duchateau, 2012; Baudry et al., 2014; Billot et al., 2010; Laughton et al., 2003), which showed that the age-related change in the postural control system involved a strategy by which older individuals increased the level of muscle activity across their lower-limb joints. However, when considering the relative torque exerted in the upright posture, no difference was observed between YA and ENF. This lack of difference may be explained by the inter-subject variability of this parameter, which was greater in ENF (coefficient of variation: 69%) than in YA (coefficient of variation: 47%) or by the fact that to produce a similar relative torque at the ankle joint, older adults needed more EMG activity. As a matter of fact, muscle fibres become weaker with aging, suggesting lower electromechanical effectiveness (Frontera et al., 2000).

The overall contribution of ankle muscles (i.e. plantar and dorsal flexors) in standing was higher in EF than in ENF.

Furthermore, PF+DF torque/AMIT exerted in the upright stance correlated strongly ($r=0.77$) with CoP/Height in the whole population. These results suggest that the decline in postural stability is associated with a greater relative contribution of ankle muscles in standing. However, as the PF+DF torque generated to maintain postural balance was no different among the three groups and the PF+DF AMIT was lower in ENF and in EF than in YA, the present study suggests that balance control is predominantly related to the maximum capacity of force production of ankle muscles. However, the ratio between CoP/Height and PF+DF torque/Mass generated in the upright stance was higher in EF than in YA, indicating that for the same contribution of ankle muscles, the CoP displacement was greater in EF. This result could be, at least in part, explained by the age-related impairment of the proprioceptive system. Indeed, it is well known that muscle spindles are affected by age. Some authors reported an increase in capsular thickness (Swash and Fox, 1972), a decrease in muscle spindle sensitivity (Kim et al., 2007; Miwa et al., 1995) and in diameter and number of intrafusal muscle fibres (Kararizou et al., 2005; Swash and Fox, 1972). Moreover, the efficiency of spinal transmission from Ia afferents to α -motoneurons is altered with aging (Baudry et al., 2014; Kido et al., 2004). As a result, the sensitivity to detect a CoP movement could be affected, and the muscle response to a CoP displacement could be altered and delayed.

PF+DF torque/AMIT needed to maintain the upright posture was higher in EF than in ENF. The statistical analysis showed that this difference was the consequence of the weakening of PF muscles rather than of DF muscles. Indeed, the present result showed that the DF torque/AMIT generated in the upright posture was no different among the three groups of subjects whereas the PF torque/AMIT was higher in EF than in non-fallers (i.e. ENF and YA). In addition, it should be noted that, in the two older groups, the torque/AMIT generated in the

upright stance was greater in PF than in DF. Since the raw data for PF and DF postural torques were similar in the two older groups, it can be concluded that the higher relative contribution of PF observed in EF was mainly due to the fact that compared with DF AMIT/Mass, PF AMIT/Mass is a better discriminator between fallers and non-fallers (-27.4% and -11.8% between ENF and EF for PF and DF, respectively). These results suggest that the decline in postural stability with aging is related to the alteration in the maximum capacity of torque production of ankle muscles, and especially of PF.

The greater relative implication of postural muscles when standing implies an increased energy cost of balance maintenance and premature muscle fatigue, which could impair postural stability. It is well known that fatigue of lower limb muscles is responsible for increased postural sway (Bisson et al., 2012; Boyas et al., 2013; Corbeil et al., 2003; Lepers et al., 1997; Nam et al., 2013; Salavati et al., 2007; Yaggie and McGregor, 2002).

5. Conclusion

We observed that in the upright stance, the contribution of ankle muscles, which was greater in older adults with history of falls than in non-fallers, was associated with increased body sway. We found that the greater contribution of ankle muscles in maintaining the upright stance was the consequence of a decline in the maximum capacity of force production, which concerned the PF muscles in particular. However, it cannot be excluded that other factors, such as impaired proprioception, could contribute to postural instability. In the light of these results, we may assume that the decline in MIT of PF muscles is strongly implicated in postural instability and the risk of falling.

It thus seems important, for prevention purposes or the rehabilitation of frail elderly patients, to set up strength training programmes focused on ankle muscles, with particular attention to PF. Strength training programmes could increase the maximum capacity of force production and

therefore decrease the relative muscular contributions needed to maintain balance in the upright stance.

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