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RESEARCH ARTICLE

Impaired Mediolateral Postural Control at the Ankle in Healthy, Middle-Aged Adults

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ABSTRACT. Elderly adults sway more than young adults. Based on the literature, the authors expected the mediolateral ankle postural control mechanism to be affected before age 60 years. Twelve healthy young adults (24.21 ± 2.50 years) and 12 middle-aged adults (51.13 ± 6.09 years) participated in the study. To challenge mediolateral stance, the conditions modified the mediolateral distance among the feet (narrow and standard distances), mandibular position (rest position, left and right laterality occlusion positions), and the occlusion with clenching (intercuspal occlusion, left and right maximal voluntary clenches). As we expected, middle-aged adults exhibited significantly reduced contribution of the ankle mechanism. It was so both in narrow and standard stances. A second objective was to show a greater contribution of the 2 mechanisms in narrow than in standard stances. The results confirmed our hypothesis. As we expected, mandibular conditions only had a significant effect on center of pressure sway. Unexpectedly, middle-aged adults did not increase their range of center of pressure sway in narrow stance. They may have overconstrained their center of pressure sway because of their ankle impairments. On the practical level, our results suggest that older adults should increase their stance width to relieve their hip and ankle control mechanisms and to stabilize their mediolateral posture.

Keywords: age-related impairments, center of pressure sway, mediolateral axis, postural control mechanisms, stance width

In stance, individuals sway all the time because of internal and external constraints (e.g., organ movement, gravity). Hence, postural control mechanisms have to work continuously to maintain postural stability (Winter, 1995). These mechanisms act conjointly and predominantly around the ankle and hip joints in the anteroposterior (AP) axis (Bardy, Oullier, Bootsma, & Stoffregen, 2002; Nashner & McCollum, 1985) and mediolateral (ML) axis (Winter, Prince, Frank, Powell, & Zabjek, 1996; Winter, Prince, Stergiou, & Powell, 1993). In the ML axis, Winter et al. (1996; 1993) identified an ankle-based mechanism (center of pressure change [COP_c]) and a hip-based mechanism (center of pressure vertical [COP_v]).

For subjects in quiet stance with a spontaneously chosen stance width (i.e., standard stance), COP_v and COP_c were shown to be the prime mechanisms involved in controlling ML and AP COP displacements, respectively (e.g., Winter et al., 1996; Winter et al., 1993). In the ML axis, COP_c was subsequently found to have a significant secondary role in controlling COP displacement in standard stance (Gatev, Thomas, Kepple, & Hallett, 1999; Termoz et al., 2008). Only Termoz et al. analyzed age-related changes in both ankle and hip postural control mechanisms. Their comparative study of young adults (*M* age = 27.1 years) and elderly adults (*M* age = 60.4 years) did not highlight any significant age-related differences in either COP_v or COP_c.

In the literature, adults over age 60 years have been found to show (a) significantly higher body sway amplitudes (Maurer & Peterka, 2005) and velocities (Prieto, Myklebust, Hoffmann, Lovett, & Myklebust, 1996) and (b) significantly lower body sway frequencies (McClenaghan et al., 1995), relative to young adults. All these effects were particularly pronounced in the ML axis (Maki, Holliday, & Topper, 1994; McClenaghan et al., 1995). Therefore, the finding that postural control mechanisms withstand the effects of aging (Termoz et al., 2008) is unexpected—especially for the ML axis. One would expect the greater ML postural sway in elderly adults to be due to significant impairment in at least one postural control mechanism (Maurer & Peterka, 2005). One significant shortcoming of Termoz et al.'s study was the instruction given to participants to move as little as possible during the trials. This requirement may have masked age-related changes in body stiffness (Cenciarini, Loughlin, Sparto, & Redfern, 2010; Maurer & Peterka, 2005) and other physiological impairments at the ankles in particular (Barr, Browning, Lord, Menz, & Kendig, 2005; Gilsing et al., 1995). Indeed, Fitzpatrick, Taylor, and McCloskey (1992) reported that standing as steadily as possible can influence reflex muscle stiffness. Another shortcoming of Termoz et al.'s study relates to the fact the researchers did not vary the difficulty of the stance conditions in the ML axis. It can be hypothesized that use of more difficult ML stance conditions would reveal age-related differences in COP_v or COP_c.

One way of challenging ML postural control is to shorten the stance width when these feet are side by side (Day, Steiger, Thompson, & Marsden, 1993; Mouzat, Dabonneville, & Bertrand, 2004). In the literature, Gatev et al. (1999) analyzed ankle and hip mechanisms in narrow and standard stances in seven healthy male subjects (*M* age = 42.3 years). The researchers compared the stances in terms of the center of mass, the COP, angular motions, and electromyography activities and performed cross-correlation analyses of the time-series. Their results showed that ankle and hip mechanisms made significantly greater contributions to ML body sway control in narrow stance than in standard stance. Another way of challenging ML postural control is to perform lateral modifications of the masticatory system (e.g., left and right laterality occlusion positions and lateral occlusion with clenching). This type of challenge may alter the stomatognathic system, which relays vestibular and visual information of importance for postural control

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(Buisseret-Delmas, Compoint, Delfini, & Buisseret, 1999). Gangloff and Perrin (2002) showed that in quiet stance with the eyes open, anesthesia of the trigeminal nerve increased the area of COP displacement. One can consider that performing intentional, lateral modifications of the masticatory system as a secondary task may destabilize the participants to some extent or is at least unlikely to increase their stability.

In the present study, our primary objective was to demonstrate significant, age-related differences in ML postural control mechanisms in general and at the ankle in particular. We included both middle-aged (under 60 years) adults and young adults because we expected to see the onset of significant changes in ML COP_c (but not in COP displacements) to occur in the adults under 60 years of age (Maurer & Peterka, 2005). Middle-aged adults usually show greater postural stability than adults over the age of 60 (Abrahamova & Hlavacka, 2008; Era et al., 2006). In order to detect age-related changes in the contributions of the two ML postural control mechanisms (ankle and hip), we deliberately modified the difficulty of ML stance by changing (a) the stance width and (b) the side on which jaw position or clenching was maintained. We assumed that more challenging experimental conditions would better reveal age-related impairments in ML COP_c. For young adults (control data), we expected to reproduce the literature findings for standard stance, that is to say a high contribution of the hip mechanism, a low but nevertheless significant contribution of the ankle mechanism (Gatev et al., 1999; Termoz et al., 2008) and no cooperation between the ankle and hip mechanisms (Winter et al., 1996; Winter et al., 1993). To elaborate on earlier findings regarding postural control mechanisms, our secondary objective was to highlight significant changes in these mechanisms under conditions that challenged ML postural control. We expected that narrow stance would increase the contribution of both ML ankle and hip mechanisms in all the participants, at least in terms of higher cross-correlation coefficients between time-series (Gatev et al., 1999). However, this effect was expected to be much greater for ML COP_c than for ML COP_v, because ML COP_v versus COP_{net} cross-correlation is already very high in healthy adults in standard stance (Termoz et al., 2008; Winter et al., 1996; Winter et al., 1993). For all the participants, we expected to find that ML COP displacement was significantly influenced by jaw conditions, as shown by certain literature studies (for a review, see Amat, 2009; Cuccia & Caradonna, 2009). However, we did not expect to see a significant effect of the different jaw conditions on the postural control mechanisms. Indeed, postural control mechanisms are known to vary little under circumstances that do not threaten postural stability (Termoz et al., 2008).

Methods

Participants

Twelve students (6 women) from the Universities of Lille and 12 middle-aged adults (8 women) participated in this study. The mean age, bodyweight, and height were 24.21 ± 2.50 years, 61.75 ± 8.58 kg, and 173.25 ± 10.65 cm for the young adults, respectively, and 51.13 ± 6.09 years, 70.05 ± 16.72 kg, and 166.17 ± 9.32 cm for the middle-aged adults. Exclusion criteria were a history of neurological or musculoskeletal disease, vestibular problems, recurrent dizziness, any kind of surgery in the preceding six months, any known or treated disabilities at the ankles and hips, previous craniofacial trauma or surgery or signs and/or symptoms of temporomandibular disorders (joint or muscle pain or signs or previous treatment of temporomandibular luxation). Previous or current orthodontic treatment and dentition were not chosen to be part of the selection criteria. The participants gave their written, informed consent to participation.

Apparatus

A dual-top force platform (AMTI, Watertown, MA, USA) was used to record COP displacement with a 100 Hz sampling frequency. The platform was placed 1.50 m from a wall. In all trials, the participants faced the wall and looked at a black dot (visual angle: 1°) placed at eye height in front of them.

Conditions

The study combined four independent variables: age (young vs. middle-aged adults), stance width (narrow vs. standard stance), mandibular position (rest position and left and right laterality occlusion positions), and occlusion with clenching (intercuspal occlusion and left and right maximal voluntary clenches). Stance width, mandibular position, and occlusion with clenching were modified specifically to challenge ML stance. In the narrow stance condition, the participants placed their feet close together, with one foot on each platform. In the standard stance condition, the participants freely chose the most comfortable foot position. We did not impose the foot position because it can lead the participants to feel uncomfortable (Kapteyn, Bles, Njiokiktjien, Kodde, & Massen, 1983; McIlroy & Maki, 1997). However, we controlled the influence of foot position in supplementary analyses (see Results). In the mandibular rest position, the teeth were not in contact with each other (as is the case after swallowing). In the left and right laterality occlusion positions, the participants had to shift their jaw as far to the left or to the right as possible (without experiencing pain) and maintain the mandibular position for the duration of the trial. In the intercuspal occlusion, the participants had to clench their teeth throughout the trial. In the left and right clenching conditions, the participants again clenched their teeth but put more pressure on the left or right side of the jaw, respectively. In all trials, the participants were told to hold their hands by the side of the body.

Procedure

Before the first trial, the foot positions (in bare feet) were marked on a piece of paper taped to each platform. The experimenter explained the different trial conditions and checked that the participants were able to comply with the instructions.

The experiment consisted of two trials per condition, giving a total of 24 trials (each lasting 35 s). The trials were run in blocks, to avoid confusion for subtle changes between conditions. Four condition blocks (mandibular position and clenching in standard and narrow stances) were performed successively and were counterbalanced for half the participants. In the four blocks, the order of the mandibular and clenching conditions was randomized. All participants in both groups performed the same trials. To aid relaxation, the participants were instructed to sit down after 12 trials and at any time if needed. The instruction to relax during trials was repeated several times.

Variables and Analyses

Stance Width and Stance Angle

Stance width was defined as the distance between the heel centers. Stance angle was defined as the angle between the lines going through the middle of the big toe and the middle of the heel for each foot (cf. McIlroy & Maki, 1997).

Postural Control Mechanisms

To investigate postural control mechanisms, we used Winter et al.'s (1993) equations:

$$\mathrm{COP}_{\mathrm{net}}(t) = \mathrm{COP}_l(t)\frac{R_{vl}(t)}{R_{vl}(t)+R_{vr}(t)} + \mathrm{COP}_r(t)\frac{R_{vr}(t)}{R_{vl}(t)+R_{vr}(t)} \quad (1)$$

$$\mathrm{COP}_c(t) = \mathrm{COP}_l(t)\times 0.5 + \mathrm{COP}_r(t)\times 0.5 \quad (2)$$

$$\mathrm{COP}_v(t) = \mathrm{COP}_{\mathrm{net}}(t) - \mathrm{COP}_c(t) \quad (3)$$

In these equations, COP_l(t), COP_r(t), R_{vl}(t), and R_{vr}(t) correspond to the COP displacement and the vertical reaction forces under the left and right feet, respectively.

The displacement of $\mathrm{COP}_{\mathrm{net}}$ corresponds to the displacement of the COP under each foot, taking into account the weight under each foot (Equation 1). The contribution of COP_c to $\mathrm{COP}_{\mathrm{net}}$ was calculated by eliminating the contribution of COP_v (Equation 2). The contribution of COP_v to $\mathrm{COP}_{\mathrm{net}}$ was then calculated by subtracting the contribution of COP_c from $\mathrm{COP}_{\mathrm{net}}$ (Equations 1 and 3). It should be recalled that $\mathrm{COP}_{\mathrm{net}}$ relates to COP displacement, whereas COP_v and COP_c relate to the contribution of the hip and ankle mechanisms. COP_v and COP_c correspond to the parts of COP displacement that are controlled by the ankle and hip mechanisms, respectively. An example of $\mathrm{COP}_{\mathrm{net}}$, COP_v, and COP_c time series is shown in Figure 1. To test our hypotheses, the computation was performed in the ML axis but not in the AP axis. Moreover, the computation of COP_v (load–unload) in the AP axis may not be right, as a single platform under each foot cannot measure the extent to which the anterior and posterior parts of the foot load and unload, respectively (i.e., two platforms under each foot would be required). If AP COP_v is not measured objectively, then it may not be differentiated from AP COP_c.

As in Termoz et al. (2008), root mean square (RMS) COP_v and RMS COP_c were computed with respect to RMS $\mathrm{COP}_{\mathrm{net}}$ (in percentage). The higher the RMS of COP displacement explained by one of the mechanisms, the higher the amplitude contribution of that mechanism. As in previous studies (Lafond, Corriveau, & Prince, 2004; Termoz et al., 2008; Winter et al., 1996; Winter et al., 1993), normalized cross-correlations with zero lag were analyzed for three relationships: COP_v versus $\mathrm{COP}_{\mathrm{net}}$, COP_c versus $\mathrm{COP}_{\mathrm{net}}$, and COP_v versus COP_c. The higher the cross-correlation coefficient between one mechanism and $\mathrm{COP}_{\mathrm{net}}$ is, the more active that mechanism is (i.e., the longer its action to control $\mathrm{COP}_{\mathrm{net}}$). Both amplitude and cross-correlation analyses are complementary to calculate the contribution of the two mechanisms to control COP displacement. Indeed, the contribution of one mechanism may be significant if the amplitude contribution is sufficiently great relative to RMS $\mathrm{COP}_{\mathrm{net}}$ and if the active

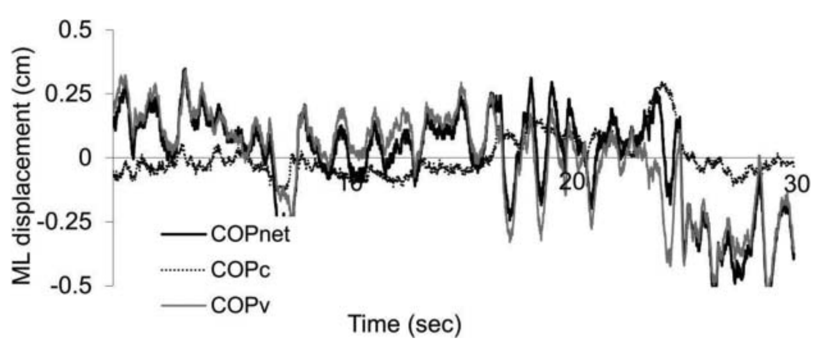

FIGURE 1. Representation of the time-series for net center of pressure ($\mathrm{COP}_{\mathrm{net}}$), center of pressure change (COP_c), and center of pressure vertical (COP_v) in one trial recorded in standard stance (cm) in the mediolateral (ML) axis. Thirty seconds of data are shown.

contribution is also sufficiently high. If the amplitude contribution were very large with no significant active contribution, the mechanism would not control COP displacement. In other words, the more similar the time series of the COP_c or COP_v and of COP_{net}, the better the contribution of that mechanism to control COP displacement. The similarity is analyzed both in terms of amplitude and phase. Supplementary information about the model is available in former manuscripts (Lafond et al., 2004; Termoz et al., 2008; Winter et al., 1996; Winter et al., 1993).

COP Displacement

As in other studies in which the stance width was modified, we used the mean velocity, standard deviation, and range to analyze COP displacement (e.g., Day et al., 1993; Kirby, Price, & MacLeod, 1987; Mouzat et al., 2004). These variables are also those typically used to compare body sway in young adults versus elderly adults (e.g., Maurer & Peterka, 2005; Prieto et al., 1996).

Analyses

In order to eliminate transitory behavior at the start of the trials, the first 5 s of data were not analyzed (Kinsella-Shaw, Harrison, Colon-Semenza, & Turvey, 2006). Each variable was calculated as the average of the two trials per condition. Repeated measures analyses of variance (ANOVA) with four factors (age, stance width, mandibular position, and clenching position) were performed on the dependent variables with a threshold for statistical significance set to $p < 05$. In secondary analyses, one-sample t tests were used to compare the amplitudes of the cross-correlation coefficients with zero. For these analyses, the threshold for statistical significance was set to $p < .025$ after Bonferroni adjustment.

Results

Amplitude Contribution of the Postural Control Mechanisms

RMS COP_c/RMS COP_{net}

The ANOVA showed a significant main effect of stance width only, $F(1, 22) = 4.72$, $\eta_p^2 = .15$, $p < .052$. The RMS COP_c/RMS COP_{net} was significantly greater in standard stance than in narrow stance (Figure 2, Table 1).

RMS COP_v/RMS COP_{net}

There was a significant main effect of stance width only, $F(1, 22) = 4.85$, $\eta_p^2 = .15$, $p < .052$. The RMS COP_v/RMS COP_{net} was significantly greater in standard stance than in narrow stance (Figure 2, Table 1).

Active Contribution of the Postural Control Mechanisms

COP_v versus COP_{net}

There was a significant main effect of stance width only, $F(1, 22) = 35.0$, $\eta_p^2 = .38$, $p < .053$. The COP_v versus COP_{net} was significantly greater in narrow stance than in standard stance (Figures 3A and 3B, Table 1).

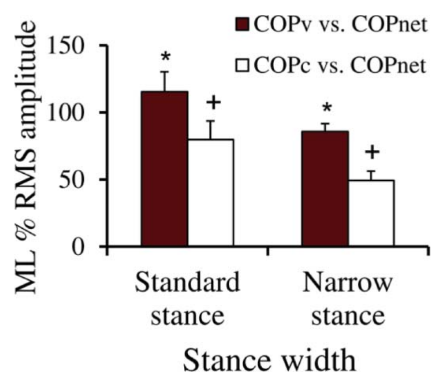

FIGURE 2. Mediolateral (ML) root mean square (RMS) amplitude of center of pressure change (COP_c) and center of pressure vertical (COP_v) expressed in percentage of the RMS amplitude of ML net center of pressure (COP_{net}). Under all conditions, the distance between the feet placed side by side (i.e., stance width) was either minimal (narrow stance) or spontaneously chosen (standard stance). * and + indicate a significant main effect of condition of the COP_v/COP_{net} and COP_c/COP_{net}, respectively. Error bars represent the standard error of the mean ($p < .05$). (Color figure available online).

COP_c versus COP_{net}

There were significant main effects of stance width, $F(1, 22) = 226.14$, $\eta_p^2 = .48$, $p < .053$, and group, $F(1, 22) = 5.12$, $\eta_p^2 = .16$, $p < .053$. The COP_c versus COP_{net} was significantly greater in narrow stance than in standard stance and significantly greater in young adults than in middle-aged adults (Figures 3A and 3B, Table 1).

COP_v versus COP_c

There were significant main effects of stance width, $F(1, 22) = 348.94$, $\eta_p^2 = .48$, $p < .053$, and group, $F(1, 22) = 4.34$, $\eta_p^2 = .14$, $p < .053$. The COP_v versus COP_c was significantly greater in narrow stance than in standard stance and significantly greater in young adults than in middle-aged adults (Figure 3A and 3B, Table 1).

The three analyses did not reveal any significant effect of mandibular or clenching position ($p = ns$).

COP Displacement

COP Standard Deviation

The ANOVA showed a significant main effect of stance width only, $F(1, 22) = 123.92$, $\eta_p^2 = .46$, $p < .054$. The COP standard deviation was significantly greater in narrow stance than in standard stance (Figure 4A, Table 1).

TABLE 1. Statistics of the Behavioral Dependent Variables as a Function of the Group (Young and Middle-Aged Adults) and the Stance Width (Standard and Narrow Stances)

	Young adults		Middle-aged adults	
Dependent variables	Standard stance	Narrow stance	Standard stance	Narrow stance
COP sway				
SD	0.15 ± 0.04^{a}	0.55 ± 0.19^{a}	0.15 ± 0.06^{a}	0.42 ± 0.10^{a}
Range	$0.82 \pm 0.20^{a,b,c}$	$2.73 \pm 0.78^{a,b,c}$	$0.80 \pm 0.04^{a,b,c}$	$0.79 \pm 0.05^{a,b,c}$
M velocity	1.37 ± 0.17^{a}	1.64 ± 0.15^{a}	1.25 ± 0.21^{a}	1.55 ± 0.21^{a}
Postural control mechanisms				
COP_v vs. COP_{net}	0.97 ± 0.01^{a}	0.99 ± 0.00^{a}	0.96 ± 0.01^{a}	0.99 ± 0.00^{a}
COP_c versus COP_{net}	$0.23 \pm 0.06^{a,b}$	$0.93 \pm 0.02^{a,b}$	$0.06 \pm 0.08^{a,b}$	$0.82 \pm 0.05^{a,b}$
COP_v versus COP_c	$0.03 \pm 0.07^{a,b}$	$0.89 \pm 0.02^{a,b}$	$-0.16 \pm 0.09^{a,b}$	$0.76 \pm 0.05^{a,b}$
RMS COP_v /RMS COP_{net}	113.47 ± 17.26^{a}	77.39 ± 5.26^{a}	117.31 ± 26.80^{a}	94.03 ± 9.52^{a}
RMS COP_c /RMS COP_{net}	83.13 ± 20.80^{a}	46.73 ± 3.27^{a}	76.28 ± 21.21^{a}	51.86 ± 13.48^{a}

Note. The dependent variables were the standard deviation, the range and mean velocity of center of pressure (COP) sway; the cross-correlation coefficients of COP_v versus COP_{net}, COP_c versus COP_{net}, and COP_v versus COP_c; and the root mean square (RMS) amplitude of COP_v and COP_c relative to the RMS of COP_{net}. The mean values ± the standard deviations of the dependent variables are given. COP_c = center of pressure change; COP_{net} = center of pressure net; COP_v = center of pressure vertical.
[a]Significant main effect of condition ($p < .05$).
[b]Significant main effect of group ($p < .05$).
[c]Significant group by condition interaction effect ($p < .05$).

COP Range

There were significant main effects of group, $F(1, 22) = 62.59$, $\eta_p^2 = .43$, $p < .05$; and stance width, $F(1, 22) = 76.05$, $\eta_p^2 = .44$, $p < .05$; and a significant group by stance width interaction effect, $F(1, 22) = 77.18$, $\eta_p^2 = .44$, $p < .054$. Only in young adults, the range of COP displacement was significantly greater in narrow stance than in standard stance (Figure 4B, Table 1).

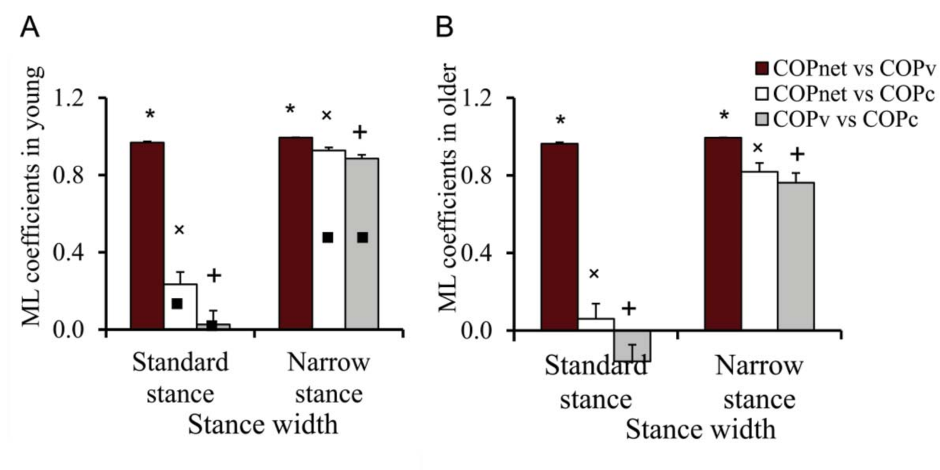

FIGURE 3. Mediolateral (ML) cross-correlation coefficients in young adults (**A**) and middle-aged adults (**B**). Black: Cross-correlation between the integrated displacement of the center of pressure (COP) under the left and right foot (COP_{net}) and the COP displacement under the control of the hip mechanism (COP_v vs. COP_{net}). White: Cross-correlation between COP_{net} and the COP displacement under the control of the ankle mechanism (COP_c vs. COP_{net}). Gray: Cross-correlation between COP_v and COP_c (COP_v vs. COP_c). Under all conditions, the distance between the feet placed side by side (i.e., stance width) was either minimal (narrow stance) or spontaneously chosen (standard stance). ×, *, and + indicate a significant main effect of condition of the COP_v versus COP_{net}, COP_c versus COP_{net}, and COP_v versus COP_c relationships, respectively. ■ indicates a main effect of group. Error bars represent the standard error of the mean ($p < .05$). (Color figure available online).

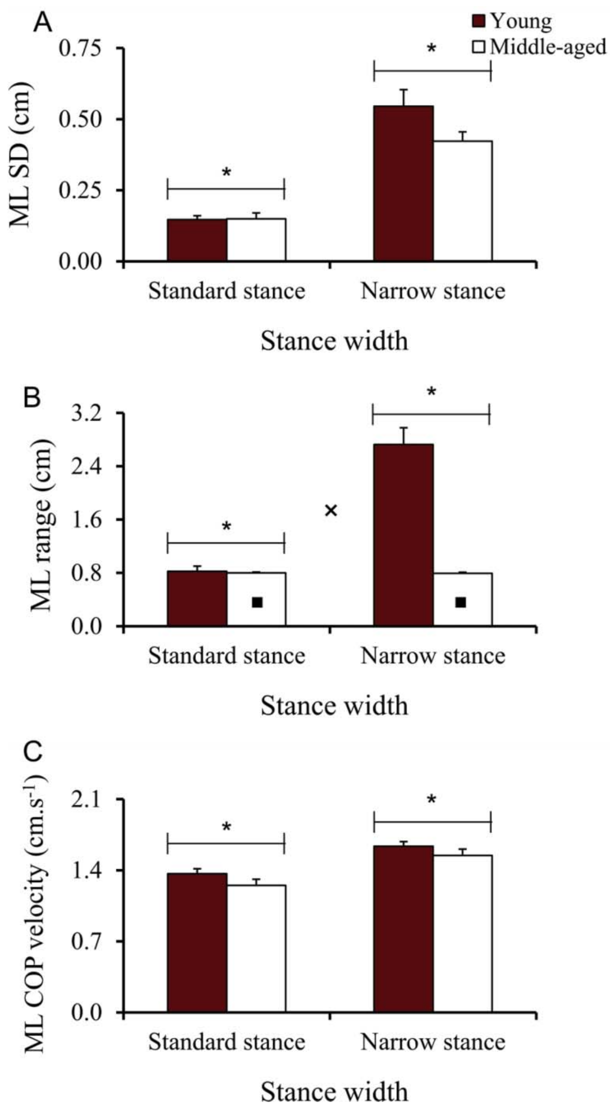

FIGURE 4. Significant effects in an analysis of variance for (**A**) the standard deviation, (**B**) the range, and (**C**) the mean velocity (velocity) of center of pressure (COP) displacement in the mediolateral (ML) axis (cm). Under all conditions, the distance between the feet placed side by side (i.e., stance width) was either minimal (narrow stance) or spontaneously chosen (standard stance). The figure displays the mean and standard error of the mean for young and middle-aged adult participants. The significant main effects of stance width and group are indicated by * and ■, respectively. The significant group by stance width interaction is indicated by × ($p <$ .05). (Color figure available online).

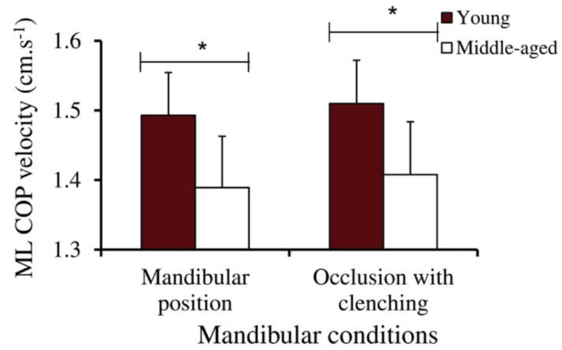

FIGURE 5. A significant effect of mandibular conditions in an analysis of variance for the mean velocity of center of pressure (COP) displacement in the mediolateral axis (ML COP velocity [cm.s^{-1}]). The mandibular position was either centered or deviated leftward or rightward. In the clenching conditions, the teeth were clenched in an intercuspal position or more strongly on the right or on the left. The figure displays the mean and standard error of the mean for young and middle-aged participants. * indicates a significant main effect of mandibular condition ($p <$ .05). (Color figure available online).

COP Mean Velocity

The ANOVA showed significant main effects of clenching, $F(1, 22) = 5.91$, $\eta_p^2 = .17$, $p < .05$ (Figure 5), and stance width, $F(1, 22) = 69.66$, $\eta_p^2 = .43$, $p < .054$. The COP mean velocity was significantly faster in occlusion with clenching (1.46 ± 0.17) than in mandibular position (1.44 ± 0.18). This significant effect may seem surprising because the means are quite similar (Figure 4C). However, it should be noted that the partial eta squared was much smaller than in other analyses. The COP mean velocity was also significantly faster in narrow stance than in standard stance (Table 1).

Significance of the Active Contribution

Above, the mandibular positions and clenches conditions did not lead to any significant effect. On this basis, only the stance width factor was included in the new analyses. One-sample t tests were used to compare only those correlation coefficients that were close to zero (i.e., ML COP_c vs. COP_{net} and ML COP_v vs. COP_c in both groups in standard stance). These t tests showed that ML COP_c versus COP_{net} in standard stance was significantly greater than zero in young adults only, $t(11) = 3.67$, $p < .025$ (Figures 3A and 3B, Table 1). This is an important finding relative to our main objective. The ML COP_v versus COP_c values did not differ from zero in either of the two groups, $t_s(11) < 1.86$, $p > .025$.

Analyses Controlling for the Foot Position, Height, Weight, and Age

The two groups were compared in terms of stance width and stance angle (Figure 6) in standard and narrow stances. Independent t tests did not show any significant inter-group differences, $t_s(22) < 1.45$, $p > .05$.

To further control for the influence of foot position (stance width, stance angle) and also height, weight and age on the dependent variables, we used the normalization procedure already adopted by Chiari, Rocchi, and Cappello (2002) and

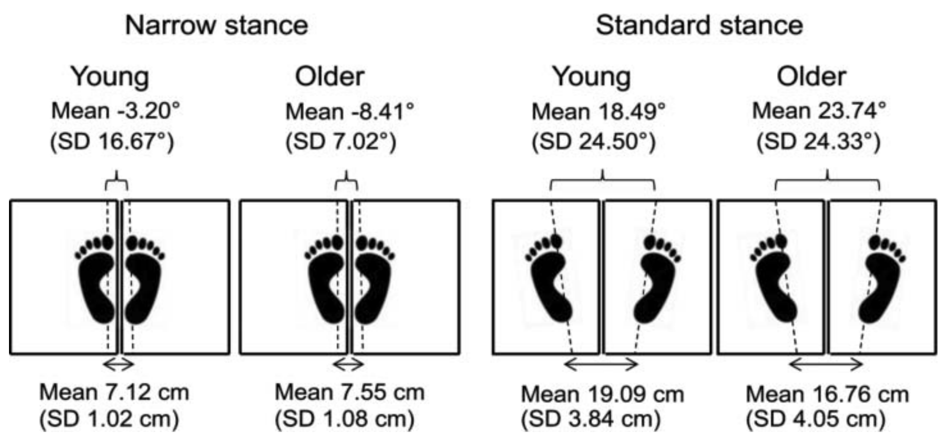

FIGURE 6. Representation of the foot positions adopted by the participants in the two stance width conditions (narrow and standard stances). The mean values and standard deviations of the stance width and stance angle are given for each stance condition.

recommended by O'Malley (1996). This procedure consists of removing the influence of a confounding variable without changing the units and range of the data. These variables were controlled in each age group separately. The main analyses (ANOVAs) were redone with the normalized variables.

The controlled analyses provided two new significant findings. After controlling for the age difference in each group, the group by stance width interaction effect was significant for the standard deviation of COP displacement (young adults standard stance: 0.15 ± 0.04 cm, narrow stance: 0.55 ± 0.19 cm; middle-aged adults standard stance: 0.15 ± 0.05 cm, narrow stance: 0.42 ± 0.08 cm). After controlling for height, a main effect of age was found for the velocity of COP displacement (young adults: 1.57 ± 0.18 cm.s^{-1}; middle-aged adults: 1.43 ± 0.25 cm.s^{-1}). Otherwise, most of these results showed a slightly greater partial eta squared with the normalized variables. Importantly, the main effect of age in the ANOVA for the cross-correlation COP_c versus COP_{net} was much stronger for each controlled variable than before the normalization ($.45 < \eta_p^2 < .46$). Overall therefore, the foot position and physical characteristics of the participants were confounding variables in the sense that they limited the amount of significant findings and the strength of the results.

Discussion

Our study featured four main findings. First, the active contribution of ML postural control mechanisms at the ankle is significantly lower in middle-aged (i.e., nonelderly) adults than in young adults. Second, the role of ML ankle postural control mechanism is much greater than has previously been found—especially for narrow stance. Third, narrow stance increased the overall contribution of the two mechanisms, but not as much as expected. Fourth, the different mandibular positions and clenching conditions significantly changed the characteristics of COP displacement but not the contributions of COP_c and COP_v.

Age-Related Changes in Postural Control Mechanisms

The study results confirmed our expectation that the ankle mechanism contributes significantly less to ML COP displacement control in middle-aged adults than it does in young adults. Indeed, middle-aged adults exhibited significantly lower ML COP_c active contribution than young adults did (Figures 3A and 3B, Table 1). In other terms, the result means that the active contribution of the ankle mechanism was less efficient in middle-aged adults than in young adults. Furthermore, the active contribution of ML COP_c was null in middle-aged adults but significantly positive in young adults (Figures 3A and 3B, Table 1). It means that the ankle mechanism had no significant role in ML postural control in middle-aged adults in standard stance while it did have a significant role in young adults. These two effects significantly diminish the collaboration between ML COP_v and ML COP_c seen in middle-aged adults, relative to young adults (Figures 3A and 3B, Table 1). This is a striking finding because it relates to a main effect of age and was not solely observed in ML-challenging conditions (Figures 3A and 3B, Table 1). Also, the effect was much stronger after controlling for foot positions and/or physical characteristics of the participants (see controlled analyses, end of Results). Our healthy, middle-aged participants may thus have shown preclinical signs of future ML postural instability. Indeed, impairments in postural control mechanisms should subsequently translate into increased postural sway (Maurer & Peterka, 2005). Our results differed from those reported by Termoz et al. (2008) probably because we did not instruct our participants to remain as steady as possible during the trials. Thus, when seeking to understand age-related or disease-related increases in ML sway, it may be more appropriate to test participants under natural conditions.

Age-Related Changes in COP Displacement

Middle-aged adults did not exhibit greater or more rapid ML COP displacement than young adults did under any of our experimental conditions (Figures 4A, 4B, and 4C, Table 1). This result was expected, as middle-aged adults are usually found to sway either as much as young adults do or significantly slightly more (Abrahamova & Hlavacka, 2008; Era et al., 2006). However, our analyses showed that middle-aged adults exhibited a significantly lower range of ML COP displacement than young adults did (Figure 4B, Table 1). As exemplified by Figure 4B, middle-aged adults did not increase their range of COP displacement in narrow stance, whereas young adults clearly did. This result was unexpected, since previous studies mostly found that COP displacement and/or postural sway were greater in narrow stance than in wider stance (Day et al., 1993; Kirby et al., 1987; Mouzat et al., 2004). Our other results (standard deviation and the mean velocity of COP displacement) were in line with literature findings (Figures 4A and 4C, Table 1). We therefore suppose that middle-aged adults controlled their narrow stance in an unexpected manner by constraining the maximum amplitude of their ML oscillation. In this difficult stance condition, they may have increased their ankle stiffness as an alternative control mechanism to compensate for putative impairments in the ML ankle postural control mechanism (Figure 3A, Table 1). This interpretation is supported by Benjuya, Melzer, and Kaplanski (2004), who showed an age-related increase in ankle muscle cocontraction (i.e., greater electromyography activities of the peripheral muscles) in narrow stances. Overall, our study results cannot confirm our initial hypothesis whereby impairments in ML COP_c appear earlier in life than any changes in ML COP displacement. However, we found that healthy adults under the age of 60 can have a significant impairment in ML postural control.

Effects of Stance Width on the Postural Control Mechanisms

For young adults in standard stance, the results were consistent with those reported by Termoz et al. (2008) and Gatev et al. (1999; i.e., significant contributions of both ankle and hip mechanisms to the control of ML COP displacement (Figure 3A, Table 1). In the comparison between standard and narrow stances, Figures 3A and 3B showed that the active contribution was significantly greater in narrow stance than in standard stance. This finding was expected based on the study by Gatev et al. However, the findings for the RMS amplitude of the mechanisms were not anticipated. Indeed, Figure 2 showed that the amplitude contribution of the mechanisms was significantly lower in narrow stance than in standard stance. In comparison, the findings for the active contribution were stronger ($\eta_p^2 > .38$) than the findings for the amplitude contribution ($\eta_p^2 < .16$), thus confirming that the overall contribution of ML COP_v and ML COP_c were higher in narrow stance than in standard stance. In other words, the two mechanisms were more active to modify the position of COP displacement in narrow stance but did it with less strength. Therefore, the overall contribution of postural mechanisms to controlling COP displacement was weakly increased, probably explaining why individuals sway more in narrow stance than in standard stance (e.g., Day et al., 1993; Kirby et al., 1987; Mouzat et al., 2004). In narrow stance, the amplitude contribution may have been weaker because the two reaction forces (one under each foot) were closer to each other, thus shortening the lever arm to control ML postural sway. Indeed, Winter et al. (1996) explained that the wider the stance width is, the less muscle activation required to maintain the same COP displacement.

In Gatev et al. (1999), seven healthy men stood in narrow and standard stance conditions with their eyes open or closed. In narrow stance, Gatev et al.'s analyses showed (a) less electromyography activity in the lower leg, (b) more relationships between linear and angular motions at the hip, (c) less correlation of body motions throughout the body, and (d) a significant correlation between hip angular motions in the AP and ML axis. Based on these findings, the researchers concluded that the hip mechanism contributed more to the control of narrow stance than to the control of standard stance. We additionally showed an unpublished finding that the collaboration between COP_v and COP_c is significantly greater in narrow stance than in standard stance (Figures 3A and B, Table 1). As expected also, the difference in the active contribution of COP_c between standard and narrow stances was much greater than the difference for COP_v in these conditions (effect size in the ANOVAs = .48 and .38, respectively). It is noteworthy that in narrow stance, the active contribution of COP_c to ML postural control was almost as great as the active contribution of COP_v (Figures 3A and 3B, Table 1). Because elderly adults have been shown to naturally adopt a significantly narrower quiet stance than young adults (McIlroy & Maki, 1997), they may have more difficulty controlling ML COP displacement in natural stance. This is an important practical message, given the very high F value found for the main effect of stance width on COP_c, $F(1, 22) = 226.14$.

Effects of Mandibular Position and Clenching on COP Displacement and Mechanisms

Some studies have shown that mandibular conditions can significantly change the area of COP displacement (Gangloff, Louis, & Perrin, 2000; Gangloff & Perrin, 2002). Our analyses with COP displacement variables did not confirm these findings (at least for the ML axis) because there was no significant effect of mandibular or clenching laterality. Although the participants exhibited significantly more rapid ML COP displacement in clenching conditions than in mandibular position conditions (Figure 5), this effect was probably meaningless. Indeed, it was not clearly apparent on Figure 5 and the effect size was low. By clenching the teeth, the mechanical increased tension in the fascia system may have spread to increase the participants' overall body stiffness, which in turn would have increased their COP displacement velocity. This increase in body stiffness must have been slight because it did not affect COP SD, COP range, or the contributions of the two postural control mechanisms (COP_c and COP_v). Overall, the mandibular conditions only had a marginal effect on postural control, even in narrow stance.

Summary and Future Work

We are not aware of any literature reports of age-related impairments in ankle and hip postural control mechanisms that could explain the known, age-related increases in ML COP amplitude and velocity (Maki et al., 1994; Prieto et al., 1996). In the present study, we expected to find and indeed observed an age-related impairment in the ML ankle postural control mechanism in under 60-year-old adults (with no changes in the ML hip mechanism). In practical terms, our study demonstrates that ML postural control may already be compromised before age 60 years—even though healthy middle-aged individuals did not exhibit significantly greater COP displacement than young adults. Our study is limited in the sense that the neuromuscular control at the ankle and hip is unknown. However, based on our findings, future researchers should be directed to search for physiological factors that may explain the lower age-related active contribution of ML COP_c: these may include a lower threshold for sensing passive inversion or eversion of the ankle (Gilsing et al., 1995) and a general reduction in the somatosensory threshold at the feet, ankles and legs (Menz, Morris, & Lord, 2005, 2006; Scott, Menz, & Newcombe, 2007; Toledo & Barela, 2010). Measures of these physiological factors as well as measures of electrophysiological factors (e.g., nerve conduction velocity of lower leg muscles) will be relevant to better explain normal and abnormal neuromuscular contributions of COP_c and COP_v to control stance. Future researchers' work and practical efforts should also check whether improving the motor performance or sensitivity of the inversion or eversion mechanism increases the contribution of COP_c and reduces ML falls (which are significantly related to hip fractures; Hayes et al., 1996; Rogers & Mille, 2003). Based on our present results, it may be important to teach elderly and middle-aged adults to increase their stance width to compensate for the lower COP_c active contribution. Alternatively, older adults could be taught to regularly check their ankle function on the sensory and motor levels.

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