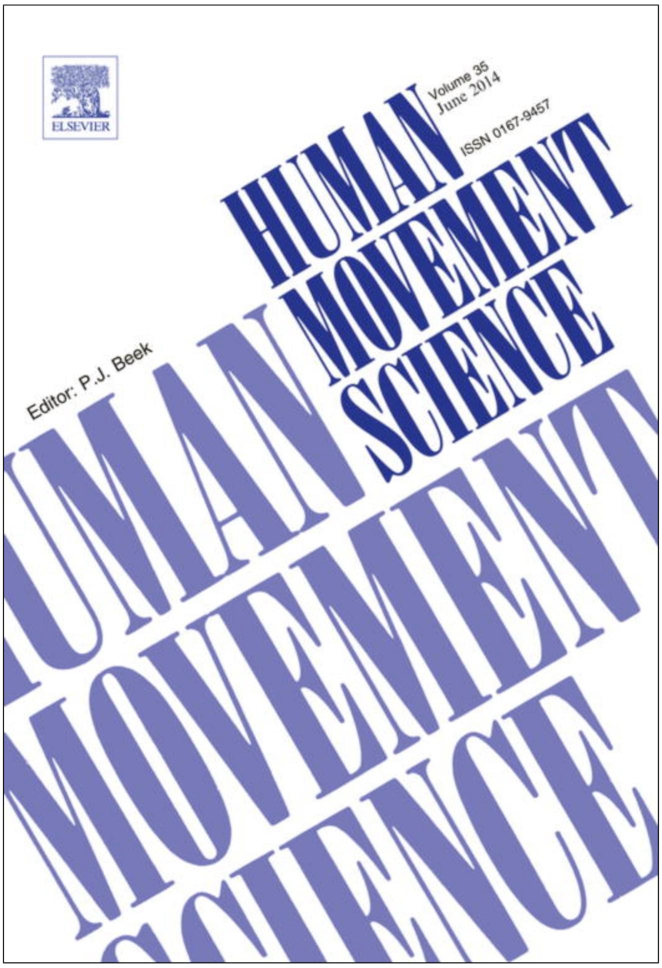
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Methodological requirement to analyze biomechanical postural control mechanisms with two platforms

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A B S T R A C T

In 1996, Winter and colleagues proposed the existence of two postural control mechanisms in both the anteroposterior and mediolateral axes: a bodyweight (loading/unloading) distribution mechanism and a complementary center of pressure location mechanism. To measure the loading/unloading forces under each foot, the feet had to be placed side by side in the mediolateral axis and one foot ahead of the other in the anteroposterior axis. Our first objective was to reexamine the validity of anteroposterior data published with the feet side by side. In that foot condition, we expected no change in the anteroposterior loading/unloading forces (regardless of the task performed), and consequently no change in the complementary mechanism. Our second objective was to confirm our hypotheses with experimental data. Twelve healthy, young adults performed three types of body oscillation in the anteroposterior axis (at the hips, at the ankles and alternately at the ankles and hips) and a quiet stance condition with the feet side by side. As expected, the bodyweight mechanism did not vary significantly. Although the complementary mechanism was significantly higher in the ankle and alternating conditions, the change was very tiny (<0.3%). Thus, we propose methodological requirements to analyze both mechanisms.

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1. Introduction

In stance, individuals sway all the time (Winter, 1995) but rarely fall. Since the mid-19th century, researchers have sought to understand how postural control operates, with a view to improving quality of life, preventing falls and creating humanoid robots. Two decades ago, Winter and colleagues (Winter, Prince, Frank, Powell, & Zabjek, 1996; Winter, Prince, Stergiou, & Powell, 1993) showed the existence of a mediolateral (ML) loading/unloading mode of coordination consisting of loading the bodyweight under one foot and thus unloading the bodyweight under the other. This mechanism seemed to act at the proximal level by using lateral hip muscles (Winter et al., 1993, 1996). This postural mechanism can be referred to as the bodyweight distribution mechanism or the loading/unloading distribution mechanism. The other ML mechanism, the center of pressure (COP) location mechanism, was assumed to act at the ankle level through inversion/eversion. This mechanism is performed by changing the COP location under the left and right foot. Both bodyweight distribution and COP location mechanisms are necessary and complementary in explaining different proportions of COP displacement (Winter et al., 1993, 1996; cf. equations in the Method).

In order to measure ML loading/unloading forces, Winter et al. (1993, 1996) clearly explained the need to have a ML foot-platform pair, that is two feet and two platforms side by side (Fig. 1A). With only one platform, it would not be possible to measure two vertical reaction forces and two center of pressure displacements, one under each foot. Winter et al. (1996) extended the existence of a loading/unloading mechanism to the anteroposterior (AP) axis. Logically, the AP loading/unloading mechanism has to be measured separately, using another foot-platform pair (forward and backward platforms and the feet in the Tandem Romberg position; Fig. 1B; cf. Winter et al., 1996). Therefore, based on these theoretical arguments, the ML loading/unloading of each foot cannot be measured in the Tandem Romberg (TR) foot position. Similarly, the AP loading/unloading forces under each foot cannot be measured with the feet side by side. In other words, to measure the loading/unloading forces under each foot, the feet need to be side by side in the ML axis (Fig. 1A) and one foot ahead of the other in the AP axis (Fig. 1B). Therefore, if both the ML and the AP COP loading/unloading forces have to be calculated accurately in the same trial, the only possible foot position may be the semi-tandem or 45° condition (Fig. 1C) with no overlap between the feet. For all these reasons, we do not understand why Lafond, Corriveau, and Prince (2004), Rougier (2007, 2008), Termoz et al. (2008) and Winter et al. (1993, 1996) calculated the AP contribution of the bodyweight distribution mechanism with the feet side by side (Fig. 1A). How can the loading of the forward foot be measured separately from the unloading of the backward foot with no foot ahead of the other? What are the "forward" foot-platform and "backward" foot-platform pairs in that side-by-side condition?

In the present study, we did not contest the validity of Winter et al.'s (1993, 1996) bodyweight distribution and COP location mechanisms. Our primary objective was to question the validity of AP loading/unloading results published with the feet side by side and to discuss a relevant methodology to measure and compute these mechanisms. Our primary hypothesis was that the computation of AP

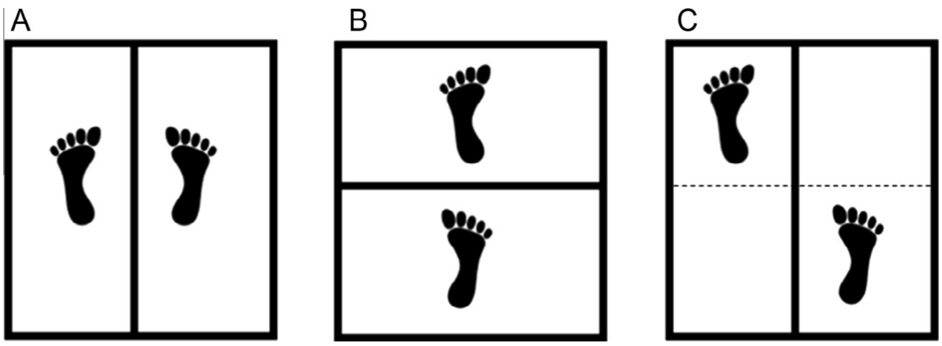

Fig. 1. (A) Representation of the foot position on each of two adjacent force platforms; (B) representation of the foot position with one platform forward and one platform backward; (C) representation of the 45° condition in which one foot is strictly ahead of the other and one foot is strictly to the right of the other. The feet can be placed on two (or four) adjacent force platforms.

loading/unloading with the feet side by side may not be relevant. In that foot condition, the platforms can measure the loading of the forward parts of both feet and the unloading of the backward parts of both feet, in some ways a half loading/unloading mechanism, which is not what Winter et al. (1996) meant by loading/unloading mechanism. Our secondary objective was to give further support to our primary hypothesis with experimental data. Indeed, a theoretical discussion alone may be insufficient to contest the experimental methodology used by several authors (Lafond et al., 2004; Rougier, 2007, 2008; Termoz et al., 2008; Winter et al., 1993, 1996).

In the feet-side-by-side condition, AP loading/unloading forces should be almost constant all the time in any kind of experimental condition because there is no foot ahead. Also, if the AP loading/unloading forces do not change in different conditions, then the proportion of the AP COP displacement explained by the other mechanism – the COP location mechanism – should not change either. Indeed, Winter et al. (1993, 1996) calculated the contribution of the COP location mechanism by eliminating the contribution of the COP loading/unloading mechanism (cf. Method and equations). Therefore, overall, the contribution of both AP mechanisms may not change with the feet side by side, even in extremely different experimental conditions. Our experiment tested these sub-hypotheses in order to give more power to our theoretical argument. To do so, we decided to modify not just one static constraint (e.g., changing the distance between the feet side by side), but several. We changed the body movement constraint (voluntary body oscillations vs. no motion), the type of body coordination used and as a consequence the amplitude of body motions. Four experimental conditions were studied in twelve healthy, young adults: quiet stance (the control condition), oscillating back and forth at the hips (the hip condition), at the ankles (the ankle condition), and at the ankles and hips at different times (the alternating condition). During the trials, the participants stood on a double force platform with the feet side by side, and one foot on each platform (Fig. 1A). The absence of any significant difference in both mechanisms in these four conditions would confirm our sub-hypotheses. In this case, both theoretical and experimental data would show that the computation of the two AP mechanisms may not be methodologically usable with the feet side by side, contrary to what is assumed in the literature (Lafond et al., 2004; Rougier, 2007, 2008; Termoz et al., 2008; Winter et al., 1993, 1996). In contrast, significant changes in any mechanism would cast doubt on our primary hypothesis (although biomechanically relevant).

2. Methods

2.1. Participants

Twelve volunteer students (four males and eight females) from the University of Lille 2 were included in the study following receipt of their written, informed consent to participate. The study was performed in accordance with the tenets of the Declaration of Helsinki. All of the participants were in good general health. Volunteers were excluded if they had any known ankle or hip problems. Their mean ± SD age, bodyweight and height were 23.00 ± 3.72 years, 65.92 ± 9.76 kg and 1.70 ± 0.10 m, respectively.

2.2. Conditions

In the quiet stance condition, the participants were told to stand comfortably. In the three other conditions, the participants were required to perform different types of AP body oscillations, changing their direction of movement in time with a metronome. The metronome was empirically set to 0.5 Hz to allow clear and feasible oscillations in the three conditions. In the hip condition, the participants had to oscillate intentionally their upper body only at their hips. In the ankle condition, the participants were instructed to oscillate intentionally back and forth only at their ankles. In the alternating condition, the participants had to oscillate intentionally their lower and upper body segments in a four-movement sequence with two movements per metronome sound signal: (i) forward, at the ankles only, (ii) further forward, at the hips only, (iii) backward, at the ankles only and (iv) further backward, at the hips only. Under all these conditions, head motions had to be continuous, broad and constant in amplitude. The goal of performing these three oscillatory conditions was simply to modify the constraint of actively moving the body (motion vs. no motion), the kind of body coordination (upper and lower body parts moving in phase, in anti-phase and in phase delay) and the amplitude of body motions (AP and vertical head motions changed greatly with the three conditions). The oscillatory conditions were also performed at the ankles and hips because the muscles at these levels are assumed to change the contribution of both the bodyweight distribution and COP location mechanisms (e.g., Winter et al., 1996). However, these conditions or body oscillations had no reference to Nashner and McCollum's (1985) ankle and hip strategies. Indeed, the AP ankle and hip postural strategies are not equivalent to, nor do they refer to, the AP ankle and hip mechanisms by Winter et al. (1993, 1996). Moreover, we did not ask our participants to perform these ankle and hip strategies and we did not compute our data to show which strategy was used.

The participants were barefoot and told to keep their feet in full contact with the force platforms throughout the trials. The participants looked at a black dot on a wall 1.90 m in front of them and held their hands behind their back. A standard stance width and a standard stance angle were adopted for all trials, 17 cm and 14°, respectively (McIlroy & Maki, 1997). There were three trials per condition, with each lasting 75 s. The order of performance of the conditions was randomized.

2.3. Apparatus

A dual-top force platform (AMTI, Watertown, MA) was used to record COP displacement with a sampling frequency of 120 Hz. A two-camera motion analysis system (Version 7.5 from Simi Reality Motion Systems GmbH, Munchen) was used to record marker displacements (marker diameter: 2.5 cm). The cameras were directed at the participant's right side and the acquisition frequency was set to 15 Hz. The markers were attached to the outer edge of the tibiofemoral joint, the anterosuperior iliac spine and the outer edge of the biceps (on the right side only). To this end, the participants wore (i) a fabric knee support (S-100 Supportiv, size 2, 35–37 cm, Oxylane, China, Shanghai) over their trousers, a belt at the anterosuperior iliac spine level, and an elbow strap (S-300 Supportiv, Oxylane) around the uppermost part of their right biceps. The knee support also helped the participants to limit their right knee motion.

2.4. Variables and analyses

We used Winter et al.'s (1996) model to investigate the contribution of the bodyweight distribution and COP location mechanisms. First, we used three equations to calculate three time series: (1) the resultant COP displacement ($\mathrm{COP}_{\mathrm{net}}$), (2) the COP displacement explained by the COP location mechanism (denoted as COP_c in the model calculation, c for changes) and (3) the COP displacement explained by the bodyweight distribution mechanism (denoted as COP_v in the model calculation, v for vertical):

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

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

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

In Eq. (1), $\mathrm{COP}_{\mathrm{net}}$ was calculated as the sum of COP displacement under each foot ($\mathrm{COP}_l(t)$ and $\mathrm{COP}_r(t)$ for the left and right feet, respectively), while taking into account the weight under each foot. $R_{vl}(t)$ and $R_{vr}(t)$ are the vertical reaction forces under the left and right feet, respectively. In Eq. (2), the contribution of COP_c was calculated by eliminating the contribution of COP_v (constant 50% of bodyweight throughout the trial). In Eq. (3), the contribution of COP_v equated to the contribution that could not be explained by COP_c. It should be noted that $\mathrm{COP}_c(t)$ and $\mathrm{COP}_v(t)$ are not real but only simulated data (no one can keep 50% of bodyweight constant throughout a trial).

Once the three time series were obtained, two complementary analyses were performed to analyze the contribution of each mechanism. The first analysis compared the amplitude of $\mathrm{COP}_{\mathrm{net}}$ and COP_c

and of COP_{net} and COP_v time series to compute the amplitude contribution of each mechanism. The root mean square (RMS) COP_c, RMS COP_v and RMS COP_{net} were computed according to the literature (e.g., Lafond et al., 2004; Termoz et al., 2008; Winter et al., 1996). If the variability of COP_c or COP_v was similar to that of COP_{net}, it was assumed that the mechanism had a high amplitude contribution to explain COP_{net}. In that case, the strength of the mechanism was high. The cross-correlations for COP_c vs. COP_{net}, COP_v vs. COP_{net} and COP_v vs. COP_c were also computed according to the literature (e.g., Lafond et al., 2004; Termoz et al., 2008; Winter et al., 1996). As in Bonnet, Mercier and Szaffarczyk (2013), we assumed that these analyses were concerned with the active contribution of the two mechanisms. Indeed, cross-correlation analyses are not concerned with the amplitude of the time series but with the direction and proportionality of the time series, thus showing a different aspect than the amplitude contribution. A high cross-correlation coefficient was assumed to show that the mechanism has a high active contribution to control COP_{net} (cf., Bonnet et al., 2013). All analyses were performed in the AP axis only. In order to exclude irregular motions, the first 10 s and last 5 s of each trial were not analyzed.

One-way analyses of variance (ANOVAs) were performed on the various dependent variables (i.e., the mean per condition). These were followed by post hoc Newman–Keuls tests. For all analyses, the threshold for statistical significance was set to $p < 0.01$.

In our study, the marker displacements were recorded only to characterize body motions in the different conditions. These data served to show that our participants moved their body parts differently in the four conditions. The participants' body motions were characterized by analyzing the angular and linear displacements of the knee, hip and shoulder markers. The Simi software directly computed the angles formed by the knee–hip and hip–shoulder segments relative to the vertical axis. There were 30 cycles per trial in the ankle, hip and alternating conditions (body motions performed at 0.5 Hz and analyzed during 1 min). The 15 forward peaks and 15 backward peaks per trial were extracted and the oscillation amplitude of each cycle was calculated and then averaged across cycles. The 30 linear forward–backward and 30 vertical oscillation amplitude of the knee, hip and shoulder were computed in the same way (i.e., in extracting and then averaging the 15 highest and 15 lowest values).

3. Results

3.1. Postural control mechanisms

There was a significant main effect of condition for AP RMS COP_c and AP RMS COP_{net} ($F_s(3,33) > 7.60$, $\eta_p^2 > 0.29$, $p < 0.01$; Fig. 2) but not for AP RMS COP_v ($F(3,33) = 1.25$, $p > 0.01$). The RMS COP_c and RMS COP_{net} were significantly lower in the quiet stance condition than in the ankle and alternating conditions ($p < 0.01$; Fig. 2).

The ANOVA showed a significant effect of condition for the COP_{net} vs. COP_c cross-correlation ($F(3,33) = 10$, η_p^2 0.32, $p < 0.01$; Fig. 3). The coefficients in the ankle and alternating conditions were significantly higher than those in the hip and quiet stance conditions ($p < 0.01$). The result of the ANOVA was not significant for the COP_{net} vs. COP_v and COP_v vs. COP_c cross-correlations ($F_s(3,33) < 0.38$, $p > 0.01$; Fig. 3).

3.2. Characteristics (amplitude, phase) of body motion in the three oscillatory conditions

In the ankle condition, the lower and upper body segments were almost aligned (mean knee–hip angle: 7.30°; mean hip–shoulder angle: 9.15°). Backward movements were less broad (mean knee–hip angle: −2.33°) than forward movements (mean hip–shoulder angle: 9.70°) (Table 1). Both knee–hip and hip–shoulder angles moved in phase (Fig. 4). The amplitude of these oscillations was situated between the maximum backward (−5°) and forward (12°) static body inclinations described by McCollum and Leen (1989). There were three main differences between the alternating and ankle conditions: the threefold greater mean hip–shoulder angle (Table 1), the threefold greater vertical displacement at the shoulder (Table 2) and a slight phase delay in knee–hip and hip–shoulder angles in the alternating condition (Fig. 4). The head moved forward further than the legs and thus further

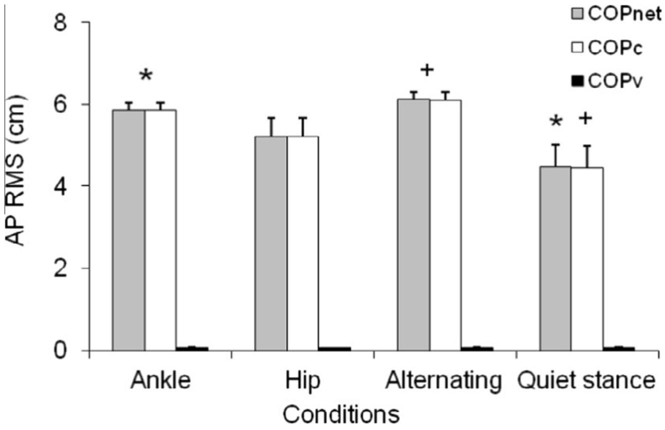

Fig. 2. The root mean square (RMS) amplitude of COP_{net}, COP_v and COP_c in the anteroposterior (AP) axis. The mean values of the RMS amplitudes and their standard errors (error bars) are represented for each of the four experimental conditions: ankle, hip and alternating oscillations and quiet stance (see the text for further details). COP_{net} is the resultant displacement of the center of pressure (COP) under the left and right feet. COP_v is the COP displacement under the control of the loading/unloading mechanism. COP_c is the COP displacement under the control of the plantarflexion/dorsiflexion mechanism. * Indicates a significant difference between the ankle condition and the quiet stance condition for both COP_{net} and COP_c. + Indicates a significant difference between the alternating condition and the quiet stance condition for both COP_{net} and COP_c. $p < 0.01$.

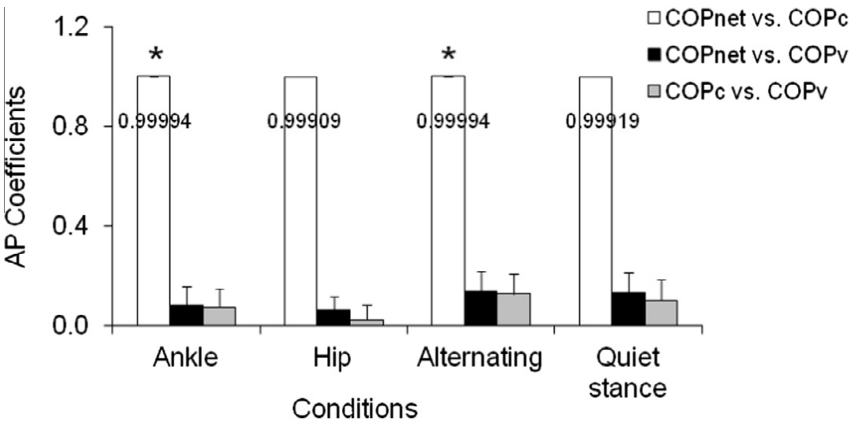

Fig. 3. The cross-correlation coefficients in the anteroposterior (AP) axis for COP_{net} vs. COP_v, COP_{net} vs. COP_c and COP_v vs. COP_c. The mean values of the cross-correlation coefficients and their standard errors (error bars) are represented for each of the four experimental conditions: ankle, hip and alternating oscillations and quiet stance (see the text for further details). For the definitions of COP_{net}, COP_v and COP_c, see the caption for Fig. 2. * Indicates a condition in which the cross-correlation coefficient (COP_c here) was significantly higher than in the conditions with no *. For clarity's sake, the COP_c mean values are indicated on the figure. $p < 0.01$.

Table 1
Angular displacements (in degrees) of two segments (knee–hip and hip–shoulder) in the three oscillatory conditions (ankle, hip and alternating). The min and max angles correspond to the mean of the 15 minimum and 15 maximum oscillation amplitude angles formed by the segment and the vertical axis during the oscillatory conditions. The values reported in the table correspond to the range (the difference between the maximum and the minimum). The standard deviation of the mean is given in brackets. For definitions of the three conditions, please refer to the text.

	Ankle			Hip			Alternating		
	Min	Max	Mean	Min	Max	Mean	Min	Max	Mean
Knee–hip angle (degrees)	−2.23 (±0.94)	5.07 (±0.98)	7.30 (±1.25)	−2.57 (±1.19)	1.59 (±0.87)	4.16 (±2.41)	−3.21 (±0.78)	3.74 (±0.77)	6.95 (±0.93)
Hip–shoulder angle (degrees)	0.55 (±1.61)	9.70 (±2.14)	9.15 (±3.06)	2.31 (±2.02)	53.28 (±2.78)	50.97 (±8.33)	1.75 (±1.34)	29.72 (±2.27)	27.97 (±7.78)

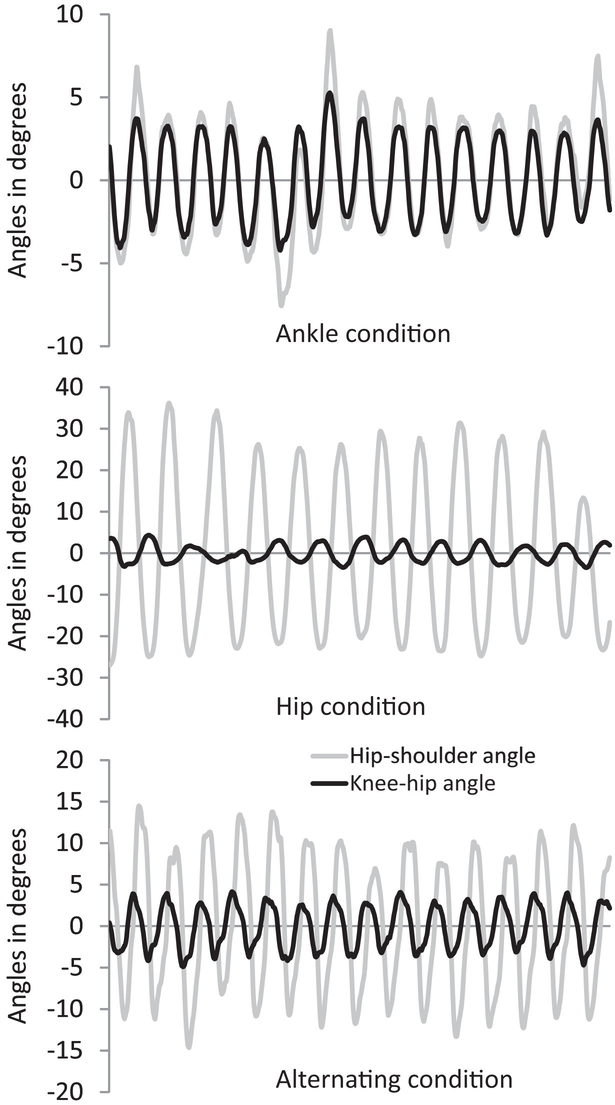

Fig. 4. Representative data of the angles formed by the knee–hip and hip–shoulder segments (relative to the vertical axis) in the ankle, hip and alternating conditions. The participants were asked to oscillate back and forth at 0.5 Hz at the ankles (the ankle condition), at the hips (the hip condition), and at the ankles and hips at different times (the alternating condition) (cf. Method for more details).

Table 2
Anteroposterior and vertical linear displacements (in centimetres) of the three markers (knee, hip and shoulder) in the three oscillatory conditions (ankle, hip and alternating). The 15 minimum displacements and the 15 maximum displacements per trial were extracted and averaged. The standard deviation of the mean is given in brackets. For definitions of the three conditions, please refer to the text.

In centimetres	Ankle	Hip	Alternating
Knee AP displacement	4.66 (±0.81)	7.87 (±3.23)	6.77 (±2.33)
Hip AP displacement	11.20 (±1.66)	9.79 (±5.44)	12.07 (±2.16)
Shoulder AP displacement	15.21 (±2.50)	11.47 (±4.41)	18.21 (±2.94)
Knee vertical displacement	0.61 (±0.21)	0.66 (±0.21)	0.79 (±0.29)
Hip vertical displacement	1.23 (±0.36)	1.58 (±0.91)	1.22 (±0.43)
Shoulder vertical displacement	1.64 (±0.71)	13.30 (±4.25)	4.82 (±2.08)

downwards in phase 2, as requested by the experimenter (see the Methods section). In the hip condition, the vertical and horizontal displacements of the shoulder were of the same magnitude (Table 2). The participants mostly did not oscillate around the lower body segment (mean knee–hip angle:

4.16°), but around the higher body segment (mean hip–shoulder angle: 50.97°), as requested by the experimenter. Also, both the knee–hip and hip–shoulder angles moved in anti-phase (Fig. 4), as can be expected to keep the center of mass relatively stable. Indeed, the lower body part should move backward – not forward – to compensate for the broad forward motion of the upper body.

Overall, the participants performed the three oscillatory conditions with different amplitude and phase. First, the two segments (knee–hip and hip–shoulder) moved differently from one another (ankle condition: in-phase; hip condition: anti-phase; alternating condition: phase delay with the knee–hip segment leading the hip–shoulder segment). Second, the mean angular displacement of the hip–shoulder segment (Table 1) and the vertical displacement of the shoulder were clearly different in the three conditions (Table 2).

3.3. Control analyses

To control for the influence of age, height and weight on COP_c and COP_v, we used the detrending normalization procedure recommended by O'Malley (1997) and adopted by Chiari, Rocchi, and Cappello (2002). This consists in removing the influence of a potentially confounding variable (here age, height and weight individually) in a time series. Normalization did not change the significance of the RMS amplitude and cross-correlation analyses but strengthened the results for RMS COP_{net} and RMS COP_c slightly ($0.30 > n_p^2 > 0.35$ instead of $n_p^2 = 0.29$). Therefore, the participants' physical characteristics were not confounding variables.

4. Discussion

In the present study, we assumed that Winter et al.'s (1996) postural control theory should not be used in the AP axis with the feet placed side by side (Fig. 1A). The experimental data were consistent with this hypothesis. Hence, we discuss methodological requirements for the feet-platform locations in order to enable relevant measurements and analyses of the bodyweight distribution and COP location mechanisms. This methodological requirement could allow the discovery of age-related and disease-related deficiencies in AP COP_v and AP COP_c in future studies.

4.1. COP_v in the AP axis

The AP RMS COP_v was very low in all conditions (Fig. 2) and COP_v did not differ significantly in the four experimental conditions. Additionally, COP_v did not interact with COP_c in any of the conditions (Fig. 3). Therefore, overall, the different kinds of AP postural coordination (Fig. 4; mean hip–shoulder angles in Table 1 and mean shoulder vertical displacement in Table 2) did not change AP COP_v when the feet were side by side. These results were expected. The simple biomechanical reason is that AP COP_v was not measured since one foot was not positioned ahead of the other. In this particular feet-side-by-side condition, the platforms measured the loading of the forward parts of both feet and the unloading of the backward parts of both feet, that is, half loading/half unloading forces under both feet. It did not measure the full loading/full unloading forces under a forward foot and a backward foot.

4.2. COP_c in the AP axis

Our data showed significantly greater values for COP_c vs. COP_{net} in the ankle and alternating conditions when compared with the hip and quiet stance conditions (Fig. 3). The data also showed significantly greater values for RMS COP_c in the ankle and alternative conditions when compared with the quiet stance condition (Fig. 2). Thus, the results potentially showed that the two platforms side by side could measure the action of plantarflexion/dorsiflexion in moving the body in the AP axis. However, the inter-condition differences in COP_c vs. COP_{net} and RMS COP_c were not meaningful in practical terms. Indeed, the COP_c vs. COP_{net} coefficients (ankle: 1.00; hip: 1.00; alternating: 1.00; quiet stance: 1.00) and the ratio RMS COP_c/RMS COP_{net} (ankle: 99.94%; hip: 99.79%; alternating: 99.85%; quiet stance: 99.56%) were almost identical in the four conditions. According to the literature, the ratio between hip and ankle motions is 4:1 when the subject bends the trunk at 45° (Alexandrov, Frolov, & Massion, 2001). Similar ratios should have been found in our study in the hip condition because the hip–shoulder angle was 50.97° on average (Table 1). Therefore, overall, we conclude that with the feet side by side, very different kinds of AP postural coordination did not change AP COP_c. This finding is logical because Winter et al. (1993, 1996) calculated the contribution of COP_c by eliminating the contribution of COP_v – unchanged in our study – to explain the COP displacement.

4.3. Methodological requirements and conclusion

The present study confirmed our theoretical hypothesis that AP loading/unloading and AP plantarflexion/dorsiflexion mechanisms cannot be measured and differentiated with the feet positioned side by side. Therefore, we question the validity of the results published in the AP axis with the feet placed side by side (Lafond et al., 2004; Rougier, 2007, 2008; Termoz et al., 2008; Winter et al., 1993, 1996). For the very same reason, we question the validity of ML COP_v and ML COP_c reported in the TR condition (i.e., Winter et al., 1996). Indeed, there is no foot to the right – or left – of the other foot in that TR condition.

In future studies, we recommend measuring and calculating (i) ML COP_v and ML COP_c only in conditions with the feet strictly to the left/right of each other and placed on separate platforms (Fig. 1A), regardless of the distance between the feet and their AP positions with respect to each other (for example one foot can be ahead of the other); and (ii) AP COP_v and AP COP_c only in conditions with one foot strictly ahead of the other and on separate platforms (Fig. 1B), regardless of the distance between the feet and their ML positions with respect to each other. If both ML and AP COP_v and COP_c need to be computed together (iii), we recommend performing conditions with the feet both strictly to the left/right and strictly behind/ahead of each other on two or four platforms (Fig. 1C), regardless of the AP and ML distances between the feet. In other foot positions, it may not be possible to dissociate which foot loads and which foot unloads in both the AP and ML axes. As such, the methodology used by Termoz et al. (2008) in their 45° condition is not perfect. Indeed, these authors measured and calculated ML and AP COP_v and COP_c with the forward foot only 80% ahead of the backward foot. The forward foot had to be entirely ahead of the backward foot, as in Winter et al. (1996; 110%).

Using four platforms instead of two may not solve the problem discussed in the present manuscript. Indeed, if the feet are side by side, the AP loading of one foot forward may still not be dissociated from the AP unloading of the foot backward because the feet overlap (as we discussed and showed). A line has to separate the two feet (Fig. 1C). In the very same way, if the feet are in the Tandem Romberg condition, the loading of the 'left' foot may not be dissociated from the unloading of the 'right' foot simply because the feet overlap (the names 'left foot' and 'right foot' are nonsensical in this condition). Hence, the 45° condition is the only condition in which the unified theory of postural control (Winter et al., 1996) can be applied to measure and study ML and AP COP_v and COP_c; but here two platforms are still sufficient if the feet do not overlap in both the ML and AP axes.

In conclusion, we did not contest the validity of Winter et al.'s (1993, 1996) bodyweight distribution and COP location mechanisms in both AP and ML axes. We simply explained that AP COP_v and AP COP_c may not be meaningful in conditions where the feet are placed side by side. We even assumed that AP COP_v and AP COP_c may not change at all in different conditions. Consistently, we found (almost) no change in COP_v and COP_c between the quiet stance and three kinds of broad body oscillations (at the ankle, at the hip and alternating at the ankle and hip). One criticism of our study may be that we did not need experimental data or possibly that the experimental conditions were not the most suitable. Indeed, Winter et al. (1993, 1996) did not notice that COP_v and COP_c could specifically control body oscillations at the hip and ankle respectively. Future conditions may still show significant changes in AP COP_v and AP COP_c even with the feet side by side. However, this would be difficult to explain at the biomechanical and physiological levels. Indeed, one can understand how the lateral hip muscles can load/unload the bodyweight forward/backward with one foot forward the other. However, we have no idea how and which anteroposterior hip muscles can load/unload the bodyweight forward/backward with the feet side by side. Previous authors did not explain the physiological basis of their computations (Lafond et al., 2004; Rougier, 2007, 2008; Winter et al., 1993, 1996). At least, our experiment confirmed that clearly different conditions (static vs. active; simple body oscillations vs. more or less broad body motions; body oscillations more controlled at the ankle vs. hip) did not lead to any significant difference in AP COP_v and almost none in AP COP_c. These findings favored our biomechanical arguments. For all these reasons, we have argued for the appropriate use of this model in the AP and ML axes. This is important methodological information in order to accurately measure and discuss the distinct roles of COP_v and COP_c in postural control. It may make it possible to better test and understand age-related (Rogers & Mille, 2003) and disease-related (Bonnet, Carello, & Turvey, 2009) changes in postural coordination.

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