Relative contribution of the pressure variations under the feet and body weight distribution over both legs in the control of upright stance

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Abstract

The resultant centre of pressure (CPRes) trajectories are aimed at controlling body movements in upright stance. When standing on two legs, these trajectories are generated by exerting reaction forces under each foot and by loading–unloading mechanisms intervening at the hip level. To assess the respective contribution of each of these factors in stance maintenance, a group of healthy individuals were tested in several conditions including standing quietly and voluntarily producing under each foot larger CP displacements in phase and in opposite phase along medio-lateral (ML) and antero-posterior (AP) axes. The results, based on the computation of coefficients of correlation between CPRes trajectories and various time series including the relative body weight applied to one leg and plantar CP trajectories, highlight some differences according to the axes along which the displacements take place and the amplitudes of the movements. Furthermore, the comparison of the CPRes trajectories resulting from each one of these two factors reveals the predominant role played by the loading–unloading mechanisms intervening at the hip level for the movements along the ML axis and those of the plantar CP displacements along the AP axis. Increasing the plantar CP displacements in phase or in opposite phase substantially modifies these contributions although without inferring a shift to the benefit of the other mechanism. The specific morphology of the ankle and hip joints implicated in this postural task plainly explains this postural control organisation. In particular, the link between the segmental configuration of the lower limbs and these mechanisms are discussed.

Keywords: Posture; Balance; Body weight distribution; Pressure variation; Humans

1. Introduction

For many decades, undisturbed upright stance maintenance has been an easy way to investigate postural control in both healthy individuals and disabled patients. Subjects are required to stand still on a force platform whose main output is generally the successive points of application of the resultant reaction forces, namely the net body or resultant centre of pressure (CPRes) displacements. This technique presents various advantages such as leaving the subjects to stand freely, with simple instructions and without any particular prior preparation. However, a major difficulty lays in the quantification and the interpretation of the CP trajectories. These reaction forces are indeed aimed at counteracting the gravity acceleration, which is exerted at the level of the centre of gravity (CG). Since the CG is endowed with body inertia, its vertical projection (CGv) cannot follow the CPRes movements through which it is controlled (Winter et al., 1996) with any precision. As a result, a gap between the two movements, namely CP–CGv, is always present and its amplitude is proportional to the horizontal acceleration communicated to the CG (Brenière et al., 1987).

Besides this approach, which legitimates the decomposition of the complex CPRes trajectories into basic components (CP–CG and CGv movements) and highlights the biomechanical relation between CPRes and CG movements, there is also the specificity of bipedal standing which has to be considered. Although having two legs is undoubtedly an interesting feature for locomotion, or for recovery from a loss of balance by a rapid enlargement of the base of support.
support, it highly complicates the ways in which the above-mentioned reaction forces can be generated. This question can be solved for the most part when postural perturbation is weak, by using a symmetrical pattern involving both feet similarly and simultaneously. For this reason, resort to a single force platform to study the postural strategies of subjects standing symmetrically with their two legs cannot be open to criticism. However, when an asymmetrical behaviour occurs either on the body weight distribution over the two feet (1), in the patterns of the two plantar CP trajectories (2), and/or also when these actions are not performed in phase, resorting to a system composed of two separate force platforms becomes indispensable. In that case, a mathematical relationship is used to compute the CP_{Res} from each support (Winter et al., 1996). To be more precise, for movements projected along the medio-lateral (ML) and antero-posterior (AP) axes, this calculation takes both plantar CP trajectories, i.e. those measured under each separate foot, weighted by the relative load applied on each leg into account:

$$\text{CP}_{\text{Res}} = \frac{\text{CP}_{L} \cdot \text{RF}_{L}}{(\text{RF}_{L} + \text{RF}_{R})} + \frac{\text{CP}_{R} \cdot \text{RF}_{R}}{(\text{RF}_{L} + \text{RF}_{R})},$$

where CP_{L}, CP_{R}, RF_{L} and RF_{R} are the plantar CP displacements and the amplitudes of the reaction forces exerted by the left and right legs, respectively.

Through this formula, one can see that a given position of the CP_{Res} can be the result of an infinite number of combinations involving the two plantar CP trajectories and body weight distribution. As highlighted by Winter et al. (1996), the distinction between these two parameters is important in the sense that it could help to specify the relative contribution of the muscular groups involved in postural control. A biomechanical analysis indeed reveals that the plantar CP trajectories in a healthy individual required to stand still are principally the result of the activation of the ankle extensors (triceps surae) along the AP axis whereas the control of the body weight distribution along the ML axis is largely achieved by the adductor–abductor muscles of the hip. However, this organisation is not necessarily set in stone. In addition, when a body weight asymmetry is seen, the more loaded a support, the larger the influence of its plantar CP movements over the CP_{Res} movements and thus over the CG_{v} movements (Genthon and Rougier, 2005).

Our main objective, in this study, is thus to quantify the relative influence of body weight and feet pressure distributions (PD) in postural control. Its general principle is based on the computation of CP_{Res} trajectories after one or the other factor has been neutralised, i.e. substituting some of the terms in the above expression with temporal averages. For instance, the role of the load–unload can be isolated by averaging over the left and right plantar CP displacements.

Lastly, depending on whether the plantar CP displacements operate in phase, or conversely in opposite phase, it can be shown that their biomechanical effects upon the CP_{Res} displacements (and thus postural control) will tend to be complementary or induce a reduced resultant effect, respectively. This particularity could be important for setting particular strategies aimed at improving postural stability, as shown recently in hemiplegics (Genthon et al., 2005). This point can be highlighted by assessing the coefficients of correlation from plantar and resultant CP trajectories.

2. Methods

Nine healthy subjects were included in this study. All subjects were adults averaging 25+6 years (mean+SD), (height: 175.6+4.8 cm; body weight: 74.6+12.7 kg), having no known neuromuscular impairment. Subjects were informed of the protocol and consented to participate.

Postural sway was measured by two rectangular (20×35 cm) force platforms (PF02, Equi+, Aix les Bains, France) collaterally installed on which the subjects placed each foot, respectively. The ground force reaction forces, issued from four vertical mono-axial load cells (range of measurements 0–150 daN) for each platform, were simultaneously monitored by the investigator during the tests. The signals issued from the dynamometric load cells were amplified and converted from analogue to digital form through a 14 bits acquisition card and then recorded with a 64 Hz frequency on a personal computer.

The subjects stood barefoot on the force platforms with their arms at their sides and the inner edges of the feet parallel and separated by 20 cm. The mean positions of the plantar and resultant trajectories were calculated with regards to a referential defined by the intersection of the line passing behind the heels and the sagittal median line between both feet. Positive values indicate that the position is situated in a forward direction and to the right with respect to these lines. Five conditions were randomly presented to all subjects. A first one consisted in standing still and upright swaying as little as possible and was used as a reference (REF). The four remaining conditions consisted of performing plantar CP displacements on a regular basis (with the assistance of a metronome recording at 0.33 Hz) along either the ML or AP axes. The subjects were required to coincide their change of direction of the plantar CP displacements with the metronome rhythm. For each axis, the subjects were instructed to produce either in phase or in opposite phase movements of constant amplitudes. Precisely, the subjects perform pronation–supination of each foot and rock forward and backward along the ML and AP axes, respectively. No instruction was given regarding the shift of the BW distribution. These conditions were called MLP, MLo, AFl, and APo, respectively. The effective performance was permanently and accurately checked by the investigator throughout the measurements.

Overall, five trials of 32 s were recorded for each condition with rest periods of similar duration between the trials and two minutes between the conditions.

Along both ML and AP axes, the coefficients of correlation between the CP_{Res} trajectories on the one hand and the various elements implicated in their computation (i.e. the percentages of body weight applied to each leg and the plantar CP displacements of each leg) on the other hand were calculated and averaged throughout the trials for each given condition. For each ML or AP axis, the determination of the respective contribution of the PD and the body weight loading-unloading (LU) mechanisms was achieved for each trial by computing two “theoretical” CP_{Res} trajectories (CP_{PD} and CP_{LU}). To be precise, instead of applying the formula given in the introduction each time, the successive values of the relative body weight applied on each leg and the successive plantar CP positions were held constant and replaced by their mean for the whole duration of the trial. This procedure slightly contrasts with the one proposed by Winter et al. (1996) who computed with the same principle the CP_{PD} and then determine the CP_{LU} trajectories by subtracting the CP_{PD} from the CP_{Res} displacements. Since, with our method, CP_{Res} displacements cannot be viewed as the simple addition of CP_{PD} and CP_{LU}
displacements, a novel procedure needs to be proposed to quantify their respective contribution. The standard deviations of these trajectories (SD(PD) and SD(LU)) were then computed and used to assess the respective contribution of the PD and LU mechanisms through the following formula:

\[
\text{Contr}_{PD} = \frac{\text{SD(CP}_{PD}\}}{\sqrt{\text{SD(CP}_{PD}\} + \text{SD(CP}_{LU}\}}},
\]

\[
\text{Contr}_{LU} = \frac{\text{SD(CP}_{LU}\}}{\sqrt{\text{SD(CP}_{PD}\} + \text{SD(CP}_{LU}\}}},
\]

Interestingly, Contr\(_{PD}\) + Contr\(_{LU}\), with this method, is equal to 1. Consequently, the statistical analysis, consisting of Anova of Friedman and the post hoc effects, through Dunn tests, was processed along each ML and AP axis with one parameter only. In all cases, the first level of significance retained was \(p<0.05\).

3. Results

3.1. Correlation analysis

Fig. 1 and Table 1 display a typical example of the various plottings obtained throughout the five conditions and the averaged coefficients of correlation for the group, respectively. It can be shown than performing reduced or large body sways either in phase or in opposite phase substantially modifies postural organisation and the relation between CP\(_{Res}\) and plantar CP\(_{L}\) and CP\(_{R}\) movements. The similar shapes of the plots relative to the body weight applied to the right foot (upper line) and the CP\(_{Res}\) trajectories projected along the ML axis should be noted for all conditions. Another striking feature that can be observed through this figure is the dependence, along the AP axis of the CP\(_{Res}\) trajectories with the plantar trajectories when body sways are performed in phase (REF and AP\(_p\)). On the other hand, the amplitudes of these CP\(_{Res}\) trajectories appear to be significantly reduced when the plantar CP movements are performed in opposite phase (AP\(_o\)). It is also worth highlighting the rather different effect the large movement conditions have upon the CP\(_{Res}\) trajectories along the ML axis. Performing plantar CP movements in phase indeed determines a much more ample CP\(_{Res}\) displacement than when these movements are performed in opposite phase.

These visual impressions from the data measured with one subject are confirmed with the computation of the coefficients of correlation involving the various components (Table 1). In particular, it appears that coefficients close to 1 are found along the ML axis between the CP\(_{Res}\) trajectories and the body weight distribution for all experimental conditions. This feature highlights the importance of the role played by body weight distribution in the determination of the ML CP\(_{Res}\) movements. This role is not modified during the performance of larger plantar CP displacements. Conversely, one may notice coefficients close to zero for the orthogonal AP axis, hence indicating that other factors are involved in these trajectories. Table 1 also indicates that AP CP\(_{Res}\) movements are mainly

![Fig. 1. Typical plots obtained from one trial of one subject across the various conditions. From top to bottom are displayed the relative body weight applied on the right foot, the plantar CP displacements for the two feet and the CP\(_{Res}\) displacements along ML and AP axis, and the planar trajectories.](image)
Table 1
Average coefficients of correlation (standard deviation) computed between the CP<sub>Res</sub> trajectories and the various time series (body weight distribution applied on the right foot, plantar CP trajectories under each left and right foot) along each ML and AP axes and reported for each condition (see text for details)

<table>
<thead>
<tr>
<th>Condition</th>
<th>REF</th>
<th>ML&lt;sub&gt;p&lt;/sub&gt;</th>
<th>ML&lt;sub&gt;o&lt;/sub&gt;</th>
<th>AP&lt;sub&gt;p&lt;/sub&gt;</th>
<th>AP&lt;sub&gt;o&lt;/sub&gt;</th>
</tr>
</thead>
<tbody>
<tr>
<td>CP&lt;sub&gt;Res&lt;/sub&gt; ML</td>
<td>BW on RF</td>
<td>0.996 (0.004)</td>
<td>1.000 (0.001)</td>
<td>0.982 (0.008)</td>
<td>0.992 (0.004)</td>
</tr>
<tr>
<td>Plantar CP (LF)</td>
<td>−0.467 (0.212)</td>
<td>−0.637 (0.392)</td>
<td>−0.045 (0.148)</td>
<td>−0.069 (0.346)</td>
<td>0.177 (0.273)</td>
</tr>
<tr>
<td>Plantar CP (RF)</td>
<td>−0.213 (0.256)</td>
<td>−0.496 (0.343)</td>
<td>0.023 (0.175)</td>
<td>−0.077 (0.318)</td>
<td>0.178 (0.210)</td>
</tr>
<tr>
<td>CP&lt;sub&gt;Res&lt;/sub&gt; AP</td>
<td>BW on RF</td>
<td>0.034 (0.173)</td>
<td>−0.032 (0.439)</td>
<td>0.049 (0.079)</td>
<td>−0.059 (0.363)</td>
</tr>
<tr>
<td>Plantar CP (LF)</td>
<td>0.942 (0.038)</td>
<td>0.427 (0.382)</td>
<td>0.879 (0.058)</td>
<td>0.997 (0.002)</td>
<td>0.431 (0.254)</td>
</tr>
<tr>
<td>Plantar CP (RF)</td>
<td>0.954 (0.035)</td>
<td>0.363 (0.365)</td>
<td>0.900 (0.069)</td>
<td>0.966 (0.004)</td>
<td>0.506 (0.130)</td>
</tr>
</tbody>
</table>

Note the high level of correlation for the CP<sub>Res</sub> trajectories along the ML axis with the body weight distribution and along the AP axis with the plantar AP CP trajectories in the AP<sub>p</sub> condition.

3.2. Relative contribution of PD and body weight LU mechanisms

The shapes of the CP<sub>Res</sub> trajectories can be explained through these two mechanisms. Fig. 2 displays these trajectories for one subject across the five conditions. One can see in particular that all the CP<sub>Res</sub> time series are rather close (or even for some conditions indistinguishable) to either the CP<sub>LU</sub> trajectories (which do not take into account the pressure variations under the feet) for the ML axis, or the CP<sub>PPD</sub> trajectories (which do not take into account the relative body weight applied to each leg) for the AP axis. Consideration of the planar plotting (i.e. in the plane of support) further highlights this organisation. Performing substantial amplitudes through these opposite control mechanisms along each orthogonal ML and AP axes indeed provides CP<sub>Res</sub> movements spreading over a large territory.

As can be seen from Table 2, these contributions are different depending on the amplitude of the plantar CP displacements and the ML or AP axes. Overall, it appears that the body weight LU plays the main role in determining the CP<sub>Res</sub> trajectories along the ML axis whereas along the AP axis, the main contribution arises from the PD. However, these trends appear to be robust since they are only slightly modified when the plantar CP displacements are increased. The Anova of Friedman reveals that the experimental conditions affect the values of these parameters for both ML ($\chi^2 (9,4) = 24.09, p < 0.001$) and AP axes ($\chi^2 (9,4) = 24.00, p < 0.001$). The results of the post hoc analysis, conducted through the Dunn tests, are displayed in the lower part of Table 2 and further explain these effects.

4. Discussion

The data of this study specifies the precise mechanisms operating in order to secure balance from bipedal stance and thus confirms the results of past studies on this topic (Day et al., 1993; Winter, 1995; Winter et al., 1996). Nevertheless, its novelty lies in the quantification of the linkage of the actions occurring under each foot and the resultant effect upon the CP<sub>Res</sub> trajectories, which allows us to control our CG and then our upright posture.

Through the correlation analysis, it was shown that these CP<sub>Res</sub> trajectories were the result of specific mechanisms operating at the hip level for movements intervening along the ML axis and at the ankle level for those intervening along the AP one. This organisation is naturally dependent on our anatomy. The orientation of these joints is indeed the main cause of these actions. The frontal plane in which the two hips are embedded favour, above all, LU mechanisms whereas the talocrural axes of the ankle joints favour the application of reaction forces in the sagittal plane. The longitudinal patterns, observed in the REF condition, confirm this organisation. From a muscular point of view, the forward projection of the CG with regard to the ankle axes explains the sole involvement of the toe flexors in this task and thus this particular pattern. Interestingly, it was shown that when the motor command was impaired, as in the case of hemiplegics, this specificity could disappear (Genthon et al., 2005).

Thanks to the computation of the coefficients of correlation between the various time series, one can see the robustness of this organisation. Even though the plantar CP displacements are voluntarily increased, a certain constancy of the coefficients obtained from the time series of body weight distribution is observed. This is particularly true for the ML movements. To be more precise, these enhanced relationships between plantar and resultant CP trajectories can be seen during the “in phase” conditions. On the other hand, during the “opposite phase” conditions, the inverse phenomenon is observed. The main reason of this effect probably arises from the higher contribution of the deterministic control
mechanisms (as opposed to the stochastic ones), which are generally observed when the movement amplitudes are enhanced. This phenomenon was for instance observed in impaired individuals, such as the elderly, who sway more spontaneously (Berger et al., 2005) and whose CPRes trajectories are endowed with less random-walk activity.

The relative contribution of PD and body weight LU mechanisms in the genesis of the CPRes trajectories also provides interesting information. By specifying that the control mechanisms operate at various joints’ levels and thus involves various muscles, these data confirm the electromyographic data classically observed for upright
stand maintenance (Okada, 1972). It also confirms the predominant role of the weight LU mechanisms in the CPRes displacements along the ML axis and the main contribution of the PD patterns in those intervening along the AP axis. It should be emphasised that for both axes, one or the other mechanism is largely predominant. However, by increasing the amplitudes of the plantar CP displacements, one can determine the range through which these values could evolve. As expected, the larger variations for our ContrPD or ContrLU indices computed from trajectories intervening along the ML axis occurred with body motions made in that direction. According to the same principle, some effects were found along the AP axis but were smaller in size. These differences in the variation ranges could be simply explained by the mechanical differences in the body structure explained by the frontal and sagittal planes, as pointed out by Day et al. (1993). According to these authors, whereas body motions can be expected through ankle, knee and hip joints independently in the sagittal plane, the hips and ankle appear to be linked in the frontal plane. This means that a change in one angle joint (for instance the right ankle) leads necessarily to predictable changes in the three remaining joints (left ankle, right and left hips) provided there is negligible movement at the knee level. Since all our conditions inducing body motions along the AP axis deliberately make use of the ankles, this would explain the lowest variation range observed in the ContrPD or ContrLU indices.

Our data also emphasises the biomechanical effects induced by reaction actions exerted in phase or in opposite phase. If we consider maximum restraint of amplitudes of the CPRes displacements as the main objective, the production of plantar CP movements in opposite phase appears to be an appropriate response. However, the main difference between control of movement along the ML and AP axes arises from the muscular synergies. Due to the morphological configuration of the lower limbs, activating agonistic muscles simultaneously and bilaterally produces symmetrical patterns along the ML axis, which, with an equal body weight distribution over the two supports, may result in a null CPRes displacement. The problem is quite different for the orthogonal AP axis since, in that case, opposite direction displacements should necessarily involve either the activation of the antagonistic muscles (tibialis anterior) and/or the relaxation of the agonistic ones. In other words, it looks as if our muscular anatomy at the foot level is organised to favour reduced displacements along the ML axis and larger ones along the AP axis. This specificity can also be discussed as a complement to the muscular synergies occurring at hip level, which principally intervene through the frontal plane. It also highlights the low level of redundancy in the implementation of a particular strategy.

Naturally, a unique motor command simultaneously activating the postural muscles of both legs should be regarded as simplifying the control process. However, as shown recently by Mochizuki et al. (2005), if the soleus muscles bilaterally act to control the standing posture, there was little evidence of a synchronisation of their motor units. Setting a strategy consisting in recruiting antagonistic muscles, although biomechanically interesting, would nonetheless significantly complicate the programming of the motor command.

To conclude, the particular anatomy of the lower limbs allows us to exert reaction forces through hip and ankle joints. Each of these joints because of its morphology is mostly called into play to control body movements either through the frontal or sagittal planes. In the REF upright position, which sees the edges of the feet being parallel, there is indeed little redundancy between the two basic mechanisms, which consist in body weight LU mechanisms upon each leg and pressure variations under the feet. However, it remains unclear whether this organisation is partly linked to the positioning of the feet with respect to the hip axis.

References


