Patients wear rigid orthoses either to prevent repetitive ankle sprains or to aid the rehabilitation process. Wearing an orthosis is thought to increase both rigidity of the joint and tactile sensory information (Feuerbach, Grabiner, Koh, & Weiker, 1994). To improve the suitability of these orthoses, satisfaction questionnaires and in situ tests can be used to adapt the product more closely to the potential user’s needs. These tests provide more objective data that are closer to normal use. Several studies over the past few decades focused on protocols testing participants standing upright and as still as possible. This task focuses on the ankle joints, at least for body movements along the anterior-posterior (AP) axis (Winter, Prince, Frank, Powell, & Zabjek, 1996; Rougier, 2007), and has been a natural choice for testing ankle orthoses in both healthy and disabled patients. The improved capacity to reduce body motions can be interpreted as an indication of improved sensorimotor coordination and, therefore, a good predictor of improved efficiency for more ecological motor tasks.

On the whole, these investigations used various protocols including one-legged (OL) or unipedal (Baier & Hopf, 1998; Bennell & Goldie, 1994; Feuerbach & Grabiner, 1993) and two-legged (TL) or bipedal standing (Calmels, Escafit, Domenach, & Minaire, 1991; Rougier, Burdet, Farenc, & Berger, 2004). By inducing movements around the subtalar and talocrural axes, the OL stance puts the ankle in conditions close to those experienced with ankle sprains and differentiates injured participants from healthy ones (Cornwall & Murrell, 1991). The TL stance is less constraining for the ankle joint, where movements occur principally around the talocrural axis. These two conditions also correspond to two levels of difficulty: it is easier to stand on two feet than one. Their biomechanical constraints, and consequently the induced postural effects, are also different. Recently, these two conditions were compared (Burdet & Rougier, 2007), showing that the more efficient participants in the OL stance were not necessarily the most efficient when using both feet. Two fundamental differences characterize

**Key words:** healthy individuals, posturography, test, upright standing

To highlight the capacity of one- and two-legged standing protocols when assessing postural behavior induced by a rigid ankle orthosis, 14 healthy individuals stood upright barefoot and wore either an elastic stocking on the preferred leg or a rigid orthosis with or without additional taping in one- or two-legged (TL) conditions. Traditional center-of-pressure (CP) measures were evaluated for the total two-feet resultant CP and under the feet (plantar CP). Focusing on the plantar CP displacements under the leg fitted with the various orthoses demonstrated particular postural behaviors for traditional parameters with main effects along the mediolateral axis. Only the TL protocol showed the limiting effects of the rigid shells on the inversion-eversion movements in healthy individuals.
the natural body frequency (Brenière, 1996).}

Most studies assessing postural performance have measured reaction forces with a force platform. In most cases, these analyzed successive positions of the application points of the resultant reaction forces, better known as resultant center-of-pressure (CP_{res}) displacements. Throughout this article, we use the term “resultant” to indicate the CP_{res} displacements resulting from the juxta-position of the two biomechanical mechanisms (pressure distribution under the feet and body weight distribution over the two feet) involved in upright stance control (for further details, see Winter et al., 1996, and Rougier (2007)). Although this measure is fine for assessing postural control in OL standing, several flaws appear when measuring TL stance. The main one is the significance of these CP_{res} displacements with regard to the above-mentioned mechanisms. It is extremely difficult with a single force platform to determine whether the altered CP_{res} displacements stem from increased or reduced plantar CP displacements and/or from a better or worse control of load distribution over two legs. Furthermore, with two force platforms it is possible to study the pattern of the plantar CP displacements. In [AQ: uninjured?] individuals, the pattern of rectilinear displacements along the longitudinal axis of a barefoot participant (Genthon & Rougier, 2005) could be substantially modified by stiffening the ankle with an orthosis. Last, increasing the stiffness of a single joint might induce compensatory strategies in the other leg, leading to normal CP_{res} displacements undetectable with a single platform.

Parallel to this is the biomechanical significance of these CP_{res} displacements regarding body displacements. The unique objective of the CP_{res} displacements is to control the center-of-gravity (CG) displacements, which cannot be motionless in these tasks. The inability of the CG to be motionless is due to inertia of the body in the standing posture that prevents the CG from following the CP_{res} displacements accurately, thus, creating a gap between them and a horizontal acceleration (Brenière, Do, & Bouisset, 1987; Gage, Winter, Frank, & Adkin, 2004). Interestingly, an upright quiet standing task presents the advantage of allowing an easy estimation of these CG displacements through a biomechanical relationship account for the natural body frequency (Brenière, 1996). The main requirement of this relationship is a constant moment of inertia, something that fits well with the TL conditions. On the other hand, OL standing, by possibly involving various trunk and upper limbs movements due to task difficulty for most individuals, prevents the use of this approach. Therefore, CG (and CP-CG) displacements only were estimated for the TL protocol.

Although OL and TL tasks have biomechanical specifics, it is unclear which is more suitable to assess the physical effects of a rigid ankle orthosis on posture. To investigate this, a sample of healthy adults was tested in both OL and TL postures. We assumed healthy participants would not be looking for adjuncts to help ankle stability and that, contrary to injured patients who are unable to stand without orthosis, there was no real consequence if the orthotic device did not increase stability. To understand the real effect of the orthosis, we chose healthy participants to (a) exclude any interaction between postural control and such factors as pain and sprain severity, and (b) decrease the intra- and interparticipant variability in a large number of trials per condition. Four conditions were investigated in each OL and TL protocol: standing barefoot, wearing elastic stockings, wearing a rigid orthosis taping, and with and wearing a rigid orthosis without taping. Although somesthetic cues should play a role in the postural effects induced by wearing the various products, we were bound to test the participants in OL protocol with their eyes open to infer a reduced task difficulty. There has been little investigation of a single leg stance with eyes closed, because repetition does not warrant a lack of interaction with fatigue, hence emphasizing the advantage of using TL protocol. Our hypothesis was that wearing the rigid orthosis would stiffen the ankle and, thus, reduce the CP_{res} displacements. When comparing OL and TL postures a priori, it appears that both talocrural and subtalar axes should be particularly sensitive to increased ankle rigidity.

**Method**

**Materials and Procedure**

Fourteen healthy, physically active men ranging in age from 21 to 43 years (M body mass = 71.4 kg, SD = 9.8; M height = 1.77 m, SD = 0.04) participated in this study. Those with no known balance pathology or previous sprain in the previous 6 months were included. Participants were informed about the protocol and provided informed consent to participate, as required by the Helsinki declaration (1964) and the local ethics committee. All had a preference for their right leg (i.e., the one displaced first when initiating a step forward). The participants performed in two protocols in random order, one consisting of standing upright on a single limb (OL protocol) and the other on both feet (TL pro-
protocol). In both, they stood barefoot on a platform with their arms at their sides and were required to minimize body motions as much as possible. If they did not follow these instructions, the trials were re-recorded. Depending on the protocol, the measurements were taken for one (for the OL protocol) or two force platforms (for the TL protocol). Four experimental conditions were carried out successively and randomly for each OL and TL protocol. The first consisted of participants performing the postural barefoot task (BAR). In the remaining conditions, participants wore on their dominant leg an elastic stocking (STOC), a rigid orthosis (ORT; Ligastrap Immo model, Thuasne, Levallois-Perret, France,) or the latter over elastic taping (ORT+T). This model was chosen because of its wide use among patients. For the last two conditions, a single investigator always attached the orthosis to standardize compression, joint position, and mobility with the adjustable strap. Before each condition, participants took several steps to adjust to the orthosis. The various conditions are depicted in Figure 1. In all cases, 10 trials lasting 32 s were recorded with rest periods of similar lengths between trials. The number of a priori trials ensured data reliability (Doyle, Hsiao-Wecksler, Ragan, & Rosengren, 2007). This duration is a good compromise between the need to collect data over long periods of time (Carpenter, Frank, Winter, & Peyser, 2001) and fatigue prevention. In addition, 32 s at 64 Hz yields 2,048 data points, an optimal number to support the Fast Fourier transform. The various parameters were computed for each trial and then averaged for subsequent statistical processing.

In the OL protocol, participants were instructed to stare at a target placed 1 m before them at eye level. As seen in Figure 1, the dominant foot was positioned along the height of the triangular (80 cm each side) force platform (PF01, Equi+, Aix les Bains, France). In particular, participants were instructed to stabilize their nonweight-bearing leg as shown in Figure 1A (i.e., the first metatarsophalangeal joint at the level of the internal malleolus). In the TL protocol, participants were instructed to distribute their weight equally over their legs and to close their eyes during each measurement (to focus on somatosensory feedback). The feet were externally rotated at 30°, with the heels 9 cm apart. Postural sway was measured by two rectangular (20 × 35 cm) collaterally installed force platforms (PF02, Equi+, Aix les Bains, France) on which participants placed both feet (see Figure 1). For both platform devices, the signals from the three or eight dynamometric load cells (for the single and double force platforms, respectively) were amplified and converted from analog to digital form before being recorded (sample frequency, 64 Hz) on a personal computer. The plantar and resultant CP displacements were calculated in different ways using specific software (Prog01 or Prog02, Equi+, Aix les Bains, France) and checked simultaneously during the tests.

**Estimation of the CP<sub>Res</sub>, CG<sub>v</sub>, and CP-CG<sub>v</sub> Displacements**

As stated, the CP trajectories for both OL and TL protocols were measured separately either by one or two force platforms. For the latter conditions, it was necessary to distinguish the plantar CP displacements involving the left (CP<sub>lf</sub>) and the right foot (CP<sub>rf</sub>) on which the orthosis or taping was positioned. The resultant CP (CP<sub>Res</sub>) displacements were calculated along each ML or AP axis from the left and right CP displacements using the following formula (Winter, 1995):

\[
CP_{Res} = CP_{lf} \times R_{lf} / (R_{lf} + R_{rf}) + CP_{rf} \times R_{rf} / (R_{lf} + R_{rf})
\]

where \(R_{lf}\) and \(R_{rf}\) were the vertical reaction forces under the left and the right foot, respectively. By definition, \(R_{lf} + R_{rf}\) oscillates about body weight in our postural tasks. A schematic representation illustrating the successive steps used for the data analysis from two force platforms is shown in Figure 2.

The CP<sub>Res</sub> displacements are noteworthy in that their action on the CG appears to facilitate its falling motion at times and counteract it at others (Rougier & Caron, 2000). Consequently, these displacements can be divided into two basic components: the vertical projection of the CG over the support plane (CG<sub>v</sub>) and the difference between these two displacements (CP<sub>Res</sub> - CG<sub>v</sub>). Because body sways during TL stance are reduced, CG displacements can be deduced from CP<sub>Res</sub> trajectories, if there is a frequency relation between the variables (Brenière, 1996; Caron, Faure, & Brenière, 1997). It is relevant, therefore, to consider that CP<sub>Res</sub> displacements operating over high frequencies do not imply appreciable CG displacements due to substantial body inertia. This method is a generalization of the
concept initially developed by Brenière (1996) for more dynamic conditions. His hypothesis was that the body constitutes a low-pass filter, which would explain the loss in amplitude observed between $\text{CP}_\text{Res}$ and $\text{CG}_v$ as frequency increases. In fact, as depicted in Figure 2, the principle consists of multiplying the data, transformed in the frequency domain, using a Fast Fourier transform (FFT), by the filter mathematically defined by the ratio:

$$\frac{\text{CG}_v}{\text{CP}_\text{Res}} = \frac{2}{(\omega_0^2 + \omega^2)} [\text{AQ: Missing symbols.}]$$

and then to recover the temporal domain by processing an inverse FFT. In this formula, $\omega_0 = \sqrt{\frac{mg}{(I_G + mh^2)}}$, where $m$, $g$, $h$, and $I_G$ are the participant’s mass, gravity acceleration, distance from CG to the ground, and moment of body inertia around the ML or AP axis with respect to the CG, respectively. This corresponds to a biomechanical constant, the natural body

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Figure 2. Calculation and data treatment; calculation: (a) collecting $\text{CP}_\text{H}$ and $\text{CP}_\text{R}$ displacement data, (b) calculating $\text{CP}_\text{Res}$ displacements along ML and AP axes, and (c) estimating $\text{CP}_\text{H}$ and $\text{CG}_v$ displacements (from Genthon & Rougier, 2005).
frequency (Brenière, 1996), and \[ \Omega f \] is the pulsation (rad/s). Because the moment of inertia differs for the ML and AP axes, two distinct relationships were used to compute it depending on participants’ weight and height (Ledebt & Brenière, 1994).

**Signal Processing**

The various displacements (\( CP_{\text{Res}} \), \( CP_{\text{lf}} \), \( CP_{\text{rf}} \), \( CG_v \), \( CP-CG_v \)) were studied through several parameters such as the mean positions and variances (sum of squared differences from the mean divided by number of positions) along the ML and AP axes, the areas of an ellipse calculated with a confidence interval (i.e., excessive displacements that can interact substantially with the computation have been removed; Tagaki, Fujimura, & Suehiro, 1985), and the mean velocities (total path lengths divided by trial duration). These parameters, commonly used in posturography because of their simplicity, measure the amount of CP displacements irrespective of their direction. All displacements were calculated with regard to the intersection of the line passing behind the heels and the sagittal median line between both feet. Positive values indicate a forward direction (positive AP axis) and to the right (positive ML axis) with respect to these lines. To compare the induced postural effects during OL and TL conditions, a one-way Friedman analysis of variance (ANOVA) with repeated measures was used, with the post hoc analysis using nonparametric Dunn tests. As shown by the Kolmogorov-Smirnov tests, all parameters computed from both protocols or the four conditions were not normally distributed, thus, justifying the use of nonparametric tests in each case. For all analyses, \( p = .05 \) (alpha) was significant. If a difference was not significant, the beta risk of an erroneous conclusion of equivalence was \( p \leq .2 \).

**Results**

The Friedman ANOVA indicated no effect of ankle support on the mean positions along both ML and AP axes in the OL protocol. A similar result was also found for the mean positions of the \( CP_{\text{Res}} \) displacements in the TL protocol (see Figure 3). However, when considering the mean positions of the plantar \( CP_{\text{lf}} \) and \( CP_{\text{rf}} \) some statistically significant effects were noted, but only for the right foot, wearing the stocking or orthosis, and only along the ML axis \[ \Omega^2 = 11.4, \ p < .01 \]. Post hoc tests revealed that the effect stemmed from the ORT and ORT+T conditions (\( p < .05 \)). Precisely, as seen in Figure 3, the mean positions for these conditions in the TL protocol were larger along the ML axis compared to the BAR condition. Furthermore, the weight distribution remained constant throughout the conditions performed in the TL protocol.

As for the \( CP_{\text{Res}} \) trajectories, the areas of the ellipse, the mean velocities, and the variances measured along each ML and AP axis did not show a significant trend between conditions for the OL and TL protocols. A similar feature was found for the \( CG_v \) and \( CP-CG_v \) displacements for the TL protocol. On the other hand, when considering the plantar CP trajectories under each foot (i.e., \( CP_{\text{lf}} \) and \( CP_{\text{rf}} \) in the TL protocol), significant results were found only for the \( CP_{\text{lf}} \) displacements. Precisely, the Friedman ANOVA indicated a significant effect for the ellipse area \[ \Omega^2 = 10.03, \ p < .02 \], the mean velocity \( \Omega^2 = 16.71, \ p < .001 \), and the variances along the ML \( \Omega^2 = 12.7, \ p < .005 \) and AP axes \( \Omega^2 = 9.2, \ p < .03 \). The bar charts in Figure 4 show these global effects are mainly due to the orthosis conditions with the post...
hoc analysis. The differences noted were between STOC and ORT conditions and between the STOC and ORT+T conditions for the four parameters.

In the TL protocol, when comparing the plantar CP trajectories from both feet, the results indicated a specific behavior for the BAR condition. Unexpectedly, the areas covered by the CP\textsubscript{rf} appear to be reduced when compared to those of the CP\textsubscript{lf} ($p < .05$). The difference for the variances along the AP axis is thought to further explain this effect ($p < .05$), as no difference was noted along the ML axis. Finally, the mean velocity also appears to be reduced for the CP\textsubscript{rf} trajectories. This latter effect was also observed for the STOC condition ($p < .05$). Interestingly, wearing the rigid orthosis (with or without additional tap-
ing) on the right foot significantly affected the patterns of the plantar CP trajectories. Some statistical effects can be seen with the ORT condition for the ellipse area ($p < .05$), the variances on both ML and AP axes ($p < .05$), and the mean velocity ($p < .01$). For the ORT+T condition, the effects were observed only for the variance along the AP axis ($p < .05$) and the mean velocity ($p < .01$).

**Discussion**

Our results showed that the CP$_{rf}$ displacements, whether measured from a single or two separate force platforms, demonstrated no statistically significant effect. In contrast, analysis of the plantar CP trajectories provided the most powerful method for differentiating postural behavior between the experimental conditions, but this required TL standing. The lack of effect along the ML axis in the OL protocol contradicted our hypothesis that OL standing would be sensitive to increased ankle rigidity caused by an orthosis with two lateral rigid shells. Also, the CP displacements in this task did not show subtle variations in the different conditions and was not enough to cause measurable differences in the CP$_{rf}$ results. The load and, thus, the constraint of the orthosis plays a role in participants’ ability to control their reaction forces.

Statistical analysis demonstrated that one effect of our TL protocol involved the mean positions of the plantar CP$_{rf}$ displacements for the ORT and ORT+T conditions. We believe this effect can be explained by the thickness of the plastic shell, which prevents the internal edge of the right foot from touching the platform wedge. Consequently, these results must be viewed as an artifact.

For the reduced CP$_{rf}$ displacement to not result in a concomitant reduction of the CP$_{rf}$ displacement along the ML axis, it should be accompanied by a similar change in the contralateral CP$_{rf}$ displacement, because both plantar CP displacements intervene in phase with each other. With these conditions and the constant weight distribution throughout the measure, the only way to reduce CP$_{rf}$ and CG displacements, when plantar CP are displaced in opposite directions, is to modulate the ML displacements under both feet. In other words, if CP$_{rf}$ displacements are reduced due to the orthosis effect, the CP$_{rf}$ should behave accordingly. Interestingly, although no statistical trend was found regarding the CP$_{rf}$, the graphic results reveal there may have been a slight—although statistically insignificant—decrease for the ORT and moderate decrease for the ORT+T conditions (see Figure 4). It is worth noting the intervention of these biomechanical strategies, even though participants spontaneously made smaller displacements under their right leg. This could be due to the participants’ right-leg preference. Previous studies showed that CP$_{rf}$ displacements along the AP axis were highly correlated with those of CP$_{lf}$ and CP$_{rf}$ (Genthon & Rougier, 2003). This shows that weight distribution plays a minor role in determining CP$_{rf}$ displacements along the AP axis (Rougier, 2007). With constant weight distribution, a reduced CP$_{rf}$ displacement can be achieved when a reduction of the plantar CP trajectory is not counteracted by an opposite trend of greater amplitude on the other leg. In other words, contrary to the ML axis, in which opposing acting on both feet is a prerequisite for a positive effect on the CP$_{rf}$ displacements, an effect can be obtained along the AP axis by lowering the CP displacements under one foot only. As expected, a major effect of the orthosis is the reduction of ankle motion around the talocrural axis and, thus, along the AP axis, when compared to the STOC condition (see Figure 4). Because these statistically significant effects were not counteracted by an opposite behavior on the left foot, a slight reduction was observed for the CP$_{rf}$ displacements. The lack of statistical results at this level can be explained by the fact that about 50% (the weight supported by the right foot) of the effect observed on the CP$_{rf}$ trajectories was distributed on the CP$_{rf}$ trajectories.

When designing this protocol, it was thought that a participant wearing an ankle orthosis would feel both an increased tactile cue and a motor impairment. Thus, we included an elastic stocking condition as a reference, because its tactile effects were expected to be similar to the rigid orthosis. By including a barefoot condition, it was also possible to assess the effects of the elastic stockings on participants. The results of the STOC condition showed that both the OL and TL protocols had no significant effect with regard to the ORT and ORT+T conditions for plantar CP areas and variances along the AP axis. In contrast, there were some decreases for the variance along the ML axis and mean velocity.

Nevertheless, analysis of the CP$_{rf}$ trajectories (i.e., those corresponding to the side wearing the stocking) provided interesting results. The variances computed along the ML and AP axes were enhanced with the stocking and implied greater between-participant variability, as shown by the standard-deviations presented in Figure 4. The orthosis model used in this study has elastic tape that can be used in addition to the two rigid shells. Because of the shells’ physical properties, the tape does not further reinforce rigidity. Therefore, its positive effect may stem from enhanced tactile information, as was the case for the stocking. Generally speaking, the same features developed for the stocking can be applied to the tape. However, some statistically significant differences between the ORT and ORT+T conditions are worth noting for the decreased CP$_{rf}$ variances along both ML and AP axes. These variances, however, did not imply a significant effect on either the CP$_{rf}$ or CG displacements. Although OL standing focuses on the talocrural and subtalar axes, our data emphasize its relative inability to highlight any postural effect while improving the joint rigidity and providing tactile informa-
tion. This result is in accordance with the work of Ross and Guskiewicz (2004), who were unable to differentiate individuals with functionally stable and unstable ankles during OL standing. On the other hand, resorting to a TL stance and separate measurement of the two feet allowed us to differentiate postural behaviors. Thus, despite an increased loading of the foot favoring increased ankle stability, as shown in chronic patients (Scheuffelen, Rapp, Gollohofer, & Lohrer, 1993), this study did not substantiate OL standing as a useful paradigm for assessing the influence of various types of orthoses on postural control strategies. The sensorimotor task of standing on both feet should be viewed as an excellent means to test the human-material interface and, hence, a good way to assess the influence of various characteristics of rigid ankle orthoses. Future investigations involving ankle-sprain patients should be based on this experimental protocol.

References


Authors’ Notes

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