Quiet Postural Control of Patients With Total Hip Arthroplasty Following Joint Arthritis

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To assess the postural strategies developed by patients after total hip arthroplasty (THA), 14 patients were measured 12 days after surgery. The respective role played by both sound and prosthetic legs and the compensatory mechanisms were assessed through a separate measure of the center-of-pressure (CP) trajectories under each foot. The movements of the center-of-gravity (CG) were estimated from those of the resultant CP to determine postural performance. The postural behavior was compared with those of a group of age-matched healthy subjects required to adopt a slightly asymmetrical weight distribution. Patient results indicate greater movements for both plantar and resultant CP displacements, principally along the antero-posterior (AP) axis, a decreased contribution of the hip mechanisms in the production of CP displacements along the medio-lateral (ML) axis, greater resultant CP and CG movements along the AP axis and increased differences between CP and CG along both ML and AP axes. The postural specificity of the THA patients appears to be due to a global sensorimotor impairment that alters the control of the loading-unloading mechanism at the hip level.

Keywords: upright stance, balance, postural asymmetry

Total hip arthroplasty (THA) is a common surgical procedure that allows individuals, often over 50 years of age, to restore their mobility capacities and diminish pain. Even though several approaches can be considered for the surgery, a common point is that both sensory and motor components are necessarily and consequently impaired (Grigg, Finerman, & Riley, 1973). Various tasks including locomotion can be investigated to highlight this impairment (Nallegowda et al., 2003). Interestingly, undisturbed upright standing can also provide various insights, as demonstrated by a few studies (Wykman & Goldie, 1989; Trudelle-Jackson & Smith, 2004). This sensorimotor task indeed requires a very precise control of the tiny movements of the body by unceasingly displacing the point of application of the reaction forces.
namely the resultant center-of-pressure ($CP_{Res}$). A biomechanical feature is that the body and its center-of-gravity vertical projection ($CG_v$), because of its considerable inertia, cannot follow these $CP_{Res}$ displacements with precision. As a result, a horizontal gap between these two movements is permanent, the main effect of which is to accelerate the body toward a direction resulting from the respective locations of both positions. The greater this horizontal difference between $CP_{Res}$ and $CG_v$ ($CP_{Res} - CG_v$), the greater the horizontal acceleration communicated to the body and the more difficult it is for the CNS to handle these disruptive motions (Brenière, Do, & Bouisset, 1987).

Because of the bipedal nature of the human upright standing position, various combinations of reaction force patterns under each leg can infer a given $CP_{Res}$ trajectory (Rougier, 2007). Solely focusing on these $CP_{Res}$ movements cannot fully explain the postural strategies developed in asymmetric upright standing. At this level, two kinds of asymmetry need to be differentiated. The first concerns the body weight distribution over the two legs, which is not equal in these patients, at least in the days following surgery. Its effect on undisturbed postural control was emphasized in healthy adults by Genthon and Rougier (2005). The second involves the reaction forces intervening under each support (i.e., the plantar $CP$ trajectories), the patterns displacements of which are not necessarily symmetrical. To distinguish clearly between and highlight these two forms of asymmetry, it is necessary to use a device with two separate force platforms.

In addition, the correlation between the two plantar $CP$ trajectories and the $CP_{Res}$ displacements calculated along both medio-lateral ($ML$) and antero-posterior ($AP$) axes indicate that the $CP_{Res}$ displacements result mainly from hip motions (inferring body weight distribution) and ankle motions (inferring plantar $CP$ movements), respectively (Genthon & Rougier, 2003; Rougier, 2007). The sensorimotor impairment of THA patients is due to the surgical approach involving the cutting of the tendon insertion of the hip rotators and the capsular excision and, therefore, mainly alters the proprioceptive information and the motor execution along the $ML$ axis (Winter, Prince, Frank, Powell, & Zabjek, 1996; Rougier, 2007). Therefore, one can expect that the prosthesis settlement would principally affect the $CP_{Res}$, and, consequently, $CG_v$ and $CP - CG_v$ movements along this specific $ML$ axis. Along these lines, one can also hypothesize that the plantar $CP$ displacements, because of the sound functioning of the ankle joint, should be weakly affected by the surgery. Interestingly, the respective contribution of ankle and hip mechanisms in the generation of $CP_{Res}$ displacements can be objectively assessed through specific indices (Rougier, 2007).

Finally, since THA determines both body weight distribution and sensorimotor impairments, it is necessary to assess the postural control organization and the expected compensatory mechanisms regarding these two aspects. A first approach could consist of involving the same patients preoperatively and comparing their data from before and soon after the surgery. However, it is likely in that case that the body weight bearing asymmetry would have been significantly modified, preventing us to adequately compare the postural strategies. It was indeed shown that the level of asymmetrical body weight distribution conditions variable $CP$ displacements and body movements (Genthon & Rougier, 2005). We, thus, opted to incorporate a control group of healthy adults of similar age required to adopt an identical body weight distribution asymmetry.
Methods

The methodology has been detailed previously in several articles (Rougier & Caron, 2000; Genthon & Rougier, 2005). Only the main points will thus be discussed here.

Experimental Procedure

Fourteen patients undergoing surgery for total hip arthroplasty because of osteoarthritis without contralateral impairment (THA group, 8 men and 6 women), ranging in age from 57 to 85 years (body weight 71.3 ± 13.1 kg; height 1.66 ± 0.07 m; mean ± SD) were included in this study. From a surgical point of view, 10 were operated on by the posterolateral approach and 4 by the anterior approach. In addition, several scores were calculated to assess a clinical profile (Table 1): muscle strength (from 1 to 5); hip range of movement (in degrees) in flexion and extension, in abduction and adduction, and in external and internal rotation; tactile, kinesthetic, and palpesthetic sensitivities (graduated from 0 to 2); pain as measured by a visual analog scale (VAS) graduated between 0 (no pain) and 10 (maximum pain); and autonomy, measured by a functional measure (FA) that evaluates various daily life tasks (scored from 0 [dependence] to 126 [independence]). To be included in the study, subjects had to have an ability to stand up for 1 min and the capacity to understand instructions and execute them adequately. Patients with psychiatric or neurological disorders that could affect balance were not included.

The posturographic and clinical measurements took place 12 ± 3 days after surgery and were processed as soon as the patients entered the rehabilitation department.

The subjects stood barefoot on a double force platform (PF02, Equi+, Aix les Bains, France) and were required to stand still with their arms at their sides. They were asked to stare at a point in front of them and, when instructed, to close their eyes until the end of the trial, when they reopened them. In all cases, three trials lasting 32 s (sample frequency, 64 Hz) were recorded with rest periods between trials of 15 s. This duration constitutes a good compromise between the need to collect data over long periods of time and to reduce the difficulty of the task for some very unstable patients. In addition 32 s at 64 Hz yields 2,048 data points, an optimal number to support Fast-Fourier transform. The feet were externally rotated at 30°, the gap between the heels being 9 cm wide. Postural sway was measured by two rectangular (20 by 35 cm) force platforms, collaterally installed, on which the subjects placed both feet.

The signals issued from the eight dynamometric load cells were amplified and converted without any filtering from analog to digital form before being recorded on a personal computer. The plantar and resultant CP trajectories were then automatically processed in different ways through specific software (Prog02, Equi+, Aix les Bains, France).

These patients were compared with 13 healthy adults (HEA group, 3 men and 10 women) of similar age ranges (65–95 years) and morphologic characteristics (body weight 60.0 ± 9.0 kg; height 1.55 ± 0.10 m). These reference subjects were required to adopt an asymmetrical body weight distribution close to that observed on average for the patients. To this aim, the healthy subjects performed a prelimi-
nary trial run during which oral feedback on body weight distribution on both legs was provided. They had then to perform several trials, as did the THA group, with the same body weight distribution. Some supplementary feedback about this body weight distribution was again given when required, although only between the trials, to avoid any possible interaction with the postural evaluation. The objective was to compare both samples through close body weight distributions.

All subjects belonging to the two groups had given their informed consent in accordance with the guidelines of the local ethics committee.

| Table 1 Characteristics of the Clinical Scores for the Patients Included in the Study |
|---------------------------------|---------|---------|---------|
| Muscular strength (0–5)         | Mean    | SD      | Range   |
| – hip flexors                   | 3.14    | 0.52    | (2–4)   |
| – gluteus medius                | 3.07    | 0.26    | (3–4)   |
| – gluteus maximus               | 3.64    | 0.48    | (3–4)   |
| – hamstring                     | 3.86    | 0.52    | (3–5)   |
| – quadriceps                    | 3.71    | 0.70    | (3–5)   |
| Range of motion (°)             |         |         |         |
| – flexion/extension             | 81.07   | 11.68   | (60–105)|
| – abduction/adduction           | 33.93   | 7.12    | (25–50)|
| – external/internal rotation    | 22.86   | 6.19    | (10–30)|
| Sensitivity (0–2)               |         |         |         |
| – tactile sense                 | 1.93    | 0.26    | (1–2)   |
| – kinesthesia                   | 1.93    | 0.26    | (1–2)   |
| – pallesthesia                 | 1.71    | 0.59    | (0–2)   |
| Pain (0–10)                     | 4.07    | 1.28    | (2–7)   |
| Autonomy FA (0–126)             | 95.50   | 5.18    | (90–110)|

Estimation of the Resultant Center of Pressure ($\text{CP}_{\text{Res}}$), $\text{CG}_v$ and $\text{CP}_{\text{Res}} − \text{CG}_v$ Movements

As stated previously, it is necessary to distinguish the plantar CP displacements involving the left ($\text{CP}_p$) and the right foot ($\text{CP}_r$). The resultant $\text{CP}_{\text{Res}}$ movements were calculated along each ML or AP axis from the left and right plantar CP movements through the following relation (Winter et al., 1996):

$$\text{CP}_{\text{Res}} = \text{CP}_p \times R_y / (R_y + R_y') + \text{CP}_r \times R_y' / (R_y + R_y'),$$

where $R_y$ and $R_y'$ are the vertical reaction forces under the left and right feet, respectively. By definition, $R_y + R_y'$ is constant and, on average, corresponds to the body weight in static conditions. A schematic representation illustrating the successive steps used for the data analysis issued from a double force platform can be seen in Figure 1. The $\text{CP}_{\text{Res}}$ displacements were computed along both ML and AP axes from
the initial left and right plantar CP movements and from the body weight distribution between the supports. For the sake of clarity and to analyze both prosthetic and sound sides in patients, as well as loaded and unloaded supports in controls, this terminology was used throughout this study to characterize the plantar CP from the unloaded ($CP_{uf}$) and loaded ($CP_{lf}$) feet.

Figure 1 — Synopsis of the different steps for calculation and data processing. The calculation methodology can be summarized in three important steps: (a) collection of $CP_{uf}$ and $CP_{lf}$ movements, (b) calculation of the $CP_{Res}$ movements along the two ML and AP axes, (c) estimation of $CP_{Res} - CG_v$ and $CG_v$ movements (from Genthon and Rougier, 2005).
The determination of the respective contribution of the ankle and hip mechanisms was achieved for each trial by computing two “theoretical” \( \text{CP}_{\text{Res}} \) trajectories (\( \text{CP}_{\text{ankle}} \) and \( \text{CP}_{\text{hip}} \)). To be more precise, instead of applying the formula every time, the successive values of the relative body weight applied to each leg and the successive plantar \( \text{CP} \) positions were held constant and replaced by their mean for the whole duration of the trial. Consequently, in addition to the measured \( \text{CP}_{\text{Res}} \) trajectories, two theoretical ones, corresponding to the \( \text{CP}_{\text{Res}} \) trajectories that would have been obtained with the single contribution of ankle and hip mechanisms, were recomputed and then processed in similar ways: \( \text{CP}_{\text{ankle}} \) relates to the contribution of ankle mechanisms to the \( \text{CP}_{\text{Res}} \) trajectories, \( \text{CP}_{\text{hip}} \) relates to the contribution of hip mechanisms to the \( \text{CP}_{\text{Res}} \) trajectories.

An example of \( \text{CP}_{\text{ankle}} \) and \( \text{CP}_{\text{hip}} \) trajectories measured from a single trial is displayed along with \( \text{CP}_{\text{Res}} \) trajectories in Figure 2.

The standard deviations of these trajectories [\( \text{SD} (\text{CP}_{\text{ankle}}) \) and \( \text{SD} (\text{CP}_{\text{hip}}) \)] were then computed and used to assess the respective contribution of the ankle and hip mechanisms as follows:

\[
\text{Contr}_{\text{ankle}} = \frac{\text{SD} (\text{CP}_{\text{ankle}})}{\text{SD} (\text{CP}_{\text{ankle}}) + \text{SD} (\text{CP}_{\text{hip}})}; \\
\text{Contr}_{\text{hip}} = \frac{\text{SD} (\text{CP}_{\text{hip}})}{\text{SD} (\text{CP}_{\text{ankle}}) + \text{SD} (\text{CP}_{\text{hip}})}.
\]

Interestingly, \( \text{Contr}_{\text{ankle}} + \text{Contr}_{\text{hip}} \), with this method being equal to 1, the statistical analysis was processed along each ML and AP axis with one parameter only.

A particularity of the \( \text{CP}_{\text{Res}} \) displacements is that their action upon the \( \text{CG} \) appears, at times, to facilitate its falling motion by increasing the induced horizontal acceleration and, at other moments, to counteract it. Consequently, it seemed interesting to decompose these complex movements into two basic components: the vertical projection of the \( \text{CG} \) over the plane of support (\( \text{CG}_v \)) and the difference between these two movements (\( \text{CP}_{\text{Res}} - \text{CG}_v \)).

Since body sways are being reduced, undisturbed stance maintenance infers a constancy of the momentum of inertia of the body. In this case, \( \text{CG}_v \) movements can thus be deduced from \( \text{CP}_{\text{Res}} \) trajectories, a frequency relation existing between these two variables (Brenière, 1996; Caron, Faure, & Brenière, 1997). It is, therefore,
this principle that was used to estimate the $CG_v$ movements. It is, thus, relevant to consider that $CP_{Res}$ displacements operating over too high frequencies do not infer appreciable $CG$ movements, because of important body inertia. This method represents a generalization of the concept initially developed by Brenière (1996) for more dynamic conditions. The hypothesis is that the body constitutes a low-pass filter, which would explain the loss in amplitude observed between $CP_{Res}$ and $CG_v$ as the frequency increases. In fact, as depicted in Figure 1, the principle consists in multiplying the data, transformed in the frequency domain, through a Fast-Fourier transform (FFT), by the filter mathematically defined by the ratio

$$\frac{CG_v}{CP_{Res}} = \frac{\Omega_0^2}{\Omega_0^2 + \Omega^2},$$

and then to recover the temporal domain by processing an inverse FFT. In this formula, $\Omega_0 = \left(\frac{mgh}{I_G + mh^2}\right)^{1/2}$ ($m$, $g$, $h$, $I_G$: mass of the subject, gravity acceleration, distance from $CG$ to the ground, and momentum of body inertia around the $ML$ or $AP$ axis with respect to the $CG$) corresponds to a biomechanical constant, the natural body frequency (Brenière, 1996), and $\Omega = 2\pi f$ is the pulsation (rad/s).

Because the momentum of inertia differs according to the $ML$ and $AP$ axes, two distinct relationships are used to compute these frequency ratios according to the body weight and height of the subjects (Ledebt & Brenière, 1994).

### Signal Processing

The various movements ($CP_{uf}$, $CP_{lf}$, $CP_{Res}$, $CG_v$, $CP_{Res} - CG_v$) have been studied through several kinds of analysis. The intertrial mean positions of the plantar and resultant trajectories were calculated with regards to a reference 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 (positive $AP$ axis) and to the right (positive $ML$ axis) with respect to these lines. The second method consists of the computation of traditional parameters such as the variances of the successive positions along both $ML$ and $AP$ axes, the areas of an ellipse calculated with a confidence interval (Tagaki, Fujimura, & Suehiro, 1985), and mean velocities (total path lengths divided by trial duration). The last method is based on parameters such as Root Mean Square ($RMS$) and Mean Power Frequency ($MPF$) issued from the Fourier Transform decomposition. According to specific bandwidths for each motion, $RMS$ and $MPF$ were calculated over $0 – 0.5$ Hz for $CG_v$ and $0 – 3$ Hz for $CP_{Res} - CG_v$, $CP_{uf}$ and $CP_{lf}$ respectively. These specific bandwidths were purposely chosen since some previous studies showed that both basic movements were not endowed with similar frequencies (Farenc & Rougier, 2000).

Lastly, to compare the postural strategies between the THA and HEA groups, Mann and Whitney $U$ tests were used. In addition, the comparison between unloaded and loaded feet parameters were assessed through a Wilcoxon test. The level of significance was set for the two unilateral tests at $p < .05$.

### Results

#### Classical Parameters

Our two samples are characterized by identical asymmetrical body weight distributions: $0.42 \pm 0.05$ and $0.42 \pm 0.06$ on the unloaded leg for the THA and HEA
groups, respectively. This result is confirmed by the statistical analysis ($U = 82, p > .05$). As displayed in Figure 3, no difference can be seen between the two groups for mean positions of both $CP_{Res}$ and planar $CP_{uf}$ and $CP_{lf}$ trajectories along both $ML$ and $AP$ axes. In particular, one can see the shifting of the mean $CP_{Res}$ toward the loaded leg.

In contrast, statistically significant results from surfaces with the 90% confidence ellipses are found for $CP_{Res}$, $CG_v$ and $CP - CG_v$ movements. To be more precise, the THA group is characterized by greater territories (ellipse areas) for all movements ($CP_{Res}$: $350.81 \pm 275.7 \text{ mm}^2$ vs $163.83 \pm 114.27 \text{ mm}^2$, $U = 34$, $p < .01$; $CG_v$: $229.74 \pm 199.7 \text{ mm}^2$ vs $104.35 \pm 79.38 \text{ mm}^2$, $U = 44$, $p < .05$; $CP_{Res} - CG_v$: $51.20 \pm 35.54 \text{ mm}^2$ vs $24.25 \pm 15.47 \text{ mm}^2$, $U = 33$, $p < .01$). The results of the plantar $CP$ trajectories reveal a statistically significant difference for the sound-loaded ($366.08 \pm 289.55 \text{ mm}^2$ vs $156.22 \pm 121.57 \text{ mm}^2$, $U = 40$, $p < .01$) and the prosthetic-unloaded leg ($342.61 \pm 422.53 \text{ mm}^2$ vs $169.73 \pm 155.85 \text{ mm}^2$, $U = 54$, $p < .05$).

Finally, the mean velocities measured from the $CP_{Res}$ and $CG_v$ displacements display statistically significant increased values for the THA group, when compared with the HEA one ($19.67 \pm 6.53 \text{ mm/s}$ vs $14.51 \pm 5.32 \text{ mm/s}$, $U = 46$, $p < .05$ and $5.01 \pm 1.84 \text{ mm/s}$ vs $3.41 \pm 1.10 \text{ mm/s}$, $U = 35.5$, $p < .01$, respectively). A difference can be also found for the $CP_{Res} - CG_v$ ($16.96 \pm 5.86 \text{ mm/s}$ vs $12.81 \pm 4.93 \text{ mm/s}$, $U = 56$, $p < .05$), but not for the plantar displacements ($CP_{uf}$: $17.99 \pm 8.99 \text{ mm/s}$ vs $13.50 \pm 6.28 \text{ mm/s}$, $U = 63$, $p > .05$; $CP_{lf}$: $18.21 \pm 6.62 \text{ mm/s}$ vs $13.42 \pm 5.29 \text{ mm/s}$, $U = 58.5$, $p > .05$).

When comparing the prosthetic-unloaded leg to the sound-loaded one, the statistical analysis highlights a difference in the THA group for the averaged absolute position, which is longer for the prosthetic leg along the $ML$ axis ($T = 20$, $p < .01$).

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**Figure 3** — Mean positions ($\pm SD$) of the various plantar trajectories (sound support on the right and prosthetic leg on the left one) and of the $CP_{Res}$ movements in the middle for both groups (THA in black and HEA in gray). Note the greater amplitudes of the trajectories for the THA group and the shift toward the sound leg expressing the body weight asymmetry.
.05). In the referential framework used, this increased distance between the mean plantar CP positions indicates that the CP_{uf} trajectories are more laterally shifted toward the external edge of the foot. No statistically significant result is observed for all parameters measured along the AP axis or for the HEA group along both axes (Figure 3). This result indicates that the degree of forward-leaning is similar for both groups and that the observed different postural behavior cannot be thus explained by this feature.

Ankle Contribution in the CP_{Res} Displacements

The contribution of the ankle mechanisms is enhanced for the THA group along the ML (0.22 ± 0.05 vs 0.17 ± 0.04) and AP axes (0.96 ± 0.03 vs 0.98 ± 0.01). In both cases, the difference between the two groups is statistically significant (ML: U = 55, p < .05; AP: U = 54, p < .05).

Frequency Parameters

All resultant and plantar movements are decomposed through the fast-Fourier transform and give rise to RMS and MPF frequency parameters. This method of analyzing the various displacements allows us to further explain the planar classical parameters (ellipse areas) by specifying whether the effect occurs mainly along one axis or along both.

Plantar CP_{lf} and CP_{uf} Trajectories. As depicted in Figure 4, the RMS computed from the plantar CP displacements of the THA group tend to be increased when compared with the HEA one for both unloaded and loaded legs and along both ML and AP axes. For the CP_{uf} trajectories, the statistical analysis reveals a significant decrease only along the AP axis (U = 46, p < .05), whereas the opposite is true along the ML one (U = 63.5, p > .05). On the other hand, a statistically significant result is found for the CP_{lf} trajectories along both ML (U = 43, p < .01) and AP (U = 37, p < .01) axes. These effects concern all the bandwidths through which the RMS are computed. Therefore, no particular effect is noticed for the MPF parameters for both CP_{lf} (ML: U = 70, p > .05; AP: U = 89.5, p > .05) and CP_{uf} trajectories (ML: U = 90, p > .05; AP: U = 88, p > .05). This means that the mean times for the trajectories to return to a similar position are equal for both the THA and HEA groups. Lastly, the comparison between CP_{lf} and CP_{uf} indicate, in all cases, nonstatistically significant results for both the THA and HEA groups and for both RMS and MPF parameters.

CP_{Res}, CG_{v}, and CP_{Res} − CG_{v} Trajectories. As it can be seen from Figure 5, greater RMS from the resultant CP_{Res} displacements characterize the THA group along this AP axis (U = 36, p < .01). This effect is seen again for the CG vertical projection (U = 46, p < .05) and for the differences CP_{Res} − CG_{v} (U = 36.5, p < .01). A similar result is found along the ML axis for the CP_{Res}, CG_{v}, and CP_{Res} − CG_{v} displacements (U = 52, p < .05; U = 50, p < .05; and U = 36.5, p < .01, respectively). As for the plantar CP displacements, the various mentioned effects concern all the frequency bandwidths, inferring nonsignificant changes for the MPF parameters for CP_{Res} (ML: U = 90, p > .05; AP: U = 86, p > .05), CG_{v} (ML: U = 80.5, p > .05; AP: U = 85, p > .05), and CP_{Res} − CG_{v}, movements (ML: U = 89, p > .05; AP: U = 82, p > .05).
Figure 4 — On the left: mean amplitudes (± SD) of the plantar CP displacements measured for the unloaded and loaded legs along ML and AP axes for both groups (THA in black and HEA in gray). On the right: bar charts illustrating the means (± the standard deviation) for the various parameters aimed at characterizing the frequency spectra. Note the increased RMS for the THA group, especially along the AP axis (**: p < .01; *: p < .05).
Figure 5 — On the left: mean amplitudes (± the standard deviation) of the resultant CP displacements ($CP_{Res}$, $CG_v$ and $CP_{Res} - CG_v$) measured along ML and AP axes for both groups (THA in black and HEA in gray). On the right: bar charts illustrating the means (± SD) for the various parameters aimed at characterizing the frequency spectra. Note the increased RMS for the THA group, especially for the $CP_{Res} - CG_v$ movements (** $p < .01$; * $p < .05$).
Discussion

Our results have highlighted the decreased capacities of patients with total hip arthroplasty to restrain their body motions as much as possible, even though they were instructed to do so, hence confirming the previous studies of Wykman and Goldie (1989). Interestingly, this decreased capacity cannot be accounted for by the asymmetrical distribution of the body weight on both legs since both HEA and THA groups were tested in close weight distributions. Furthermore, by separately investigating the reaction forces exerted under both feet, it can be seen that the mean positions of the points of application of the reaction forces are similarly positioned. This is true when comparing each foot for both groups but also, interestingly, when comparing sound and prosthetic legs of the THA group. However, it is worth highlighting some significant differences in the amplitudes of both plantar CP and resultant trajectories. These effects can either be seen or not along both ML and AP axes, depending on the nature of the trajectories. This is particularly important since it indicates the anatomical levels at which the postural control alterations take place.

Depending on the ML or AP Axis, the Plantar CP Displacements Are Not Used Similarly for Determining the CP_res Displacements

Postural behavior in asymmetrical standing needs to be assessed through the respective contribution of the plantar CP trajectories on the one hand and the body weight distribution variation between the two legs on the other. This principle, through which the resultant CP trajectories can be computed from a double force platform device, thus constitutes the key point to understanding the specific behavior encountered in our subjects. A recent study (Rougier, 2007) has emphasized that a strong link characterizes the plantar and CP_res trajectories along the AP axis. In other words, the greater the plantar CP displacements under the feet, the greater the CP_res displacements, even though large weight variations might occur. In contrast, the degree of correlation between these two movements is considerably lowered when considering the ML axis. This suggests that the CP_res displacements are poorly linked to the plantar displacements and are mainly explained by the weight distribution variations. Because of these differences, which are principally explained by the anatomy of the feet and the characteristics of the ankle joints, the results relative to each of these axes will be discussed separately. It is noteworthy that these principles only apply since the mean positions of the plantar CP trajectories are similar along the AP axis.

Along the AP axis, the best way to obtain reduced CP_res movements is to reduce either plantar CP displacements or, at least, one of them. In fact, the antero-posterior limits of these CP_res displacements are graphically bounded by a straight line passing through the extremes of the plantar CP trajectories. In this organization, the correlation between these displacements and body weight distribution has been demonstrated as being particularly low (Genthon & Rougier, 2003; Rougier, 2007). These AP movements of the plantar CP displacements are the biomechanical consequence of the reaction forces exerted under the feet, which consists, for the most part, in displacements along the longitudinal axis of the feet.
and involve the ankle joint. One must therefore question the possible link between these augmented displacements and the hip arthroplasty. It, thus, seems that it is the control of all of the body motions that is impaired (maybe because of fear linked to the risk of falling), inferring greater levels of cocontractions and, therefore, CG horizontal accelerations and then greater CP displacements. As deduced from its definition, the CG expresses the relative positions of all body segments at a given time. By weakening the control of the upper parts of the body (head, trunk, and arms) relative to the lower ones (legs), total hip arthroplasty might also facilitate these greater CG movements.

Along the ML axis, the general principle observed in bipedal stance to limit the \( CP_{Res} \) excursion is primarily to limit the variations of body weight distribution and secondarily to generate opposite plantar CP displacements on both feet. Indeed, if perfect mirror trajectories in which the subject distributes his body weight equally between the two supports could be performed, an immobility of the \( CP_{Res} \) should be seen. When an asymmetrical body weight distribution is performed, as in this study, the opposite displacements should be inversely proportional to this distribution to reach this goal. In other words, if the body weight is, for instance, distributed in a 1/3 – 2/3 proportion, a 1 cm displacement under the loaded foot must be accompanied by a 3 cm displacement under the unloaded foot to secure null \( CP_{Res} \) displacements along the ML axis.

As shown by the RMS values, the plantar CP displacements for both legs are greater along the ML axis when comparing both the THA and HEA groups. However, because of the specific position of the feet on the platform (feet abducted at 30°), the greater plantar CP displacements along the AP axis infer, in turn, increased movements along the ML one. The symmetry of the plantar CP displacements suggests that these effects upon \( CP_{Res} \) displacements are weak and that the other factor (i.e., the weight distribution variations) concurs in this impairment. This is not really surprising since the \( CP_{Res} \) displacement along this ML axis is mainly achieved by the control of hip abductor and adductor muscles (Winter et al., 1996). This point is also highlighted by the correlation computation from plantar and resultant CP trajectories (Genthon & Rougier, 2003; Rougier 2007) and by the lower \( \text{Contr}_{\text{ankle}} \) index in the THA group. The prosthesis implantation with its sensorimotor impairment would, thus, principally affect the capacity to limit the variations of body weight distribution over the two supports.

From a theoretical point of view, when one mechanism (ankle or hip) is impaired, an adaptive way to restrain the \( CP_{Res} \) displacements along an axis is to position the plantar CP trajectories at different places with respect to both feet. This point was highlighted by Winter et al. (1996) through the tandem stance position. An adaptive strategy for the THA patients to lower \( CP_{Res} \) displacements along the ML axis could, therefore, be to modify their foot position or the pressure distribution under each foot. As indicated by the present data, such an adaptive strategy was not seen in the present case, hence, suggesting that the observed increased \( CP_{Res} \) movements along the ML axis do not really disturb balance control and or more ample \( CP_{Res} \) displacements along the AP axis is not desirable.

Lastly, the statistical analysis, by revealing increased levels of significance between the groups for the trajectories issued from the more loaded leg, emphasizes the major role played by this a priori sound support in impaired postural control. Interestingly, this specificity also appears to apply to other asymmetrical patients.
such as hemiplegics (Genthon et al., in press) and amputees (Rougier & Genthon, 2005).

**THA Patients Display Large $CP_{Res}$ Displacements Partly Affecting Body Movements**

The biomechanical principles for controlling upright standing indicate that greater $CP_{Res}$ displacements do not necessarily induce greater $CG$ movements. As explained previously, a frequency relationship exists between these two movements because of the moment of inertia of the body. As demonstrated by Brenière (1996), the higher the $CP_{Res}$ frequency, the smaller its effect upon the $CG$ movement. For the present data, the augmented $CP_{Res}$ displacements are explained by both greater $CG$, and $CP_{Res} - CG$ basic movements along both $ML$ and $AP$ axes. However, from a statistical point of view, one may note that the greater differences are observed for the $CP_{Res} - CG$ displacements. As emphasized in the backward-forward leaning of the body (Rougier, Burdet, Farenc, & Berger, 2001), these $CP_{Res} - CG$ trajectories constitute a fair expression of the neuromuscular means called into play for controlling the $CG$ movements. This link is mainly explained by the fact that this greater activity induces greater horizontal $CG$ accelerations, inferring greater forces to be generated to handle the body motion. The reduced effects upon the $CG$ movements, when compared with those of the $CP_{Res} - CG$, suggest that the THA patients are able to develop strategies aimed at minimizing these disturbing effects.

To conclude, analyzing the strategies developed to secure balance during undisturbed upright stance has provided various insights that can mainly be explained by the global sensorimotor impairment due to surgery and prosthetic insert. As mentioned in the presentation of the patients, all of them suffered from osteoarthritis before the surgery. One can, thus, assume that numerous reasons, including pain, skeleton deformity, limited joint mobility, muscular amyotrophy, or impaired proprioception may also contribute to the sensorimotor impairment, and, thus, the postural instability that was likely observable before the surgery and still present 12 days later. Therefore, one may wonder what is the role of the osteoarthritis and the role of the surgery itself in the clinical deficits and postural control organization. Although two surgical approaches were present in our group, the reduced sample in each case prevents us from highlighting any possible difference. Future investigations could, thus, be to compare the postural behaviors induced by anterior and posterolateral approaches and before and after the surgery.

**References**


