Undisturbed Upright Stance Control in the Elderly: Part 2. Postural-Control Impairments of Elderly Fallers

L. Berger  
Laboratoire de Modélisation des Activités Sportives  
Université de Savoie, France

M. Chuzel  
Centre hospitalier general  
Belley, France

G. Buisson  
Centre hospitalier general  
Aix les Bains, France

P. Rougier  
Laboratoire de Modélisation des Activités Sportives  
Université de Savoie, France

ABSTRACT. A common way of predicting falling risks in elderly people can be to study center of pressure (CP) trajectories during undisturbed upright stance maintenance. By estimating the difference between CP and center of gravity (CG) motions (CP – CG1), one can estimate the neuromuscular activity. The results of this study, which included 34 sedentary elderly persons aged over 75 years (21 fallers and 13 nonfallers), demonstrated significantly increased CG1 and CP – CG1 motions in both axes for the fallers. In addition, the fallers presented larger CG1 motions in the mediolateral axis, suggesting an enlarged loading–unloading mechanism, which could have reflected the adoption of a step-initiating strategy. As highlighted by fractional Brownian motion modeling, the distance covered by the CP – CG1 motions before the successive control mechanisms switched was enhanced for the fallers in both axes, therefore increasing the risk that the CG would be outside of the base of support.

Key words: center of gravity, center of pressure, elderly, fallers, postural control, sedentary

Fall-related injuries represent a major threat to functional ability and quality of life of the elderly. Approximately 30% of the elderly population experiences a fall each year (Tinetti & Ginter, 1988). The risk of falls increases with age over 60 years and is greater in women than in men (Wild, Nayak, & Isaacs, 1981). Consequently, falls in older people have a great impact on the healthcare system. Factors traditionally viewed as contributing to an increased risk for falls have been categorized as extrinsic (i.e., those associated with environmental features) and intrinsic (i.e., those internal to the individual). Intrinsic factors include changes in muscular strength associated with the normal aging process (Jette, Branch, & Berlin, 1990), decreases in joint flexibility (Hughes, Dunlop, Edelman, Chang, & Singer, 1994), and decline of visual perception, vestibular function, and somatosensory sense (Manchester, Woollacott, Zederbauer-Hyton, & Marin, 1989). The ineluctable decrease in muscular strength with age is associated with a greater degree of inactivity. Therefore, we characterized the participants in this study according to their levels of inactivity, as deduced from the distance they walked daily and the similarity in their degree of immobility.

Overall, intrinsic factors do not provide a sufficient explanation for the falls because other deficits in the postural system have been reported. Those deficits include changes in the spatiotemporal sequencing of the muscles called into play for reacting to a loss of balance (Horak, Shupert, & Mirka, 1989; Woollacott, Shumway-Cook, & Nashner, 1986), a decreased ability to appropriately select sensory information (Woollacott et al.), and an increased dependency on visual cues (Perrin, Jeandel, Perrin, & Béné, 1997). Visual information is indeed crucial for elderly individuals; for that reason, investigators, in most cases, have evaluated postural control in the elderly while the participants’ eyes are open.

To our knowledge, few investigators who have performed measurements of undisturbed upright stance have suggested that larger center of pressure (CP) areas are associated with an increased risk of falling (Balah, Corona, Jacobson, Enrietto, & Bell, 1998; Lord, Rogers, Howland, & Fitzpatrick, 1999). Moreover, Maki, Holliday, and Topper (1994), who reported an axial specificity of center of pressure (CP) trajectories, considered that the CP displacements in the mediolateral (ML) axis constitute a fair predictor of the risk of falling.

An investigation of impaired balance necessitates recourse to a method that provides additional information regarding

Correspondence address: L. Berger, Laboratoire de Modélisation des Activités Sportives, Université de Savoie, Domaine Scientifique de Savoie-Technolac, F 73 376 Le Bourget du Lac Cedex, France. E-mail address: laetitia.berger@univ-savoie.fr

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the control processes and the biomechanical changes. Using the undisturbed upright stance maintenance paradigm, for example, one can dissociate measurements of the displacements of CP—which correspond to the successive points of application of the resultant ground reaction forces—into two elementary superimposed components (Rougier, 2003; Rougier, Burdet, Farenc, & Berger, 2001; Rougier & Farenc, 2000): the horizontal motions of the center of gravity (CGh) and the differences between the CP and the vertical projection of the CG (CP – CGv). The difference CP – CGv is proportional to the horizontal CG acceleration as long as upright standing can still be modeled as an inverted pendulum (Brenière, Do, & Bouisset, 1987), and it constitutes a good expression of the muscular activity. Previous investigators have indeed proposed that the CP – CGv frequency decomposition constitutes a fair expression of the resultant ankle stiffness (Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998). The CGh motions express the body sway and, consequently, the net postural performance. In addition, CGh and CP – CGv motions can be advantageously modeled as fractional Brownian motion (fBm; Mandelbrot & Van Ness, 1968), which enables one to assess the nature and the number of processes involved in their control. Applied to those elementary motions, fBm modeling reveals for the CP displacements (Collins & De Luca, 1993; Rougier, 1999) that two control mechanisms operate successively: one over the shortest time intervals Δt and one over the longest Δt. By objectively evaluating their spatiotemporal limit, one can now disentangle the respective effects, induced, for instance, by a particular physical disability, on resultant ankle muscular stiffness, on the one hand, and on body sway, on the other hand, as deduced from the CP – CGv and CGh motions, respectively. One of the main virtues of that method is its adaptability for the study of nonstationarity of CP (Riley, Balasubramaniam, & Turvey, 1999). Another main advantage of that method is the possibility of suppressing that bimodal organization by using a surrogate data method to remove the time correlation between past and future increments (Rougier & Caron, 2000; see Berger, Chuzel, Buisson, & Rougier, 2005, Part 1 in this issue).

With those particular modeling techniques, Berger et al. (2005) demonstrated that elderly participants elevate the horizontal acceleration communicated to the CG and the mean distance covered by the CG until they have to initiate a corrective process. Elderly individuals mobilize higher neuromuscular energy than do young adults and seniors. Consequently, the origin of the balance impairment could be partly the degree to which the participants’ sedentary nature induces less muscular strength. In parallel, as compared with younger individuals, the elderly are capable of maintaining a good quality of control, which permits them to maintain a relative ability to stabilize their CG in the base of support, but solely in the AP axis. However, that ability appears to be less than that of younger individuals. In the ML axis, the larger CP displacements may reveal that the elderly select a loading–unloading process aimed at facilitating a step-initiating strategy so that they can counteract the ongoing disequilibrium by suddenly enlarging the base of support.

Our aim in this study was thus to pinpoint and compare the behavioral characteristics, for a similar level of immobility, of elderly fallers and elderly individuals who have not experienced falls. In particular, our goal was to highlight the reasons that certain participants are more or less able to stabilize their body motions, and consequently their CG, than others. Complementarily, we also questioned whether the neuromuscular activity, as deduced from CP – CGv motions, is concomitantly affected and whether possible effects are similar for ML and AP axes.

Materials and Method

Participants

Participants aged over 75 years who volunteered for this study lived in a community dwelling. We contacted them, with agreement of their treating doctor, and they gave their informed consent before the study. For all participants, we performed the experiments in the community dwelling in the presence of the physical therapist.

We excluded individuals who were receiving neuroleptic medication or who presented with various disabilities, such as cerebrovascular accidents, vestibular pathology, severe cognitive impairment, neurological pathologies (Alzheimer’s or Parkinson’s diseases), knee prosthesis, joint fusion, neuropathy and acute pain, and visual impairments. We checked those criteria via a standardized questionnaire, in which we asked participants to estimate the number of falls over the last year. An elderly person was considered a faller (F) if he or she reported one or more falls in the last 12 months that were not related to a known intrinsic event (i.e., acute medical illness) or an overwhelming hazard (i.e., ice). The responses to the questionnaire enabled us to estimate the participants’ inactivity level and difficulties in daily life. All the retained participants in this group were highly sedentary because they all walked less than 1 km/day. In addition, some of the fallers walked less than 300 m/day, going only to the dining room and minimizing the distance traveled in the home, and had some difficulties, for example, going up and down stairs.

In all, 34 participants ranging in age between 75 and 96 years (9 men and 25 women) gave their informed consent and were included in this study. Among them, 21 (6 men and 15 women) were considered fallers (F), whereas 13 (3 men and 10 women) had not experienced any falls (NF).

Experimental Procedure

The participants stood barefoot on a triangular force platform (PF01, Equi+, Aix les Bains, France), in a position with feet abducted at 30° and heels separated by 9 cm, with their arms at the sides and with their eyes open. We asked them to decrease the amount of body sway as much as possible.

We amplified and converted the signal that issued from the dynamometric load cells from analogue to digital form.
before we recorded it on a personal computer. As we discuss later, the CP trajectory was then automatically processed in different ways via a specific software program (ProgO1, Equi+, Aix les Bains, France). The horizontal CP trajectory was decomposed along mediolateral (ML) and anteroposterior (AP) axes. The experiment included, for each participant, three trials of 32 s (sampled at 64 Hz) and a rest period of at least similar duration between each trial.

Data Analysis

The method used for the signal processing has been reported previously (Rougier, 2003; Rougier et al., 2001; Rougier & Caron, 2000; Rougier & Farenc, 2000). This method is identical to that reported in Berger et al. (2005).

Estimation of CGh and CP – CGv motions. As mentioned before, we determined CGh and CP – CGv motions from the complex CP trajectories. We used the biomechanical relationship in the frequency domain between the amplitude ratio of the CGh and the CP trajectories (CGh/CP) to estimate the CGh and, consequently, the CP – CGv motions (Brenière, 1996). The amplitude ratio constitutes a low-pass filter, which we calculated accurately for each participant by taking into account some anthropometric parameters (height and body weight) so that we could assess the moments of inertia of the body around the ML and AP axes (see Figure 1, part 1).

Signal processing. In a first approach, we analyzed CGh and CP – CGv motions through classical parameters such as the area covered, the mean velocity, and the variances for ML and AP axes for both motions. Complementarily, we used a frequency approach and a mathematical model termed fractional Brownian motion (fBm) on the various trajectories. In the former approach, one computes two parameters, root mean square (RMS) and median frequency (MF), to characterize the mean frequency decompositions of the elementary motions on specific bandwidths.

![FIGURE 1. Spectral mean decomposition for each mediolateral (ML) and anteroposterior (AP) axis characterizing the entire sample population for both center of gravity horizontal motions (CGh) and CP – CG motions for fallers and nonfallers. Note the differences in amplitudes between the two groups for both CGh and CP – CG motions and for both axes. CP = center of pressure; CGh = vertical center of gravity motions.](image-url)
RMS is proportional to the mean amplitude independently of the frequency distribution, whereas MF expresses the respective contribution of lowest and highest frequencies in the amplitudes and thus characterizes the central tendency. In the latter approach, CGh and CP – CG motions can be advantageously modeled as fBm (Mandelbrot & Van Ness, 1968); that model allows one to evaluate the relative contribution of the random processes in the control mechanisms involved in each motion. Used initially for studying CP trajectories (Collins & De Luca, 1993), this model was recently extended to the elementary CGh and CP – CG motions in various conditions (Rougier, 2003; Rougier et al., 2001; Rougier & Farenc, 2000). All the steps of the data analysis are detailed in Figure 2 of Berger et al. (2005). It is worth noting that variograms computed for such experiments classically display two successive portions, either for both the ML and the AP axes or for both the CGh and the CP – CG motions, indicating that a somewhat deterministic scaling regime precedes or succeeds a completely random one. Thus, for each ML and AP axis, we objectively extracted two scaling exponents (indexed as short and long latencies: H₃ and H₀) as well as the spatiotemporal coordinates of the transition point (<Δr² > and Δt) by comparing experimental and average random variograms. Specifically, the maximal distance between an experimental variogram computed from the unfiltered CP trajectories and the straight line characterizing a complete random walk is thought to correspond to the Δt coordinate of the transition point. Through those spatiotemporal coordinates, one can in fact express the mean distance covered by the motion (<Δr² >) at the onset of the corrective process and the mean time (Δt) passed until that onset.

By applying fBm modeling, one can demonstrate that, during the shortest Δt, the control is primarily focused on the CP – CG motions, that is, the horizontal forces applied to the CG, whereas during the longest Δt, the control concerns only the CG motions. As a result, when one elementary motion is controlled, the other is characterized by a complete random process (Rougier & Caron, 2000).

Last, to compare the elderly F and NF, we performed a Mann–Whitney U test; we set the first level of significance at p < .05.

### Results

The characteristics of the population (mean age, height, and weight) are presented in Table 1. No difference was observed between the F and NF groups for any of those variables.

### Classical Parameters and Frequency Parameters

The areas covered by both CGh and CP – CG motions appeared to be significantly increased for the fallers as compared with those of the nonfallers. That feature, which is displayed in Table 2, was true for all motions (U = 52, p < .01, for CGh motions; U = 69, p < .01, for CP – CG motions). On the other hand, only the mean velocities for CP – CG motions appeared significantly different between the two groups (U = 69, p < .05; see Table 2).

As revealed in Table 3, the frequency parameters displayed statistically significant differences between the F and the NF groups, and in opposite ways for ML and AP axes. In the ML axis, some significantly larger amplitudes were similarly encountered for both CP – CG and CG motions (13% and 15%, respectively; Us = 52 and 49, ps < .01, for CP – CG and CG motions, respectively). As can be seen in Figure 1, which represents the average spectra for both the F and NF groups, those increases, when observed, affected the totality of the bandwidth for the CP – CG motions. As a result, no statistically significant difference was found for those motions between the two groups for MF (Table 2). On the contrary, a significant decrease of the MF was observed for the CGh motions (U = 75, p < .05).

In the AP axis, a significant increase in the amplitudes of the spectra was noticed for CP – CG and CG motions (Us = 77 and 81, ps < .05, respectively; see Table 2); the increase affected the totality of the bandwidth for both motions. Consequently, the MF remained unchanged for those motions.

### Parameters That Resulted From fBm Modeling

As shown in Table 2, on the one hand, the degree to which the CP – CG motions were controlled over the shortest Δt (H₀) did not present any statistically significant difference between the F and NF groups and between the ML and the AP axes. On the other hand, the temporal coordinates Δt of the transition points in the ML and AP axes were similar for

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**TABLE 1. Anthropometric Characteristics of Participants**

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Participants with falls (n = 21)</th>
<th>Participants without falls (n = 13)</th>
<th>Independent t test</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (yr)</td>
<td>M 85.4, SD 9.3, Range 75–94</td>
<td>M 84.3, SD 7.1, Range 75–96</td>
<td>ns</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>M 158.9, SD 8.6, Range 150–174</td>
<td>M 160.0, SD 8.4, Range 146–178</td>
<td>ns</td>
</tr>
<tr>
<td>Weight (kg)</td>
<td>M 65.0, SD 17.1, Range 40–112</td>
<td>M 65.1, SD 14.5, Range 40–81</td>
<td>ns</td>
</tr>
</tbody>
</table>

*Note. All 34 participants were sedentary. SD = standard deviation; ns = nonsignificant.*
### TABLE 2. Classical and Frequency Parameters for CP – CGᵥ and CGₕ Motions in Both Axes for All Participants

<table>
<thead>
<tr>
<th>Participant</th>
<th>Area (mm²)</th>
<th>Mean velocity (mm/s)</th>
<th>RMS ML (mm)</th>
<th>MF ML (Hz)</th>
<th>RMS AP (mm)</th>
<th>MF AP (Hz)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fallers</td>
<td>32 (30)</td>
<td>0.11 (0.05)</td>
<td>1.08 (0.20)</td>
<td>0.19 (0.08)</td>
<td>1.17 (0.19)</td>
<td></td>
</tr>
<tr>
<td>Nonfallers</td>
<td>12 (9)</td>
<td>0.07 (0.02)</td>
<td>1.11 (0.20)</td>
<td>0.14 (0.08)</td>
<td>1.23 (0.13)</td>
<td></td>
</tr>
<tr>
<td>p value</td>
<td>**</td>
<td>**</td>
<td>ns</td>
<td>**</td>
<td>ns</td>
<td></td>
</tr>
</tbody>
</table>

**CP – CGᵥ motions**

| Fallers     | 262 (210) | 5.9 (2.1)   | 0.95 (0.37) | 0.14 (0.05)| 1.41 (0.67)| 0.16 (0.04)|
| Nonfallers  | 108 (56)  | 5.4 (2.1)   | 0.62 (0.17) | 0.17 (0.05)| 1.05 (0.30)| 0.18 (0.05)|
| p value     | **         | ns          | **          | ns         | ns          |

**CGₕ motions**

*Note.* Data are expressed as mean (± standard deviation). RMS = root mean square; MF = median frequency; ML and AP = mediolateral and interoposterior, respectively.

*p < .05, **p < .01. ns = nonsignificant.

### TABLE 3. Fractional Brownian Motion Parameters for CP – CGᵥ Motions in Both Axes for All Populations

<table>
<thead>
<tr>
<th>Population</th>
<th>Δt ML (s)</th>
<th>&lt;Δ²&gt; ML (mm²)</th>
<th>H₀ ML</th>
<th>Δt AP (s)</th>
<th>&lt;Δ²&gt; AP (mm²)</th>
<th>H₀ AP</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fallers</td>
<td>0.60 (0.24)</td>
<td>2.70 (2.60)</td>
<td>0.74 (0.08)</td>
<td>0.48 (0.26)</td>
<td>8.48 (8.50)</td>
<td>0.79 (0.15)</td>
</tr>
<tr>
<td>Nonfallers</td>
<td>0.59 (0.20)</td>
<td>1.07 (0.68)</td>
<td>0.71 (0.08)</td>
<td>0.41 (0.17)</td>
<td>3.49 (3.97)</td>
<td>0.75 (0.14)</td>
</tr>
<tr>
<td>p value</td>
<td>ns</td>
<td>*</td>
<td>ns</td>
<td>ns</td>
<td>ns</td>
<td>ns</td>
</tr>
</tbody>
</table>

*Note.* ML = mediolateral; AP = anteroposterior; Δt = time interval; <Δ²> = mean square distance covered by a given point; H₀ = shortest Δt; CP – CGᵥ motions = difference between CP and CGᵥ motion, where CGᵥ = center of gravity motion in the vertical direction. Data are expressed as mean (± standard deviation).

*p < .05. ns = nonsignificant.

### TABLE 4. Fractional Brownian Motion Parameters for CGₕ Motion in Both Axes for All Populations

<table>
<thead>
<tr>
<th>Population</th>
<th>&lt;Δ²&gt; ML (mm²)</th>
<th>H₀ AP</th>
<th>&lt;Δ²&gt; AP (mm²)</th>
<th>H₀ AP</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fallers</td>
<td>4.88 (6.72)</td>
<td>0.11 (0.11)</td>
<td>8.22 (15.19)</td>
<td>0.10 (0.11)</td>
</tr>
<tr>
<td>Nonfallers</td>
<td>2.41 (1.65)</td>
<td>0.06 (0.10)</td>
<td>3.88 (3.48)</td>
<td>0.06 (0.13)</td>
</tr>
<tr>
<td>p value</td>
<td>ns</td>
<td>ns</td>
<td>ns</td>
<td>ns</td>
</tr>
</tbody>
</table>

*Note.* ML = mediolateral; AP = anteroposterior; Δt = time interval; <Δ²> = mean square distance covered by a given point; H₀ = longest Δt; CGₕ motions = center of gravity motion in the horizontal direction. Data are expressed as mean (± standard deviation).

*p < .05. ns = nonsignificant.

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both groups. However, the mean spatial coordinates <Δ²> for the CP – CGᵥ motions were significantly increased in both axes for the F group (Us = 80 and 76, ps < .05, in ML and AP axes, respectively), whereas those of the CGᵥ motions appeared unaffected (Figure 2). Finally, the efficiency of the corrective control involving the CGₕ motions during the longest Δt, expressed through the scaling regimes H₀, did not separate the postural behaviors of both groups on either the ML or the AP axis (Table 4).

**Discussion**

At the Onset of the Corrective Process, Elderly Fallers Have to Counteract Larger CG Accelerations

The fallers showed increased amplitudes of both CP – CGᵥ
and CGₘ motions in both axes and, consequently, in the CP motions. Therefore, our results confirm those of previous studies showing increased areas of CP trajectories and mean velocities in elderly fallers compared with those in elderly nonfallers (Baloh et al., 1998; Baloh et al., 1994; Lord et al., 1999). On the one hand, from a biomechanical point of view, the increase in the amplitude of the CP – CGₘ motions revealed some increase in the horizontal accelerations communicated to the CG (Brenière et al., 1987). On the other hand, using the principle of progressive recruitment of the motor units (Hennemann, Somjen, & Carpenter, 1965), one can explain some larger amplitudes of the CP – CGₘ motions as an enhancement of motor activity. Consequently, the increase of the amplitude of the CP – CGₘ motions, as revealed by the RMS parameter, can be interpreted as expressing the development of greater strength. Fallers, as compared with nonfallers, would thus demonstrate increased neuromuscular activity in quiet stance. Recently, Daubney and Culham (1999) showed that fallers are characterized by a force decrease in lower extremity muscles, especially in ankle dorsiflexor and hip extensor muscles. However, a force decrease of all muscles is generally observed in elderly individuals, and muscular impairment is a characteristic of aging (Jette et al., 1990). Both groups in the present study shared a high degree of inactivity and consequently a decrease in muscular strength. As a result, we suggest that the fallers’ capacity to generate force through the musculature could be somehow insufficient to enable them to develop the minimal strength necessary to adequately perform the correction and thus to maintain postural stability. That was especially true for the population of fallers, who were found to mobilize more neuromuscular resources. However, not all elderly participants fall, and the decline of strength of the muscles cannot be the sole explanation for falls.
A Shift in the Frequency Distribution of the CG\textsubscript{n} Motions in the ML Axis May Constitute a Fair Predictive Measure of the Risk of Falling

As revealed by the frequency analysis, elderly fallers exhibited in the ML axis a shift of the frequency distribution of CG\textsubscript{n} motion amplitudes to lower frequencies, whereas similar results were not observed in the AP axis. Maki et al. (1994), in a study designed to evaluate the relationship between postural control and falls, found similarly that a shift of the mean frequency of the CP motion in the ML axis characterized the faller participants. The augmented MF observed in the ML axis signifies, by definition, a diminution of the period needed for the CG to return to a similar position. For the ML axis, the primary stabilizing response occurs generally at the hip level (Maki, Holliday, & Topper, 1992), which can induce, in turn, a loading–unloading mechanism. It is evident that elderly fallers, more than elderly nonfallers, choose compensatory strategies such as preparatory limb unloading. Those results were also noticed in a recent study by Kemoun, Watelain, Defevre, Guiei, and Destee (2002), who investigated the evolution of the postural patterns connected with falls and found that a majority of fallers systematically use many hip strategies.

The Distances Covered by the CP – CG\textsubscript{n} Motions in Both Axes Before the Corrective Control Mechanisms Intervened Were Enhanced for the Fallers

The fBM modeling provided some insight into the degree to which both CP – CG\textsubscript{n} and CG\textsubscript{n} motions are controlled; the results were identical for the two groups. Because the temporal coordinates of the transition points were not affected, the mean square distances covered by the CP – CG\textsubscript{n} motions \(<\Delta x^2>\) before the successive control mechanisms switched were enhanced for the fallers in both axes. In addition, those augmented spatial coordinates of the transition points observed in both axes served mainly to explain the increased RMS noticed for the CG\textsubscript{n} motions. Those increases indicate that some larger horizontal accelerations are communicated to the CG. Fundamentally, maintenance of the CG\textsubscript{n} over the base of support can be considered only if the postural correction is initiated before too long a distance from the equilibrium position has been covered by the CG\textsubscript{n}. Consequently, fallers may present a major risk of going outside of the base of support.

It is widely accepted that most falls are the result of both extrinsic and intrinsic factors and are not caused by a single factor. Rather, they are often the result of multiple factors that occur simultaneously. The efficiency of postural control depends on the limited time required to completely recover equilibrium (Horak et al., 1989). That time increases with age (Horak et al.; Blaszczyk, Lowe, & Hansen, 1994) and causes substantial delays in control. One cannot exclude the possibility that fallers need more time than nonfallers to initiate the corrective processes. Nevertheless, that postural strategy may not be available only because the CG sways too close to the limits of the stability border. The distances covered by the CG\textsubscript{n} motions before the corrective process could come close to those limits. Consequently, sometimes the fallers cannot adapt the corrective process when the perturbation occurs.

The correction process necessitates a rapid and precise perception of the body movements. An age-related decrease in sensory input is a constant. On the one hand, it is possible that there are variable levels of somesthetic, vestibular, and vision sensitivities between the present participants and other groups. We estimated only the vision of our participants. On the other hand, no abnormal loss of vestibular, tactile, and proprioceptive functions were noticed by medical therapists. However, the degree of loss of sensory input could be different between the present groups. A possible explanation for the enhanced mobilization of muscular strength could be a higher level of proprioceptive input. Consequently, that feature may correspond to an adaptation to the loss of sensory information. The results of this study seem to open a possible means to an understanding of some age-related falls. As deduced from the CP – CG\textsubscript{n} motions, it was evident that fallers, more than healthy nonfalling elders, mobilize more neuromuscular resources and involve the ML and AP axes in similar ways to maintain undisturbed upright posture.

Our two groups’ low level of activity may favor a decrease in muscular strength and, consequently, lessen their ability to mobilize muscle. The faller population could have less sensory input than the nonfallers, hence explaining in part the balance impairments. Another explanation unexplored in this study was the fear of falling, a factor that is significantly associated with a greater variation in balance performance (Carpenter, Frank, & Silcher, 1999). In addition, many fallers are indeed afraid to fall, and fear is known to induce a decrease in voluntary mobility.

Evidently, some exercises may improve balance and mobility functions (Crilly, Willems, Trenholm, Hayes, & Delaquerrière-Richardson, 1989; Hu & Woollacott, 1994). In particular, some exercises aimed at reinforcing body perception, such as mirror feedback (Rougier, 2002), whose main effect, in healthy adults, is a decrease of the CP – CG\textsubscript{n} motions in the AP axis, must be considered as a possible rehabilitation tool for elderly fallers.

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