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Control of Equilibrium in Humans: Sway over Sway

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In humans, the postural control of a segment or the whole body about a reference position is achieved by passive and active restoring forces applied to the system under control. Under this rationale, the control of whole-body posture during upright standing has been modeled as an inverted pendulum oscillating about a fixed position. This simple representation has been very useful for understanding many aspects of human postural control. However, some behaviors observed during upright standing are not well captured by this representation. For example, we conducted a series of studies on natural (unconstrained) prolonged (several minutes) upright standing and showed that individuals tend to oscillate about a moving reference position (Duarte and Zatsiorsky 1999; Duarte et al. 2000; Duarte and Zatsiorsky 2000; Duarte and Zatsiorsky 2001; Freitas et al. 2005b; Prado and Duarte 2009).

In fact, there are no mechanical or neural constraints requiring that humans regulate their upright posture around a reference position somewhat aligned with the vertical axis. An alternative idea is that humans simply adopt a strategy to maximize the safety margin for falling. For example, Slobounov and colleagues (1997) have proposed that we regulate our upright posture by maximizing the time the body center of pressure (COP) would take to contact the stability boundaries at any instant, given the instantaneous position, velocity, and acceleration of the COP at that instant (this time was termed the *virtual time to contact*). By maximizing the virtual time to contact, a standing person would avoid a fall. Although it remains to be shown to what extent virtual time to contact is incompatible with posture control around a reference point, this theory has not been disproved.

The concept of COP is very useful for understanding the regulation of postural control. The COP expresses the position of the resultant vertical component of the ground reaction force applied to the body at the ground surface. The COP is a two-dimensional position dependent on the acceleration of the body and its segments (because it is related to the forces applied to the body). A related concept to COP is the center of gravity (COG). The COG is defined as a specific point in a system of particles (or segments for the human body) that behaves as if the weight of all particles were concentrated at that point. From simple mechanics, if the vertical projection of the COG (COGv) steps out of the base of support (the area at the floor that circumscribes the region of contact of our body with the ground, e.g., for the bipedal posture, it is the area circumscribing the feet), the body will be unable to apply restoring forces to maintain the upright posture. In this sense, the contour of the base of support can be viewed as the stability boundaries for controlling posture.

In fact, the limits of stability standing humans are able to use are smaller than the physical limits given by the contours of the feet. Figure 10.1 presents mean values for the limits of the base of support, limits of stability ~~the~~ adults can voluntarily reach during standing, COP area of sway during prolonged natural (unconstrained) standing for 30 minutes, and COP area of sway during standing still for 40 seconds. Figure 10.1 shows that, while standing still, humans occupy a very small area of the base of support and that during unconstrained standing this area is much larger. With regards to the amount of sway produced during standing, in general, it is assumed that more sway means more instability and is an indication of a deteriorated posture control system. This rationale is based on many experiments on aging and pathological conditions that showed increased sway in those conditions (see for example, the reviews of Horak et al. 1989; Bonnet et al. 2009). However, this is not always the case. Patients with Parkinson disease in some cases demonstrate reduced postural sway compared to elderly adults, despite the fact that patients with Parkinson disease do present severe problems of postural control (Romero and Stelmach 2003). Another proposition for postural control is that at least part of the sway during upright posture is, in fact, an intentional sway (Riccio et al. 1992; Riley et al. 1997; Riley and Turvey 2002; Stoffregen et al. 2005; Bonnet et al. 2009). In this view, more sway does not necessarily imply more instability; neither it is an indication of a deteriorated posture control system. In addition, although elderly persons, when asked to stand as still as possible for a short period of time, commonly show increased postural sway during standing compared to younger persons, elderly persons show the opposite behavior during prolonged unconstrained standing (Freitas et al. 2005b).

In this chapter, we briefly review the control of equilibrium in humans during quiet standing and findings about prolonged unconstrained standing, and we discuss the implications of these findings for understanding the control of equilibrium in humans. But first we describe how postural sway can be evaluated. Throughout this chapter, we employ biomechanical principles to understand the control of equilibrium. The use of biomechanics to

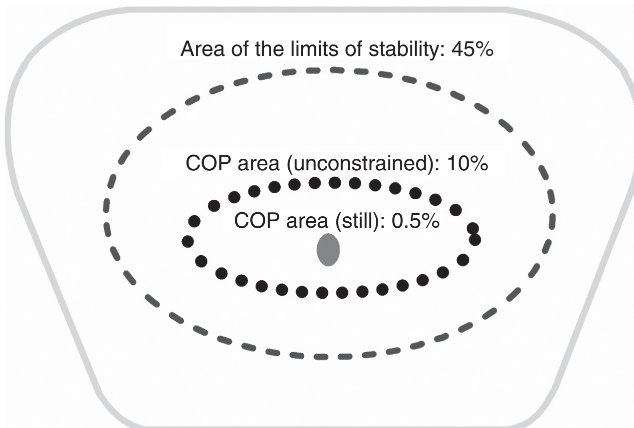

Figure 10.1 Mean values for the limits of the base of support (*solid line*), limits of stability individuals can voluntarily reach during standing (*dashed line*), center of pressure (COP) area of sway during prolonged unconstrained standing for 30 min (*dotted line*), and COP area of sway during standing still for 40 s (*filled area in the center*).
Data adapted from Duarte, M., W. Harvey, and V.M. Zatsiorsky. 2000. Stabilographic analysis of unconstrained standing. *Ergonomics* 43: 1824–39; Duarte, M., and V.M. Zatsiorsky. 2002. Effects of body lean and visual information on the equilibrium maintenance during stance. *Experimental Brain Research* 146: 60–69; Freitas, S.M., J.M. Prado, and M. Duarte. 2005a. The use of a safety harness does not affect body sway during quiet standing. *Clinical Biomechanics (Bristol, Avon)* 20: 336–39; and Freitas, S.M., S.A. Wieczorek, P.H. Marchetti, and M. Duarte. 2005b. Age-related changes in human postural control of prolonged standing. *Gait & Posture* 22: 322–30, all with permission of their respective publishers.

understand the control of locomotion and the connection between motor control and biomechanics in general are addressed in the other two chapters of this section.

QUANTIFICATION OF POSTURAL SWAY DURING STANDING

Before 1950, postural sway was studied mainly by recording head oscillation, a method that is called *ataxiography*. Since then, ataxiography was almost completely forgotten and the recording of ground reaction forces and COP displacement became the prevailing approach. The quantitative evaluation of body sway via force recordings is called *posturography*, which has been divided into static posturography, when the postural control of a person is evaluated by asking that person to stand as still as possible, and dynamic posturography, when the postural responses to a perturbation applied to the person are evaluated. The most frequent measurement used in posturography is COP displacement, which can be easily measured using a force plate. The most common task used for the evaluation of postural control is the quiet standing task, in which the person is asked to stand "as still as possible," commonly while looking at a fixed target. The COP displacement is then measured and analyzed to quantify the postural sway.

Although the most utilized instrument to evaluate postural control is the force plate and the most commonly measured variable is COP displacement, there is no agreement about which variables derived from the COP signal should be used to evaluate postural sway (see, for example, Kapteyn et al. 1983; Gagey and Weber 2005; Visser et al. 2008). Typically, COP displacement during a standing task can be visualized in two ways: in statokinesigram and stabilogram plots (Figure 10.2). The *statokinesigram* is the map of the COP displacement in the sagittal plane (anteroposterior direction, COP ap) versus the COP displacement in the frontal plane (mediolateral direction, COP ml); whereas the *stabilogram* is the time series of the COP displacement in each direction. Customarily, posturographic analysis has been divided into global and structural analyses. *Global analysis* is related to the quantification of the total amount of body sway, whereas *structural analysis* quantifies particular events or components of body sway.

A large number of measures have been used to describe the amount of postural sway (Winter et al. 1990; Prieto et al. 1996; Duarte and Zatsiorsky 1999; Baratto et al. 2002; Raymakers et al. 2005; van der Kooij et al. 2005; Piirtola and Era 2006; Rougier 2008). Among them, the most common measures are COP spatial displacement (usually standard deviation in each direction or total area), mean speed or velocity, and frequency variables (usually mean or median frequency). Some of the most common variables used in the quantification of body sway in the time and frequency domains are presented in Table 10.1, and an example of the power spectral density

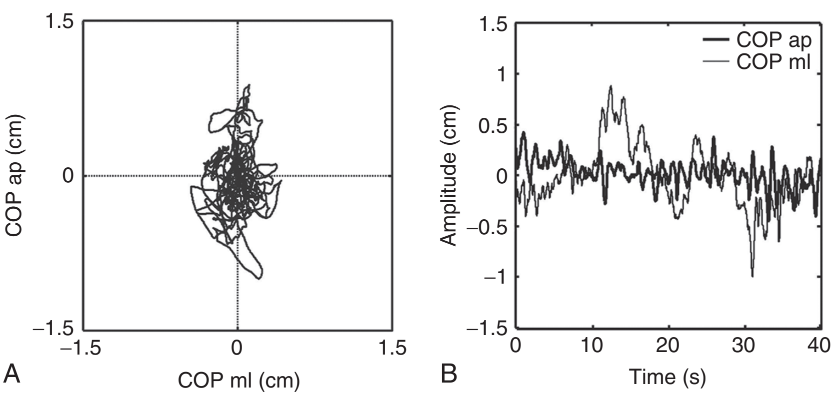

Figure 10.2 Examples of ***statokinesigram*** (A) and stabilogram (B) of center of pressure (COP) displacement during standing as still as possible on a force plate.

Table 10.1 Usual variables used in the global analysis of COP displacement and examples on how to compute these variables by using Matlab software

Variable	Matlab code
Standard deviation	std(COP)
RMS (Root Mean Square)	sqrt(sum(COP.^2)/length(COP))
Range of COP displacement	max(COP) - min(COP)
Sway path	sum(abs(COP))
Resultant sway path	sum(sqrt(COPap.^2 + COPml.^2))
Area (95% of the COP data inside)	[vec,val] = eig(cov(COPap,COPml)); Area = pi*prod(2.4478*sqrt(svd(val)))
Mean speed or velocity	sum(abs(diff(COP)))*frequency/length(COP)
Resultant mean velocity	sum(sqrt(diff(COPap).^2+diff(COPml).^2))* frequency /length(COPap)

Power spectral density
Peak (Fpeak)
Mean (Fmean)
Median (F50) frequency and the frequency band that contains up to 80% of the spectrum (F80)
	nfft = round(length(COP)/2);
	[p,f] = psd(detrend(COP),nfft,
	frequency,nfft,round(nfft/2));
	[m,peak] = max(p);
	area = cumtrapz(f,p);
	F50 = find(area >= .50*area(end));
	F80 = find(area >= .80*area(end));
	Fmean = trapz(f,f.*p)/trapz(f,p)
	Fpeak = f(peak)
	F50 = f(F50(1))
	F80 = f(F80(1))

estimation and its outcome variables for COP displacement during quiet standing is presented in Figure 10.3.

Baratto and collaborators (2002) examined 38 posturographic measures calculated from COP time series and examined the reliability and power of the measures to discriminate three different groups of individuals: normal individuals, parkinsonian patients, and osteoporotic patients. They concluded that only four measures were valuable for clinical practice: total sway path, frequency band, and two measures from COP decomposition called *sway-density plots*. The first two variables were derived from global analysis, whereas the other two measures were derived from structural analysis. The mean speed or velocity of the COP migration has been considered as the measure of greater consistency across repetitions (Lafond et al. 2004a; Cornilleau-Peres et al. 2005). On the other hand, Doyle and collaborators (2005) reported that peak velocity and area presented indexes of the lowest and highest reliability, respectively. Raymakers and collaborators (2005) observed that the velocity of the COP displacement was more

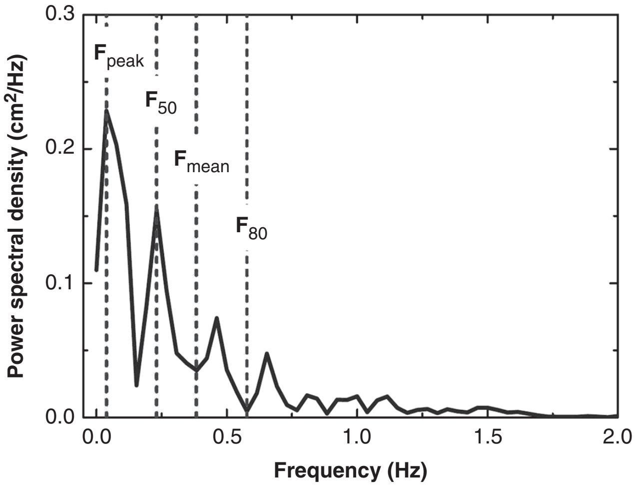

Figure 10.3 Example of the power spectral density estimation of center of pressure (COP) displacement during quiet standing. The peak (Fpeak), mean (Fmean), and median (F50) frequencies and the frequency band that contains up to 80% of the power spectrum (F80) are also shown.

reliable for comparisons between different aging groups and between groups with different health conditions. These different results can be due to the absence of standardization of the methods used to evaluate equilibrium control, such as differences in time duration (10–120 s), number of repetitions (three to nine repetitions), and frequency of data acquisition (10–100 Hz).

Center of Gravity Estimation

Typically, in posturography, instead of measuring the sway of each segment, measures of whole body sway are used; the displacements of COGv and COP are the most common measures of body sway. (However, bear in mind that the COP is not a direct measurement of postural sway of the body or its segments.) Although COP displacement can be easily measured with a force plate, the direct measurement of COGv is more complicated and typically subject to a larger error. The direct measurement of COGv is computed by recording the position of each body segment and estimating each segment mass, using an anthropometric model. More commonly, the displacement of the COGv is indirectly determined from the COP displacement, and different methods are available that produce similar results (Lafond et al. 2004b). In one of these methods, the COGv displacement is obtained by double integration of the horizontal force in combination with information from the COP displacement (King and Zatsiorsky 1997; Zatsiorsky and King 1998; Zatsiorsky and Duarte 2000). A computational algorithm implementing this method is available on the Internet (http://demotu.org/software/gline.m). A simpler method to derive the COGv displacement is to apply a low-pass filter to the COP displacement (Benda et al. 1994; Caron et al. 1997; Baratto et al. 2002). The use of a low-pass filter is motivated by modeling the mechanics of the standing-still task as an inverted pendulum, as illustrated next. Let us consider for now only the movement in the anteroposterior direction (in the sagittal plane) of a person standing still. Let us represent the human body as composed of two rigid segments articulated by a single hinge joint (feet, rest of the body, and ankle joint). Given this simplification, all the mechanical quantities important for understanding the motion of the body are represented in Figure 10.4.

Applying the second Newton-Euler equation of motion to the inverted pendulum system in this two-dimensional problem, and after a few simplifications, the following equation represents the relation between the *COGv* acceleration and the *COGv* and *COP* displacements:

$$\frac{d^2COG_v}{dt^2} \approx \frac{mgd}{I}\left(COG_v - COP\right) \qquad \text{(Eq. 1)}$$

where I is the moment of inertia of the body around the ankle.

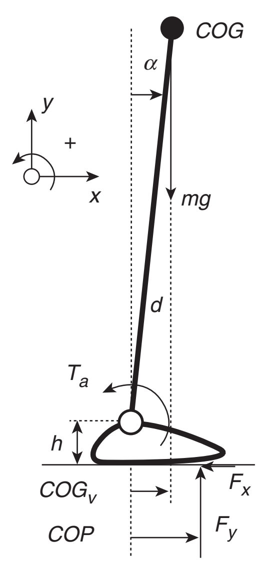

Figure 10.4 Single inverted pendulum model for the representation of a human standing. *COG*, center of gravity; COG_v, *COG* vertical projection in relation to the ankle joint; *COP*, center of pressure in relation to the ankle joint; m, g, body mass, acceleration of gravity; F_x, F_y, horizontal and vertical components of the resultant ground reaction force; T_a, torque at the ankle joint; d, distance between the *COG* and ankle joint; h, height of the ankle joint to the ground; α, angle of the body.

If we rewrite ~~(~~Equation 1 in the frequency domain by computing its Fourier transform, we obtain:

$$\frac{\overline{COGv}(\omega)}{\overline{COP}(\omega)} = \frac{\omega_0^2}{\omega^2 + \omega_0^2} \qquad \text{(Eq. 2)}$$

where ω is the angular frequency and $\omega_0 = \sqrt{mgd/I}$ represents the natural frequency of the pendulum.

The term on the right side of ~~(~~Equation 2 is always lower than 1 and indicates that COGv is indeed a filtered version of the COP in the frequency domain. For a person with 70 kg of mass and 1.70 m of height, ω_0 is equal to 3 rad/s, and the filter with this parameter will be similar to a low-pass filter with a cutoff frequency in the range of 0.4–0.5 Hz (Benda et al. 1994; Caron et al. 1997). Table 10.2 shows a Matlab code implementation of this method. A reliable estimation of the COGv based on this method depends on the assumption that the dynamics of COP and COGv can be captured by the inverted pendulum model. In addition, because of the Fourier transform, the COP data should be suitably long (at least 30 s). The mentioned above "double integration of the horizontal force" method is free from this requirement and has an additional advantage in that the ~~CoG~~ position is determined at each instant in time and not on average over the period of observation, as in those methods based on data filtering.

Table 10.2 Matlab code for estimation of COGv from the COP displacement

```
function COGv = cogve(COP,freq,m,H)
%COGVE estimates COGv from COP using a FFT filter
%SYNTAX:
% COGv = cogve(COP,freq,m,H)
% cogve(COP,freq,m,H)
%INPUTS:
% COP: column vector of the center of pressure [m]
% freq: sampling frequency [Hz]
% m: body mass of the subject [kg]
% H: height of the subject [m]
%OUTPUT:
% COGv: column vector of the center of gravity vertical projection [m]
%cogve(COP,freq,m,H) with no output plots the COP and COGv data.

%Remove mean to decrease instabilities at the extremities:
mcop = mean(COP); COP = COP - mcop;
%Parameters:
%Height of the COG w.r.t. ankle:
h = 0.56*H - 0.039*H; %(McGinnis 2005; Winter 2005)
%Body moment of inertia around the ankle:
I = m*0.0533*H^2 + m*h^2; %(Breniere 1996)
```

(*Cont'd*)

Table 10.2 Matlab code for estimation of COGv from the COP displacement (*Continued*)

```
%Gravity acceleration:
g = 9.8;
%Pendulum natural frequency:
w02 = m*g*h/I;
%Make sure COP is a column vector:
if size(COP,1)==1; COP=COP’; disp(‘COP transformed to column vector’), end
%Number of data:
ncop = length(COP);
nfft = 2^nextpow2(ncop);
%COP fft:
COPf = fft(COP,nfft)/ncop;
%Angular frequency vector:
w = 2*pi*freq/2*linspace(0,1,nfft/2+1)’;
w = [w; -w(end-1:-1:2)];
%Transfer function:
TF = w02./(w.^2 + w02);
%COGv:
COGv = real(ifft(COPf.*TF)*ncop);
COGv = COGv(1:ncop);
%Get back the mean (COP & COGv have same mean):
COP = COP + mcop; COGv = COGv + mcop;
%Plot:
t = (1:ncop)’/freq;
figure, plot(t,COP,‘b’,t,COGv,‘r’,‘LineWidth’,2)
legend(‘COP’,‘COGv’,‘Location’,‘best’)
xlabel(‘Time [s]’), ylabel(‘Amplitude [m]’)
```

This code is also available at http://demotu.org/software/cogve.m

PROLONGED UNCONSTRAINED STANDING

Under natural standing conditions, in which persons are not obliged to stand as still as possible, people usually adopt asymmetrical postures and tend to change their body position periodically while adopting relatively fixed body postures for certain periods of time. In natural standing, continuous low-amplitude and slow swaying of the body, which is normally observed during standing still, is commonly interrupted by postural changes characterized by fast and gross body movements. Hereafter, we will refer to such a task—standing for several minutes without a requirement to stay still, but with a requirement either not to change feet positions on the ground or not to step off the force plate—as prolonged unconstrained standing.

To better understand what people do during prolonged unconstrained standing, Duarte and Zatsiorsky (1999) analyzed the COP displacement of young and healthy individuals standing for 30 minutes with an upright bipedal posture on a force platform. The individuals were allowed to change their posture at any time, and there were no specific instructions on how to stand, except for the requirement to not step off the force platform. To reproduce the fact that people actually stand to do something else, the individuals were allowed to chat occasionally with another person in front of him or her. Figure 10.5 shows exemplary data for the COP displacement for one individual.

A few distinct characteristics of prolonged unconstrained standing can be noted on the COP data. First, when the COP displacement is mapped in the anteroposterior versus mediolateral plane (statokinesigram), two typical patterns can be observed: multiregion and single-region standing (Figure 10.5). In multiregion standing, the individuals tend to change the

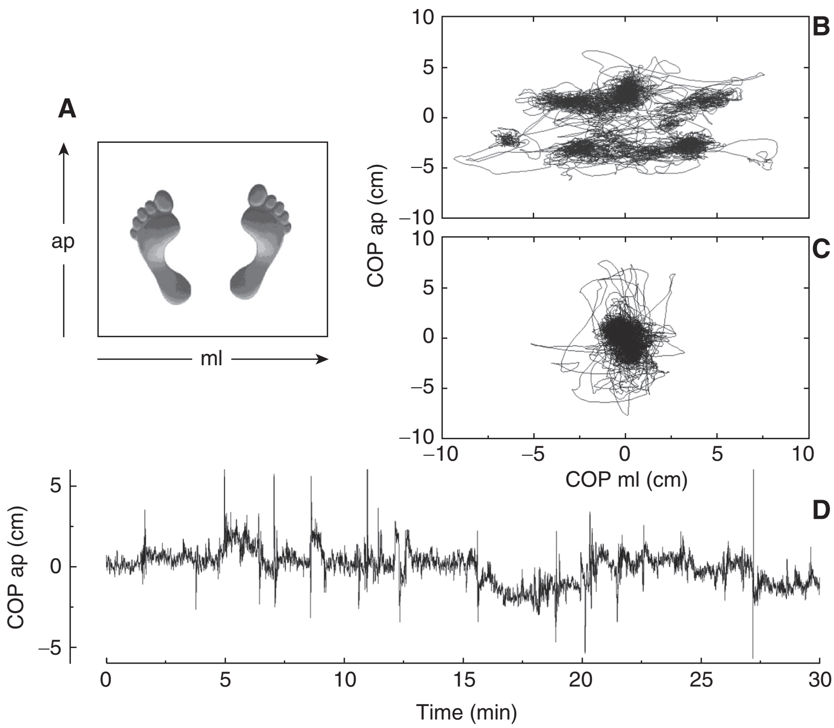

Figure 10.5 Position of the subjects on the force plate and axes convention (**A**). Two examples of statokinesigrams during prolonged unconstrained standing for 30 minutes: multiregion (**B**) and single-region standing (**C**). Exemplary center of pressure (COP) stabilogram (**D**).
Data from Duarte, M., and V.M. Zatsiorsky. 2001. Long-range correlations in human standing. *Physics Letters A* 283: 124–28, with permission of the publisher.

average location of the COP several times during the trial. Second, when the COP displacement is plotted against time (stabilogram), other two characteristics can be observed (Fig. 10.5): (a) the presence of specific (localized) events of larger amplitude, which have been classified as COP migration patterns of specific types; and (b) the presence of very low frequencies in the COP displacement, a typical signature of a long-range correlation process or long-memory process. These distinct characteristics of prolonged unconstrained standing are discussed next.

The specific (localized) events in the COP data during prolonged unconstrained standing have been classified as COP migration patterns of the following types (Duarte and Zatsiorsky 1999):

- *Shifting*: a fast (step-like) displacement of the average position of the COP from one region to another;
- *Fidgeting*: a fast and large displacement and returning of the COP to approximately the same position; and
- *Drifting*: a slow continuous displacement of the average position of the COP (linear or nonlinear trend).

Figure 10.6 shows a representative example of the three patterns in a COP time series of the present study. In general, these three patterns are always observed, in varying quantities, during prolonged unconstrained standing. In fact, these patterns can be seen as different forms of shifting. Fidgeting is a shifting followed by another in the opposite direction, and drifting is a very long shifting in time.

Duarte and Zatsiorsky (1999) parameterized the COP migration patterns in terms of a few quantities, and they were able to objectively identify such parameters with computational algorithms. For recognition of shifting, any two consecutive nonoverlapping moving windows, W_1 and W_2, satisfying Equation 1 were classified as a shift:

$$\left| \frac{\bar{x}_{W1} - \bar{x}_{W2}}{\sqrt{SD^2_{W1} + SD^2_{W2}}} \right| \geq f_{shift} \qquad \text{(Eq. 3)}$$

where $\bar{x}_{Wi}$ $(i=1,2)$ is the mean of the COP data for the windows W_1 and W_2, SD_{Wi} is the standard deviation of the COP data in the window Wi, and f_{shift} is the threshold value of the amplitude of the shift pattern (in units of $SD_{W1} + SD_{W2}$). The amplitude of the shift is defined as $\left|\bar{x}_{W1} - \bar{x}_{W2}\right|$. The estimated width of the shift (the time taken to shift the COP position) is given by the interval, W_S, separating the two consecutive windows.

For recognition of fidgeting, any peak or valley satisfying Equation 4 was classified as a fidget:

$$\left| \frac{x_F - \bar{x}_W}{SD_W} \right| \geq f_{fidget} \qquad \text{(Eq. 4)}$$

where x_F is the amplitude of the peak or valley, $\bar{x}_W$ is the mean COP data for the window W, SD_W is the standard deviation of the COP data in the window W, and f_{fidget} is the threshold value for the amplitude of the fidget pattern (in units of SD_W). The amplitude of the fidget is defined as $\left|x_F - \bar{x}_W\right|$. The width of the fidget, W_f, was estimated by the full width at half maximum of the fidget (see Fig. 10.6).

For recognition of drifting, the data between two consecutive shifts were smoothed using a low-pass filter with a variable cutoff frequency $F_C = {}^1\!/_{2W_D}$, where W_D was the preselected minimal drifting width. This procedure preserves only the low-frequency trend (drift) in the data. If the difference

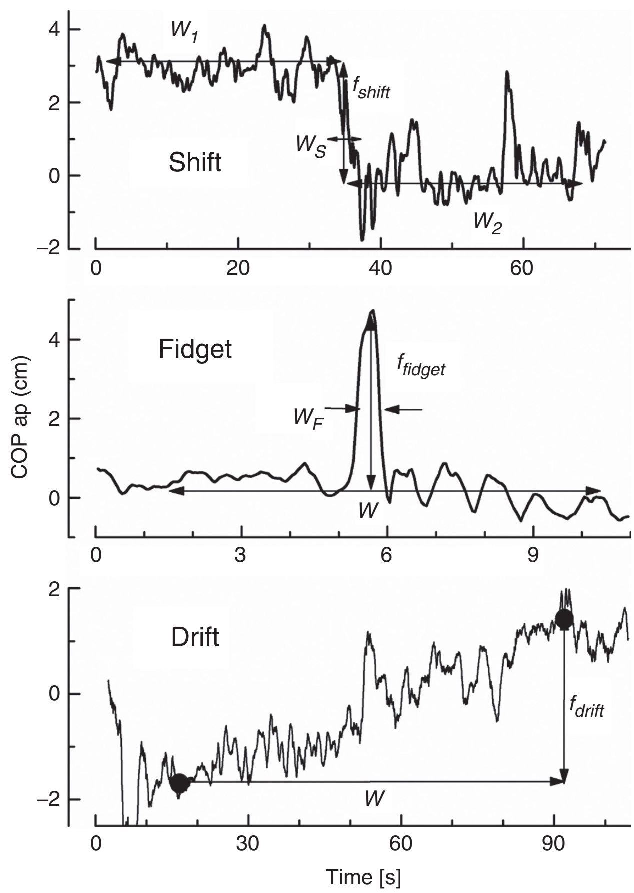

Figure 10.6 An example of shifting, fidgeting, and drifting patterns during prolonged unconstrained standing, with the corresponding parameters used for identification.
Data from Duarte, M., W. Harvey, and V.M. Zatsiorsky. 2000. Stabilographic analysis of unconstrained standing. *Ergonomics* 43: 1824–39, with permission of the publisher.

between the amplitudes of two consecutive local maximum and minimum values satisfied Equation 5, the COP displacements between the consecutive maximum and minimum were classified as a drift.

$$\left|\frac{x_{\max} - x_{\min}}{SD_W}\right| \geq f_{drift} \qquad \text{(Eq. 5)}$$

where x_{max} and x_{min} are the consecutive local maximum and minimum amplitudes, SD_W is the standard deviation of the COP data in the window containing the data between the maximum and minimum values, and f_{drift} is the threshold value of the drift amplitude (in units of SD_W). The amplitude of the drift is defined as $|x_{\max} - x_{\min}|$.

Typical criterion values chosen for classifying the data as shift, fidget, or drift patterns are respectively: a minimum shift amplitude of 2 SD, a maximum shift width of 5 s, and a base window of 15 s; a minimum fidget amplitude of 3 SD, a maximum fidget width of 4 s, and a base window of 60 s; and a minimum drift amplitude of 1 SD, with a minimum drift width of 60s.

By performing the above computations, Duarte and Zatsiorsky (1999) observed that, during prolonged unconstrained standing by young healthy adults, the most common COP pattern was fidgeting, followed by shifting, and then drifting. On average, one postural change was produced every 20 s, either at the anteroposterior or at the mediolateral direction.

Why exactly postural changes are produced during prolonged unconstrained standing and how our posture is regulated by the presence of these postural changes are questions discussed next.

Basis for Postural Changes During Prolonged Unconstrained Standing

A qualitative observation of individuals during prolonged unconstrained standing reveals that the three COP patterns result from various movements of the body segments and/or the body as a whole. The most commonly observed body segments motions are arms, head, and trunk movement, as well as a redistribution of the body weight from one leg to another.

The existence of shifting and drifting fits well to the hypothesis of Lestienne and Gurfinkel (1988). These authors suggested that the motor control system responsible for balance maintenance is a hierarchical two-level system. The upper level ("conservative") determines a reference frame for an equilibrium, with respect to which the equilibrium is maintained. The low level ("operative") maintains the equilibrium around the predetermined reference position. This hypothesis was supported in studies by Gurfinkel and collaborators (1995), where the supporting surface was rotated slowly. They found that individuals maintained for some time a fixed body orientation with respect to the surface (the perceived vertical) rather than with respect to the real vertical. Lestienne and Gurfinkel (1988), as well as Gurfinkel and collaborators (1995), did not address in their studies a possible reference point migration during natural standing. The observations of shifting and drifting during prolonged unconstrained standing indeed suggest that such migration takes place.

Postural changes are most commonly viewed as a mechanism to avoid or minimize physiological fatigue and discomfort in the musculoskeletal system by decreasing venous pooling in the lower extremities, decreasing occlusion of blood flow through some regions of the sole of the foot caused by the continuous pressure in static standing, or alleviating the pressure on joints by "repumping" the cartilage fluid (Brantingham et al. 1970; Cavanagh et al. 1987; Zhang et al. 1991; Kim et al. 1994). However, at least another reason for postural changes during natural standing would be a mechanism to interact (e.g., with another person), explore, and gather information from the environment, mainly by using the visual sensory system (Riccio and Stoffregen 1988).

Duarte and collaborators (2000) tested some of these hypotheses for postural changes during prolonged unconstrained standing, employing again the analysis of COP migration patterns. They manipulated the load on the individual, under the rationale that, with the addition of an external load, the pressure on joint cartilage in the lower extremities and on the plantar sole increases. If the main reason for postural changes is to "repump" the cartilage fluid, the number of postural changes should increase when holding a load. If the reasons for postural changes are solely to decrease venous blood pooling in the lower extremities and to allow momentary blood flow through some regions of the foot sole, the number of postural changes would not vary during loaded standing. This last result would occur because the pressure on the plantar sole during normal or unloaded standing is already large enough to occlude the circulation of blood in this region (Cavanagh et al. 1987). Duarte and collaborators (2000) did not observe any increase in the number of postural changes, and so they discarded the idea of postural changes to "repump" the cartilage fluid. They also requested that participants stand with eyes closed in order to remove the visual system that is used to interact with the environment. They hypothesized that if postural changes during prolonged unconstrained standing were performed to explore the environment through the visual sensory system, then the absence of vision would lead to fewer postural changes. They also did not observe any decrease in the number of postural changes during eyes-closed standing, and so they discarded this hypothesis.

Prolonged Unconstrained Standing as a Fractal Process

The presence of fast and large fluctuations, as well as of slow and small fluctuations, in the COP displacement during prolonged unconstrained standing is a typical characteristic of a fractal process. A fractal is "a rough or fragmented geometric shape that can be split into parts, each of which is (at least approximately) a reduced-size copy of the whole" (Mandelbrot 1983). The observation of "reduced-size copy of the whole" in space is termed the *self-similarity* property; this observation in time is termed the *self-affinity* property (which can also be referred as a *long-range correlation* or *long-memory* process because the dependence of data farther apart is higher than it is expected for independent data). These two properties can be seen as scaling laws of the spatial and temporal variability of a fractal process. Duarte and Zatsiorsky (2000, 2001) analyzed the COP displacement during prolonged unconstrained standing and indeed observed that natural standing is a fractal process exhibiting these two properties, as illustrated in Figure 10.7.

The presence of long-range correlations in the COP data during natural standing is, in fact, not surprising. The branching in trees and in our lungs, the fluctuations of the waterfall sound and of our heartbeats are a few examples of the ubiquity of fractals in nature. Nevertheless, this "trivial" characteristic has important implications for the study of postural control in humans. Here are a few of them: One important issue in postural studies is the period of data acquisition (i.e., for how long should one collect data to capture essential properties of human standing?). In a frequently referenced study, Powell and Dzendolet (1984) reported low frequencies in the COP data of 130 seconds of duration. Different authors have cited this paper as a reference to justify the acquisition of data for no more than 2 minutes. Our studies suggest that, with longer acquisition time, even lower frequencies of COP could be observed. The important conclusion is that the choice of period of acquisition has to be based on which periods (frequencies) are regarded as relevant for the study in question. Another issue is whether the stabilogram can be considered a stationary process. The distinction between nonstationarity and long-range correlations in time series analysis is an ill-posed problem, and studies on stationarity in COP data have indeed shown discrepant results. Given our findings, such differences are the consequence of different periods of observation, and the investigators have tested only small portions of a longer process. Because of the presence of long-range correlations, apparent nonstationarities in short COP time series might actually represent fluctuations of a longer stationary process. Thus, the issue of stationarity cannot be adequately addressed using short time series of up to few minutes. Finally, the property of self-affinity in the COP data implies that, to properly compare COP data of different lengths (different periods of acquisition), the COP data should be scaled by the fractal exponent.

Prolonged Unconstrained Standing and Aging

It is common—but not ubiquitous—to find an increase in postural sway with aging when an individual is asked to stay as still as possible for a short

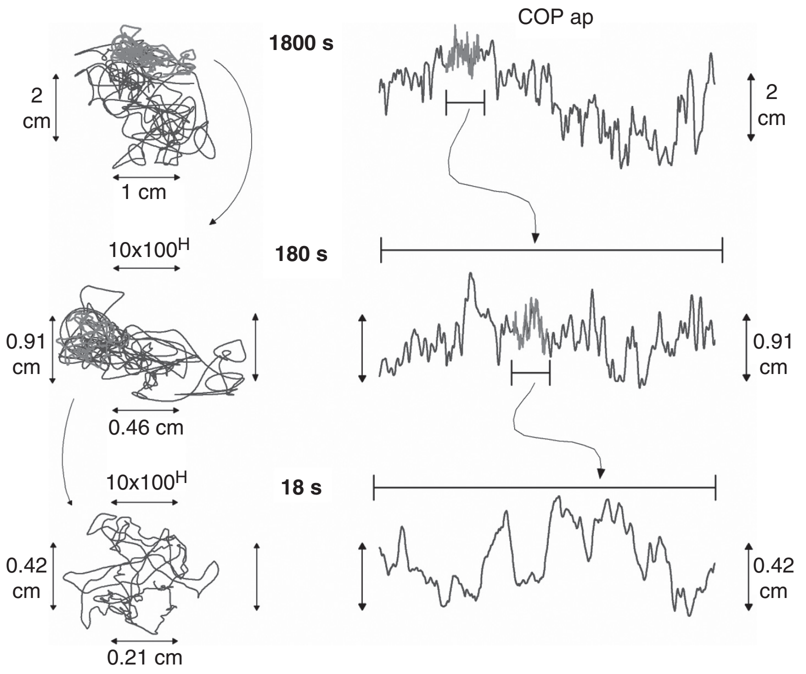

Figure 10.7 A: Statokinesigrams (*left*) and stabilograms in the anterior-posterior (ap) direction of the entire data set during natural standing (1,800 s, first row), for 1/10 of the data set (180 s, second row), and for 1/100 of the data set (18 s, third row). The *Hurst* exponent (*H*) for this example is 0.34, giving a reduction of 2.2 in the amplitude scale for each 10-fold of reduction of the time scale. Both scaled and real axes are indicated in the 180 s and the 18 s plots for illustration. Notice that after each scaling (that is related to the fractal exponent and to the period of time), the three statokinesigrams and stabilograms present roughly the same amplitudes in space. For the sake of clarity, not all points are shown for the 1,800 s and 180 s plots. The difference in the fine structure, observed for the 18 s time series compared to the other two time series, is due to the fact that center of pressure (COP) displacements for short intervals up to 1 s display a different behavior.

Data from Duarte, M., and V.M. Zatsiorsky. 2000. On the fractal properties of natural human standing. *Neuroscience Letters* 283: 173–76; Duarte, M., and V.M. Zatsiorsky. 2001. Long-range correlations in human standing. *Physics Letters A* 283: 124–28, with permission of the publishers.

period of time. However, Freitas and collaborators (2005b) found that elderly individuals tend to show an opposite behavior compared to young adults during prolonged unconstrained standing. This was because elderly individuals produced fewer large-amplitude postural changes compared to young adults. Specifically, the elderly individuals produced smaller shift patterns that resulted in fewer COP multiregion patterns. Figure 10.8 shows typical examples of COP displacement in the anteroposterior direction versus the mediolateral direction during standing still and during

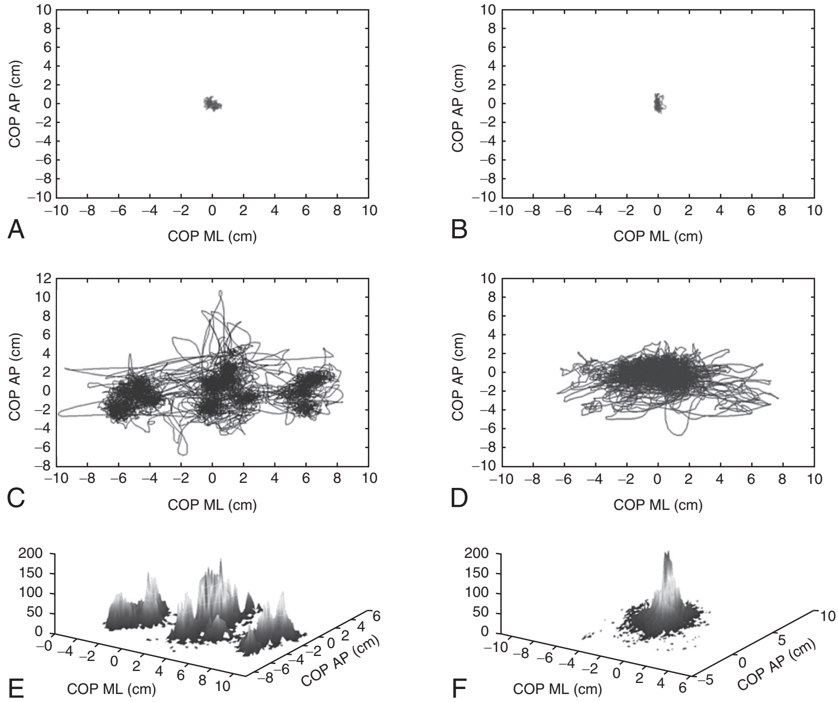

Figure 10.8 Examples of the center of pressure (COP) in the anterior-posterior (AP) direction versus the medio-lateral direction (ML) for one young adult (**A**, **C**) and one elderly individual (**B**, **D**) during standing still (**A**, **B**) and prolonged unconstrained standing (**C**, **D**). Graphs **E** and **F** show the respective COP histograms of the prolonged unconstrained standing trials.

prolonged unconstrained standing for one young adult and one elderly individual (data from Freitas et al. 2005b). The difference in the behavior of each representative participant is clear. On average, young adults (n = 14) exhibited six COP multiregion patterns, whereas the elderly individuals (n = 14) showed only one COP multiregion pattern during the 30 minutes of unconstrained standing (Freitas et al. 2005b). These results indicate that elderly individuals adopt a "freezing" strategy during prolonged standing. Because a decrease in mobility is typically observed in elderly individuals (Gunter et al. 2000; Hatch et al. 2003), Freitas and collaborators (2005b) hypothesized that the lack of mobility in elderly individuals might be responsible for the decreased numbers of postural changes of large amplitude.

More detailed examinations of the freezing strategy and the possible effect of decreased mobility on the postural changes during prolonged unconstrained standing in elderly individuals were undertaken by Prado and Duarte (2009). The hypothesis for this study was that elderly individuals produced fewer postural changes due to reduced mobility and, specifically, due to the inability to use the load/unload mechanism to transfer weight from one leg to the other. The specificity of the hypothesis is because shifts are typically a reflection of unloading and loading the body weight from one leg to the other in the mediolateral direction. As Freitas and collaborators (2005b) observed a decrease in the number of large shifts in elderly adults, this could be an indication of the inability to use such a load/unload mechanism. As early as 1913, Mosher stressed the importance of shifting the body weight from side to side for comfort (Zacharkow 1988). So, to investigate this hypothesis, Prado and Duarte (2009) used the dual force plate paradigm (one force plate under each leg) and measured the mobility of individuals. They examined 20 elderly individuals (70 ± 4 years) and 20 young adults (25 ± 4 years) without any known postural or skeletal disorders. Subjects performed two tasks: quiet standing for 60 seconds and prolonged standing for 30 minutes, standing with each leg on a force plate. In the prolonged unconstrained standing, the participants were allowed to change their posture freely at any time, and there were no specific instructions on how to stand except for the requirement not to step off the force plates. In the prolonged unconstrained standing, participants watched a television documentary displayed 3 m in front of them. The mobility of each individual was measured with the timed up-and-go test (Podsiadlo and Richardson 1991).

A change larger than 25% of the body weight in the vertical ground reaction force (Fz) on any of the force plates was counted as one transfer of weight from one leg to the other (termed shift). Different criteria for the detection of weight transfer, ranging from 0.1 to 0.5, were tested, and the comparison across groups was not affected by the criterion value. Figure 10.8 illustrates exemplary time series of the vertical ground reaction forces from the left and right legs during prolonged unconstrained standing by one young adult and one elderly adult. Again, the "freezing" behavior that elderly adults seem to adopt is evident by looking at these plots. One can observe less weight transfer from one leg to the other for the elderly adult than for the young adult, and in this particular case, the elderly adult had more weight on the left leg to begin with, with the weight share slowly increasing over time. The elderly adults significantly ($p < 0.001$) produced less weight transfers from one leg to the other and with lower amplitudes than did the young adults during prolonged unconstrained standing (Figure 10.9). For the quiet-standing task, neither group produced any weight transfer. The elderly adults may have produced less weight transfers just because they adopted a narrow base of support with their feet, and this makes it difficult to transfer weight. To investigate this possibility, Prado and Duarte (2009), using a motion capture system, also looked at the kinematics of the feet during prolonged unconstrained standing by recording the position of reflective markers placed on the participant's feet. However, they found that the base of support width of the

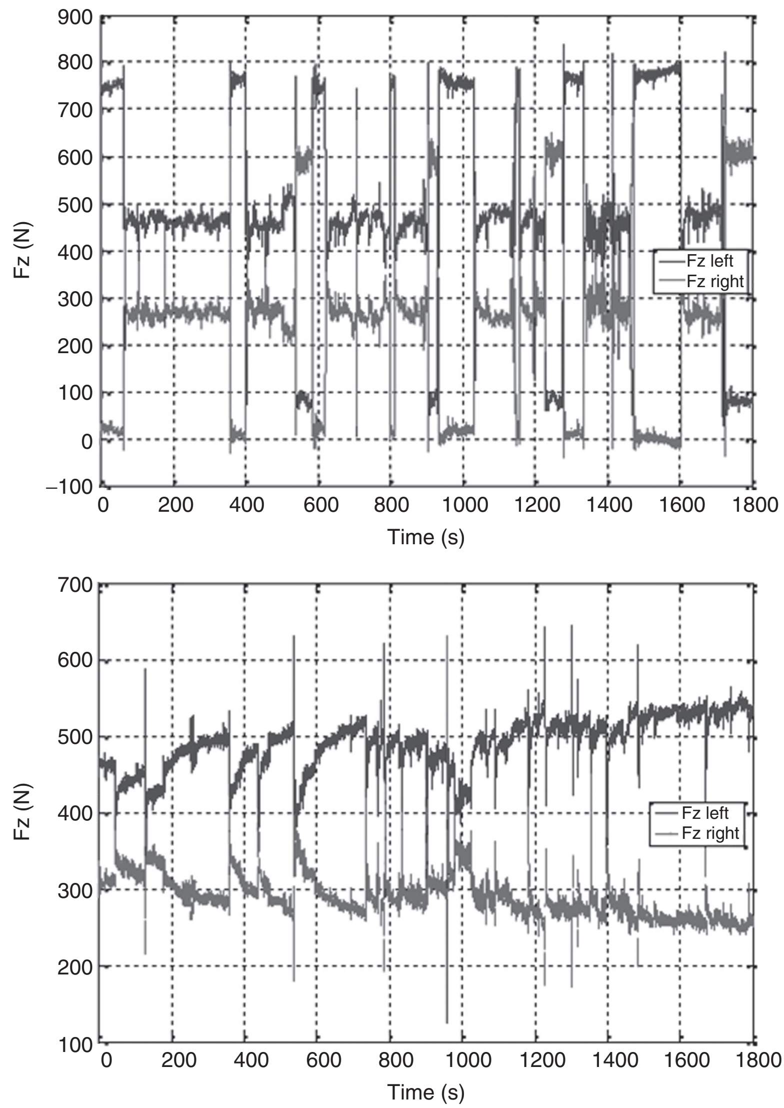

Figure 10.9 Exemplary time series of the vertical ground reaction forces from the left (Fz left, *black line*) and right (Fz right, *gray line*) legs during prolonged unconstrained standing by one young adult (*top*) and one elderly adult (*bottom*).

elderly adults was no different from that of the young adults during prolonged unconstrained standing (Figure 10.10).

Despite the decreased capacity of elderly adults to produce weight transfer, which could be viewed as a sign of decreased mobility in this group, in fact, the specific measure of mobility, the timed up-and-go

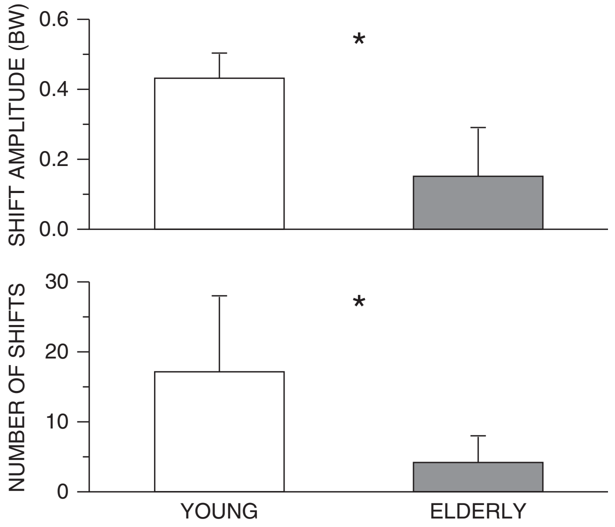

Figure 10.10 Mean and standard deviation of amplitude (*top*) and number (*bottom*) of shifts (weight transfers from one leg to the other) for young adults and elderly adults during prolonged unconstrained standing. * $p < 0.001$.

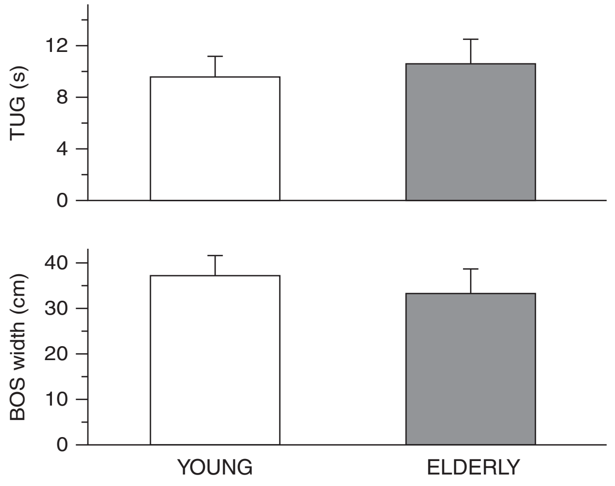

Figure 10.11 Mean and standard deviation of the base of support width for young adults and elderly adults during prolonged unconstrained standing (*left*). Mean and standard deviation of the timed up and go (TUG) test for young adults and elderly adults (*right*).

test, was not different between the young adults and the elderly adults (Figure 10.11). These results are intriguing; if the elderly adults were simply more cautious or afraid of falling, and so did not move during unconstrained standing, we should have observed a similar cautious behavior during the timed up-and-go test. But this was not the case. Therefore, we can only speculate why we observed such a difference between young and elderly individuals in producing weight transfers, but no difference in the mobility measures. If we assume that postural changes are a likely response to reduce musculoskeletal discomfort, then they are somehow initiated by proprioceptive information signaling such discomfort. It is possible that the observed decreased capacity of weight transfer by elderly adults is due to diminished somatosensory information, which is not triggering such postural changes.

Prolonged Unconstrained Standing and Musculoskeletal Problems

The observed decrease in the number and amplitude of postural changes during unconstrained standing in elderly adults may not be exclusively due to the aging factor. Rather, an immediate cause of the observed age trend could be musculoskeletal disorders associated with aging. Lafond and collaborators (2009) used the quantification of COP patterns paradigm to investigate how individuals with chronic low-back pain behave during prolonged unconstrained standing. Prolonged standing is linked with the onset of low-back pain symptoms in working populations (Macfarlane et al. 1997; Xu et al. 1997). Lafond and collaborators (2009) hypothesized that the onset of pain during prolonged standing might be due to the inability to produce postural changes during such tasks. Indeed, they found that individuals with chronic low-back pain presented fewer postural changes in the anteroposterior direction, with decreased postural sway amplitude during the prolonged standing task in comparison to the healthy group. Lafond and collaborators (2009) suggested that this deficit may contribute to low-back pain persistence or an increased risk of recurrent back pain episodes. They also linked this deficit to reduced proprioceptive information from the low back or altered sensorimotor integration in individuals with chronic low-back pain.

CONCLUDING REMARKS

Although we often refer to the postural control system as the entity responsible for the control of equilibrium during a certain body posture, this system is neither a single neuroanatomic structure in our body nor a single task that we perform. Postural control is dependent on a rich and fine integration of many sensorimotor processes in our body, the goal we are trying to accomplish, and the surrounding environment. Natural (unconstrained) standing is a good example of this complexity. The series of studies we have conducted might be useful not only for understanding how humans behave during such a task but in general how the postural control system works. The fact that during natural (unconstrained) standing humans control their posture around different reference positions with no difficulty might suggest that the postural control system in fact never specifies a fixed and unique reference position, even when standing still. When standing still, we would consciously try to avoid any postural change; however, what our postural control system adopts as an exact reference position, if it adopts one, probably is beyond our voluntary will.

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REFERENCES

Baratto, L., P.G. Morasso, C. Re, and G. Spada. 2002. A new look at posturographic analysis in the clinical context: Sway-density versus other parameterization techniques. *Motor Control* 6: 246–70.

Benda, B.J., P.O. Riley, and D.E. Krebs. 1994. Biomechanical relationship between center of gravity and center of pressure during standing. *Rehabilitation Engineering, IEEE Transactions* [see also *IEEE Transactions on Neural Systems and Rehabilitation*] 2: 3–10.

Bonnet, C., C. Carello, and M.T. Turvey. 2009. Diabetes and postural stability: Review and hypotheses. *Journal of Motor Behavior* 41: 172–90.

Brantingham, C.R., B.E. Beekman, C.N. Moss, and R.B. Gordon. 1970. Enhanced venous pump activity as a result of standing on a varied terrain floor surface. *Journal of Occupational Medicine* 12: 164–69.

Breniere, Y. 1996. Why we walk the way we do. *Journal of Motor Behavior* 28: 291–98.

Caron, O., B. Faure, and Y. Breniere. 1997. Estimating the centre of gravity of the body on the basis of the centre of pressure in standing posture. *Journal of Biomechanics* 30: 1169–71.

Cavanagh, P.R., M.M. Rodgers, and A. Iiboshi. 1987. Pressure distribution under symptom-free feet during barefoot standing. *Foot & Ankle* 7: 262–76.

Cornilleau-Peres, V., N. Shabana, J. Droulez, J.C. Goh, G.S. Lee, and P.T. Chew. 2005. Measurement of the visual contribution to postural steadiness from the COP movement: methodology and reliability. *Gait & Posture* 22: 96–106.

Doyle, T.L., R.U. Newton, and A.F. Burnett. 2005. Reliability of traditional and fractal dimension measures of quiet stance center of pressure in young, healthy people. *Archives of Physical Medicine and Rehabilitation* 86: 2034–40.

Duarte, M., W. Harvey, and V.M. Zatsiorsky. 2000. Stabilographic analysis of unconstrained standing. *Ergonomics* 43: 1824–39.

Duarte, M., and V.M. Zatsiorsky. 1999. Patterns of center of pressure migration during prolonged unconstrained standing. *Motor Control* 3: 12–27.

Duarte, M., and V.M. Zatsiorsky. 2000. On the fractal properties of natural human standing. *Neuroscience Letters* 283: 173–76.

Duarte, M., and V.M. Zatsiorsky. 2001. Long-range correlations in human standing. *Physics Letters A* 283: 124–28.

Duarte, M., and V.M. Zatsiorsky. 2002. Effects of body lean and visual information on the equilibrium maintenance during stance. *Experimental Brain Research* 146: 60–69.

Freitas, S.M., J.M. Prado, and M. Duarte. 2005a. The use of a safety harness does not affect body sway during quiet standing.*Clinical Biomechanics (Bristol, Avon)* 20: 336–39.

Freitas, S.M., S.A. Wieczorek, P.H. Marchetti, and M. Duarte. 2005b. Age-related changes in human postural control of prolonged standing. *Gait & Posture* 22: 322–30.

Gagey, P.-M., and B. Weber. 2005. *Posturologie: Régulation et dérèglements de la station debout*. Paris: Masson.

Gunter, K.B., K.N. White, W.C. Hayes, and C.M. Snow. 2000. Functional mobility discriminates nonfallers from one-time and frequent fallers. *Journals of Gerontology. Series A, Biological Sciences and Medical Sciences* 55: M672–76.

Gurfinkel. V.S., P. Ivanenko Yu, S. Levik Yu, and I.A. Babakova. 1995. Kinesthetic reference for human orthograde posture. *Neuroscience* 68: 229–43.

Hatch, J., K.M. Gill-Body, and L.G. Portney. 2003. Determinants of balance confidence in community-dwelling elderly people. *Physical Therapy* 83: 1072–79.

Horak, F.B., C.L. Shupert, and A. Mirka A. 1989. Components of postural dyscontrol in the elderly: a review. *Neurobiology of Aging* 10: 727–38.

Kapteyn, T.S., W. Bles, C.J. Njiokiktjien, L. Kodde, C.H. Massen, and J.M. Mol. 1983. Standardization in platform stabilometry being a part of posturography. *Agressologie* 24: 321–26.

Kim, J.Y., C. Stuart-Buttle, and W.S. Marras. 1994. The effects of mats on back and leg fatigue. *Applied Ergonomics* 25: 29–34.

King, D.L., and V.M. Zatsiorsky. 1997. Extracting gravity line displacement from stabilographic recordings. *Gait & Posture* 6: 27–38.

Lafond, D., A. Champagne, M. Descarreaux, J.D. Dubois, J.M. Prado, and M. Duarte. 2009. Postural control during prolonged standing in persons with chronic low back pain. *Gait & Posture* 29: 421–27.

Lafond, D., H. Corriveau, R. Hebert, and F. Prince. 2004a. Intrasession reliability of center of pressure measures of postural steadiness in healthy elderly people. *Archives of Physical Medicine and Rehabilitation* 85: 896–901.

Lafond, D., M. Duarte, and F. Prince. 2004b. Comparison of three methods to estimate the center of mass during balance assessment. *Journal of Biomechanics* 37: 1421–26.

Lestienne, F.G., and V.S. Gurfinkel. 1988. Posture as an organizational structure based on a dual process: A formal basis to interpret changes of posture in weightlessness. In *Vestibulospinal control of posture and locomotion*, eds., O. Pompeiano and J.H.J. Allum, vol. 76, 307–313. Amsterdam New York: Elsevier.

Macfarlane, G.J., E. Thomas, A.C. Papageorgiou, P.R. Croft, M.I.V. Jayson, and A.J. Silman. 1997. Employment and physical work activities as predictors of future low back pain. *Spine* 22: 1143–49.

Mandelbrot, B.B. 1983. *The fractal geometry of nature*. New York: W.H. Freeman.

McGinnis, P.M. 2005. *Biomechanics of sport and exercise*. Champaign, IL: Human Kinetics.

Piirtola, M., and P. Era. 2006. Force platform measurements as predictors of falls among older people: A review. *Gerontology* 52: 1–16.

Podsiadlo, D., and S. Richardson. 1991. The timed "Up & Go": A test of basic functional mobility for frail elderly persons. *Journal of the American Geriatric Society* 39: 142–48.

Powell, G.M., and E. Dzendolet. 1984. Power spectral density analysis of lateral human standing sway. *Journal of Motor Behavior* 16: 424–41.

Prado, J.M., and M. Duarte. 2009. Age-related deficit in the load/unload mechanism during prolonged standing. In *Progress in motor control VII*. Marseille, France.

Prieto, T.E., J.B. Myklebust, R.G. Hoffmann, E.G. Lovett, and B.M. Myklebust. 1996. Measures of postural steadiness: differences between healthy young and elderly adults. *IEEE Transactions in Biomedical Engineering* 43: 956–66.

Raymakers, J.A., M.M. Samson, and H.J. Verhaar. 2005. The assessment of body sway and the choice of the stability parameters. *Gait & Posture* 21: 48–58.

Riccio, G.E., E.J. Martin, and T.A. Stoffregen. 1992. The role of balance dynamics in the active perception of orientation. *Journal of Experimental Psychology. Human Perception and Performance* 18: 624–44.

Riccio, G.E., and T.A. Stoffregen. 1988. Affordances as constraints on the control of stance. *Human Movement Science* 7: 265–300.

Riley, M.A., S. Mitra, T.A. Stoffregen, and M.T. Turvey. 1997. Influences of body lean and vision on unperturbed postural sway. *Motor Control* 1: 229–46.

Riley, M.A., and M.T. Turvey. 2002. Variability of determinism in motor behavior. *Journal of Motor Behavior* 34: 99–125.

Romero, D.H., and G.E. Stelmach. 2003. Changes in postural control with aging and Parkinson's disease. *IEEE Eng Med Biol Mag* 22: 27–31.

Rougier, P.R. 2008. What insights can be gained when analysing the resultant centre of pressure trajectory? *Neurophysiology Clinics* 38: 363–73.

Slobounov, S.M., E.S. Slobounova, and K.M. Newell. 1997. Virtual time-to-collision and human postural control. *Journal of Motor Behavior* 29: 263–81.

Stoffregen, T.A., C.-M. Yang, and B.G. Bardy. 2005. Affordance judgments and non-locomotor body movement. *Ecological Psychology* 17: 75–104.

van der Kooij, H., E. van Asseldonk, and F.C. van der Helm. 2005. Comparison of different methods to identify and quantify balance control. *Journal of Neuroscience Methods* 145: 175–203.

Visser, J.E., M.G. Carpenter, H. van der Kooij, and B.R. Bloem. 2008. The clinical utility of posturography. *Clinical Neurophysiology* 119: 2424–36.

Winter, D.A. 2005. *Biomechanics and motor control of human movement*. Hoboken, NJ: John Wiley & Sons.

Winter, D.A., A.E. Patla, and J.S. Frank. 1990. Assessment of balance control in humans. *Medical Progress through Technology* 16: 31–51.

Xu, Y., E. Bach, and E. Orhede. 1997. Work environment and low back pain: the influence of occupational activities. *Occupational and Environmental Medicine* 54: 741.

Zacharkow, D. 1988. *Posture: Sitting, standing, chair design, and exercise*. Springfield, IL: Thomas.

Zatsiorsky, V.M., and M. Duarte. 2000. Rambling and trembling in quiet standing. *Motor Control* 4: 185–200.

Zatsiorsky, V.M., and D.L. King. 1998. An algorithm for determining gravity line location from posturographic recordings. *Journal of Biomechanics* 31: 161–64.

Zhang, L., C.G. Drury, and S.M. Wooley. 1991. Constrained standing: evaluating the foot/floor interface. *Ergonomics* 34: 175–92.