Assessing quiet and perturbed balance: a review

HSIAO-WIECKSLER Elizabeth T.

1. Abstract

Human balance and postural control during upright stance can be assessed using a variety of methods. Various analysis techniques have been used to explore balance during both quiet and perturbed stance conditions. Quiet stance assessment typically involves the examination of postural sway, as measured by fluctuations of the center of pressure under the feet. Perturbed stance or dynamic balance can be examined through a variety of methods, ranging from mild perturbations that require a sway response to large perturbations that require one or more steps to recover balance. In this paper, we review experimental design and data analysis issues related to quiet and perturbed stance balance assessment with an emphasis on techniques utilize by our research group.

2. Introduction

Assessment of balance during upright stance has been done for a variety of reasons to assess, for example, differences due to age, fall risk, pathology, intervention treatments, etc. (e.g., (Maki et al. 1990; Thelen et al. 1997; Schiiffman et al. 2006; Yang et al. 2007). To do these assessments, examination techniques have varied from experimental studies of quiet stance to situations when an external perturbation causes a disturbance to balance. Various definitions have been used to describe the postural control system used to moderate balance in these situations. In this paper, we will refer to static balance and postural control as that used to maintain an upright position (usually stance) when no external disturbance is applied to the body (e.g., quiet stance). We will use the term dynamic balance and postural control as that used to maintain or restore an upright position during volitional movement or after an externally applied perturbation (Prieto et al. 1996). This review focuses on assessing static and dynamic balance during upright stance and not during locomotion.

This paper includes a review of postural sway measures used to characterize quiet stance balance, guidelines for design of quiet stance balance experiments with regard to selection of length and number of test trials, and review of different methods to assess dynamic balance due to both large and small externally applied perturbations. Some examples from our work that used these techniques are also presented. Finally, we discuss a promising method for predicting the dynamic postural control response to a mild perturbation using only quiet stance postural sway data through the application of a statistical mechanics theorem. This final work provides an approach that overlaps quiet and perturbed stance assessment.

3. Quiet stance assessment of balance

Frequently, quiet stance static balance is assessed by examining the postural sway of the body. Typically, this is done by examining the fluctuations of the center of pressure under the feet (COP), which is computed from ground reaction force and moment measurements.
corded by a force platform (Winter 2005). To do these assessments, an individual stands on one or two force platforms, i.e., both feet on one platform or one foot per platform. The COP is considered to be the point of application of the ground reaction force vector on the force platform. It provides a convenient summary measure that represents the movements of all body segments while attempting to remain upright. The planar characterization of the COP is called a stabilogram (Fig. 1). A number of methods for examining stabilograms have been proposed. These consist of traditional descriptive summary measures that characterize the time- and frequency-domain behaviors of the COP fluctuations. Stochastic measures based on a method called Stabilogram Diffusion Analysis have also been used to interpret the behavior of the postural control system. These measures are usually examined along one- or two-dimensions. One-dimensional metrics characterize behaviors in only the anterior-posterior or medial-lateral directions, while two-dimensional metrics characterize transverse plane or the composite radial direction behaviors.

Instructions during the assessment can influence balance measurements. We use the instruction to “stand quietly”, which means to not move voluntarily. We have noted anecdotally that researchers should not tell the subject to “stand still” as some individuals interpret this instruction as meaning to stand rigidly, which in some cases increases postural sway. Frequently, balance assessments are conducted with the eyes open. For such cases, it has been recommended that subjects should be instructed to focus on a stationary visual target to reduce extraneous postural sway due to volitional head or eye movements (Lee and Lishman 1975). Usually quiet-stance balance assessments are conducted over multiple trials. Stance width has been shown to influence balance measures, such that postural sway increases when stance width decreases due to reduced base of support (Kirby et al. 1987; Day et al. 1993). Therefore, the same foot position is recommended for all trials within a testing session (Mcllroy and Maki 1997). Some researchers have also suggested using anthropometric and foot position measurements to normalize stabilogram measures (Chiari et al. 2002).

3. 1 COP measures during quiet stance

3. 1. 1 Traditional time and frequency domain summary measures

Traditionally, balance performance has been assessed using summary measures derived from time- and frequency-domain analyses of the COP fluctuation. To determine which measures may be more sensitive to changes in balance and postural control due to age and vision, Prieto et al. (1996) compiled an exhaustive list of 36 measures (i.e., 11 basic parameters evaluated for three directions: anterior-posterior, medial-lateral, and radial, and three area parameters). These measures included time-domain measures of “distance”: mean sway velocity and mean distance, root mean square (RMS) distance, and range about the mean; time-domain measures of “area”: 95% confidence circular area; 95% confidence ellipse; time-domain “hybrid” measures: sway area, mean (or rotational) frequency, fractal dimension; frequency-domain measures: total power, median (50%) power frequency, 95% power frequency, centroidal frequency, and frequency dispersion.

More recently, Rocchi et al. (2004) used principal component analysis to develop guidelines for establishing a set of standard summary COP balance measures. Starting with the list of parameters examined by Prieto et al. (1996) and an additional term, angular deviation of the principal sway direction from the AP axis (Oliveira et al.
Assessing quiet and perturbed balance

Fig. 2 Schematic of a stabilogram-diffusion plot that is created when the mean square COP distance for a given direction $j$ ($\langle \Delta j^2 \rangle$) is plotted as a function of the time interval between assessment points ($\Delta t$), where $\Delta j$ represents the distance in the AP, ML, or RAD direction. (Adapted from (Collins and De Luca 1993))

1996), the authors identified 10 key parameters. These included the RMS distance in the anterior-posterior (AP), medial-lateral (ML), and combined radial (RAD) directions; AP and ML mean sway velocity; AP and RAD 95% power frequency; AP and RAD frequency dispersion; and angular deviation.

3.1.2 Stabilogram Diffusion Analysis measures

Stabilogram Diffusion Analysis (SDA) is based on the assumption that the postural control system operates as a coexistence of deterministic and stochastic processes, such that statistical mechanics tools can be used to analyze the fluctuations of the COP. SDA assumes that the COP during quiet stance can be modeled as a system of coupled, correlated random walks (Collins and De Luca 1993). These random walks are characterized by examining certain metrics, such as diffusion coefficients, scaling exponents, and critical transition point parameters.

The metrics derived from the SDA quantify the stochastic behavior of the COP profile and have been interpreted as providing information on the underlying control processes at work during quiet standing. Alternative methods of investigating the dynamic nature of the COP profile exist (e.g., Newell et al. 1997; Chiari et al. 2000); however, we will focus on the SDA approach proposed by Collins and De Luca (Collins and De Luca 1993).

SDA parameters are determined by examining COP time series data in the three directions (AP, ML, and RAD). If the squared distance between two points on the COP profile for a given direction is plotted against the time interval that they are separated, a stabilogram-diffusion plot is created (Fig. 2). The slope of this plot can be used to determine the diffusion coefficient. If both axes of the plot are converted to log scale, then the slope determines the scaling exponent. The diffusion coefficient ($D$) is an average measure of the stochastic activity of a random walker. This coefficient provides an indication of the relative stability of the system. The scaling exponent describes the tendency of a particle (i.e., COP) to drift toward or away from a relative equilibrium point. From the stabilogram-diffusion plot, it becomes apparent that there are at least two distinct regions on the plot (a short-term region, which tends to be below 1-2 s, and a long-term region of larger time intervals). This transition point is referred to as the critical point. It is quantified by the critical point coordinates - the critical time interval and critical value. It has been suggested that the critical point gives an indication of when postural control changes from a primarily open-loop to a primarily closed-loop control process (Collins and De Luca 1993). The diffusion coefficient ($D$) is one half of the slope of the resultant log-linear plot of the mean square COP distance ($\langle \Delta f^2 \rangle$) as a function of the time interval between assessment points ($\Delta t$), where $\Delta j$ represents the distance in the AP, ML, or RAD direction for a given $\Delta t$. The scaling or Hurst exponent ($H$) describes the relationship between successive COP locations at increasing time intervals. $H$ is one half of the slope of the resultant log-log plot of $\langle \Delta f^2 \rangle$ as a function of $\Delta t$. Slopes are determined using the method of least squares (Collins and De Luca 1993). The point of intersection between the short and long term regions of the linear-log plot is the critical point C. The coordinates for the critical point ($CT$, $CV$) provide the measures of the critical time interval, i.e., $CT = \Delta t_c$, and critical value, $CV = \langle \Delta f^2 \rangle_c$. In our work, SDA parameters are calculated based on the algorithm and MATLAB code by Collins and Stamp (1997). For a detailed review see (Collins and De Luca 1993).

SDA has been used to study, for example, aging and fall risk (Laughton et al. 2003), individuals with Parkinson’s disease and profound vestibular loss (Mitchell et al.
1995; Dozza et al. 2005), and load carrying (Schiffman et al. 2006). It has been shown that SDA may be more sensitive than traditional measures at detecting subtle age-related differences in postural control (Collins et al. 1995).

3.2 Experimental design (appropriate trial length and number)

There is little standardization in data collection protocols (e.g., trial number and length) used to analyze COP. For example, the number of trials in past studies have ranged from 1 to 10 trials, and trials have varied from 30 to 90 s in length (Collins and De Luca 1995; Collins et al. 1995; Bosek et al. 2005). To compare results across clinics, research groups, and participant populations, reliable protocols and assessment techniques should be used. Minimal work has been done to assess the reliability of these traditional and SDA COP measures during quiet standing (Goldie et al. 1989; Collins and De Luca 1993; Le Clair and Riach 1996; Schiffman et al. 2006). To that end, we investigated the reliability of COP measures using Generalizability Theory in order to propose optimal experimental protocols that produce acceptable levels of reliability (Doyle et al. 2007, 2008).

Generalizability Theory (G-Theory) is a statistical technique that allows researchers to investigate measurement design and reliability. G-Theory is a reinterpretation of Classical Test Theory that allows one to investigate the reliability of a measure over a number of measures (Shavelson and Webb 1991). G-Theory, however, also allows researchers to investigate possible sources of error (or facets) to the specific measurement (Safrit and Wood 1989). For example, in COP measures of quiet standing, trial length and number of trials can be factors that influence reliability. Identifying sources of error enables researchers to examine which aspects of the experimental protocol influence the reliability of the scores and then design a better, more efficient measurement procedure to reduce error and achieve the desired reliability.

G-Theory provides an estimate of the variance components of the selected facets (G-study), and this information is used to design a measurement procedure that results in acceptable reliability (D-studies). First, the Generalizability Study (G-study) is performed. The G-study determines how the individual facets and facet interactions contribute to the measurement variance. In our works, the facets were participants (P), number of trials (T), and length of trials (L), and their interactions (P x T; P x L; T x L; P x T x L, e). The term “e” represents error attributed to unidentified facets. The output of the G-study is the percent of overall variance that can be contributed to each facet or interaction. Decision Studies (D-studies) use the results of the G-study to design a measurement protocol that reaches a desired reliability level, i.e., minimize error and maximize reliability (Shavelson and Webb 1991). D-studies adjust the desired facet (s) and provide feedback on the reliability of different protocol designs in the form of the G-coefficient (G). The G-coefficient is analogous to the reliability coefficient (r) of the Classical Test Theory (Shavelson and Webb 1991; Brennan 2001). G-coefficient values of 0.8 or higher are desired, with 0.7 to 0.79 considered acceptable (Safrit and Wood 1989).

In our studies, we examined the effect of these facets on a limited set of traditional COP measures (AP and ML standard deviation, mean sway velocity, and 95% confidence ellipse area) (Doyle et al. 2007) and the complete set of SDA measures (short- and long-term diffusion coefficients, scaling exponents and critical point coordinates in three directions) (Doyle et al. 2008). Adult subjects (19.9 ± 1.3 y) conducted ten trials with 90 s trial length under both eyes open and eyes closed conditions. The COP measures were then calculated using the first 30 s, 60 s, and 90 s of each trial, and averaged over one to ten trials. Results from the G-study for both groups of measures found that the number of trials did not contribute much to the overall variance (range = 0-1.2% for traditional measures, and 0-12.3% for SDA). Trial length contributed slightly more to the overall variance (range = 0-7.6% and 0-27.6%, respectively). The number of trials and trial length tended to result in larger percent variance in SDA measures than traditional summary measures: highlighting the sensitivity of SDA measures to these facets due to the statistical mechanics nature of the SDA technique that assumes large data sets. The participant facet (P) was found to be the largest contributor to variance. These findings suggest that the majority of the variance in the investigated COP measures is primarily due to differences between par-
ticipants. Adding trials or increasing trial length once an acceptable level of reliability is achieved does not significantly reduce variability.

The D-studies for both group of measures found that acceptable levels of reliability differed by measure and visual condition, such that different combinations of trial lengths and number achieved the desired condition of $G \geq 0.7$. For example, standard deviation showed improved reliability during the eyes closed condition. AP standard deviation ($SD_{AP}$) with eyes closed reached acceptable reliability levels based on four 30 s trials ($G = 0.71$) or three 60 s trials ($G = 0.70$). Whereas, $SD_{AP}$ with eyes open reached an acceptable reliability after five 60 s trials ($G = 0.71$). The 30 s trial length for the eyes open condition did not reach the acceptable level of reliability. Mean velocity reached acceptable reliability with the least number of trials, $G = 0.77$ with eyes open and $G = 0.79$ with eyes closed based on two trials at 30 s trial length. The diffusion coefficient for all conditions also reached acceptable reliability with two trials at 30 s. This result suggests that if a researcher is interested in only examining mean sway velocity and diffusion coefficients, then only two 30 s trials are necessary. See (Doyle et al. 2007, 2008) for detailed data tables and graphs that provide G-coefficient results for each COP measure as functions of the facets (trial number and trial length). The results from these papers can be used to assess reliability of the COP measures that were examined in a completed quiet stance balance study. If designing a new study, we recommend that to achieve acceptable levels of reliability ($G \geq 0.7$) for the majority of the COP measures investigated in these two studies, researchers should use trial lengths of 60 s and no less than five trials. It should be noted that the above results and recommendations were derived from data collected on young healthy adults. Our recent work (Doyle 2008) has similar conclusions and recommendations for healthy middle-aged (40-60 yrs) and older adults (65 and over) using the traditional measures recommended by Rocchi et al. (2004) and all SDA measures (Collins and De Luca 1993).

3.3 Balance during pregnancy

The following section reviews one of our studies that applied traditional and SDA measures to examine possible static balance changes due to pregnancy (Jang et al. 2008). Pregnant women often remark that their balance degrades during pregnancy; however, minimal work has been published documenting a woman’s perception of her balance or actual measured changes in balance throughout pregnancy or after delivery (Fries and Hellebrandt 1946; Pickering et al. 1999; Davies et al. 2002; Butler et al. 2006). In this study, 30 women (15 pregnant and 15 non-pregnant control subjects) were tested at four week intervals during pregnancy and also three times over the six months after delivery. For each session, the subject performed ten 30 s quiet stance trials with her arms to the side and eyes open, looking at a stationary picture placed at eye level and 3 m away. During the first trial of each session, a new foot tracing was created to observe any changes in preferred stance width over time. The subject placed her feet within the tracing for all subsequent trials. Additionally, to quantify perceived balance degradation, the subject rated her balance on a scale from 0 (normal) to 10 (extremely unstable). The previous session’s score was provided as a reference. Based on recommendations from Rocchi et al. (2004), nine traditional COP parameters were examined: standard deviation of the displacement about the mean ($SD_{dev}$) in the AP, ML, and RAD directions; mean sway velocity ($Vel$) in the three directions; AP and RAD 95% power frequency ($95 Freq$); and angular deviation. SDA parameters reported in this study were short-term and long-term diffusion coefficients ($D_s, D_l$) and scaling exponents ($H_s, H_l$) in the three directions. Reliability of these traditional and SDA measures have been found to be fair to excellent, ranging from 0.56 to 0.95 (Doyle et al. 2007, 2008).

Perceived degradation in balance during pregnancy was found to strongly relate to increasing postural instability in the AP direction and increasing stance width. Lateral stability was maintained during pregnancy and likely accomplished by increasing stance width. For the pregnant group, perceived balance degradation and stance width were highly correlated ($r = 0.94$). Both increased during pregnancy ($p \leq 0.016$) and dropped to near-control levels after delivery ($p \leq 0.004$). Compared to the control group, pregnant subjects displayed increased sway, especially in AP and RAD measures of $StDev$, $D_s$, $D_l$, and $H_l$ ($p \leq 0.039$). Anterior-posterior
sway measures (StDev, Vel, Ds) strongly correlated with perceived balance (0.82 > r > 0.72) and also decreased significantly between the third trimester and postpartum (p ≤ 0.029). Interestingly, ML balance measures varied little during pregnancy, but increased after delivery. The postpartum subject was found to return to her pre-/early-pregnancy stance width. In normal healthy individuals, increasing stance width has been shown to reduce postural sway, especially lateral sway (Kirby et al. 1987; Day et al. 1993). These findings with the pregnant subjects may suggest that there is a residual effect of stance width reductions that resulted in increased lateral sway.

SDA results suggest differences in postural control mechanisms due to pregnancy, in that greater postural sway activity was noted among the pregnant group. Group differences were particularly apparent for postural sway during long time intervals in the AP and RAD directions (Ds,α, Ds,αα: p ≤ 0.018). These results suggest that, compared to controls, pregnant subjects had increased instability during periods of closed-loop control. Short-term postural sway in the AP direction (characterized by Ds,α) was found to increase during the latter stages of pregnancy and drop postpartum. Ds,AP was the only SDA parameter to correlate with perceived balance degradation and stance width. This result suggests strong relationships exist between open-loop AP postural control, balance perception and base of support size. Long-term postural sway in the mediolateral direction (Ds,αα) was found to increase in pregnant subjects especially postpartum when compared to controls. This suggests increasing ML instability during periods of closed-loop control particularly after delivery. Little to no changes in short-term and long-term scaling exponents (Hs, Hl) suggest little effect of pregnancy on the COP drift tendencies about a relative equilibrium point.

4. Assessing dynamic balance

Preventing falls, especially in older adults, is a worldwide health initiative. To this end, understanding how dynamic balance mechanisms and balance recovery ability after an unexpected perturbation are affected by age has become an active research area. Upright stance is usually maintained with a variety of possible sway responses in the event of mild to moderate perturbations. Sway about the ankles alone or ankles and hip are two commonly observed recovery behaviors, referred respectively to as ankle and hip strategies (Horak et al. 1989). In the event of large perturbations, stepping is the primary means to recover balance (Hsiao and Robinovitch 1998). In this section, we discuss how dynamic balance and recovery ability can be examined due to large and small perturbations. We start by discussing work on balance recovery strategies that require one or more steps to restore upright stance.

4.1 Balance recovery after a simulated fall

The tether-release method provides a useful experimental paradigm for assessing biomechanical and age-related differences in the response used after a simulated fall (Hsiao-Wecksler 2008). This method provides a repeatable perturbation that simulates the initial unbalanced configuration of the body during a trip or slip. In this technique, the test subject is released from an initial forward or backward inclined position (Fig. 3).

This technique has been used to explore various biomechanical parameters, or step characteristics, in an effort to understand which attributes allow for successful balance recovery by stepping or stumbling (see review in (Hsiao-Wecksler 2008)). These parameters are a combination of kinematic and kinetic metrics, such as step timing, step length, joint kinematics and joint torques. The time at step contact was found to associate with the magnitude of the initial lean angle θi such that step contact time decreased with increasing lean angle until a plateau was reached at approximately 460 ms (Do et al. 1982; Hsiao and Robinovitch 1999). Age-related differences were found for women (Wojcik et al. 1999; Hsiao-Wecksler and Robinovitch 2007) and marginally for men (Thelen et al. 1997), such that older adults used longer times than young. Step length was found to increase as θi increased (Do et al. 1982; Thelen et al. 1997; Hsiao and Robinovitch 1999; Wojcik et al. 1999). Conversely, larger magnitudes of θi were achievable by taking longer steps (Hsiao-Wecksler and Robinovitch 2007). For a given non-taxing lean angle, mixed results have been found regarding age-related differences in recovery step length. Theilen et al. (1997) noted a non-
Fig. 3 Simulation of a forward fall using tether-release method. The subject is suddenly released from an initial lean angle $\theta_0$ after triggering a release mechanism. (Reprinted from (Hsiao-Wecksler 2008) with permission)

significant trend with male subjects such that older subjects used slightly shorter step lengths than young subjects. Whereas, Wojcik et al. (1999) found that older female subjects used significantly longer step lengths than young females and older males. However, when challenged to their maximum recoverable lean angle, older women used shorter steps than young women (Hsiao-Wecksler and Robinovitch 2007). Joint range of motion and peak angular velocities were also found to increase with increasing lean angle (Wojcik et al. 2001; Madigan and Lloyd 2005). At respective maximum lean angles, young adults used larger knee and hip ROM (Wojcik et al. 2001) and faster peak joint velocities (Madigan and Lloyd 2005) than older adults. Joint torques appear to be affected by gender, age and phase in the stepping response. For example, during the initiation and swing phases of the stepping response, older women used larger peak ankle and knee torques in the stepping leg than young women for a given lean angle (Wojcik et al. 2001). We found that during the post-contact phase young women used larger ankle and hip torques than older women, while knee torques were similar between age groups (Hsiao-Wecksler and Robinovitch 2007). Opposite trends were found with male subjects, i.e., older men used larger ankle and hip torques than young, and knee torques were smaller during the post-contact phase (Madigan and Lloyd 2005).

We also developed a mathematical model to further explore how these biomechanical parameters affect balance recovery by stepping (Hsiao and Robinovitch 1999). In the model, the stance leg and trunk were simulated with a single-link inverted pendulum and the stepping leg with a linear spring. Our model predicted that successful balance recovery is governed by a coupling between step length, step contact time, and leg strength; therefore, declines in one capacity must be offset by enhancements in the others to ensure successful balance recovery. This suggests that one’s risk for falls may be affected more by small but diffuse neuromuscular impairments than by a larger impairment in a single motor capacity. This finding supports epidemiological evidence that fall risk increases dramatically with number of impaired motor functions (Tinetti et al. 1988).

These tether-release studies found strong associations between recovery ability (maximum lean angle) and biomechanical parameters such as step length, step timing, and joint torques. These results point to the importance of neuromuscular capacities that relate to lower extremity flexibility, reaction time, and strength. Therefore, the maintenance or enhancement of these necessary attributes should be considered when developing fall intervention programs for older adults.

4.2 Postural control response to a mild impulse perturbation

Previous studies of dynamic postural control have focused mainly on using continuous or persistent perturbations, e.g., (Maki et al. 1990; Peterka 2002). Most losses of balance, however, are often sudden momentary events that occur during gait (e.g., slips or trips) or while performing activities of daily living (e.g., standing, sitting, reaching, pushing, or pulling) (Cumming and Klineberg 1994). In our more recent work, we have been interested in examining the response of the postural con-
trol system to a mild, but sudden impulse perturbation (Hsiao-Weckslers et al. 2003; Hur et al. 2007). In these studies, a weak impulse backward tug force is applied to generate a postural sway response (Fig. 4). We have used this experimental paradigm to propose a new metric for assessing the robustness of the postural control system based on the sensitivity function (Hur et al. 2007). We also demonstrate that it is possible to predict this perturbed-stance postural sway response using only COP data derived when standing quietly (Hsiao-Weckslers et al. 2003).

4.2.1 Robustness Analysis

To assess the relative stability of an individual to a mild perturbation, a new metric was proposed for characterizing the robustness of the postural control system to an impulsive perturbation (Hur et al. 2007). Robustness, or relative stability, of a modeled system was quantified as the maximum magnitude of the sensitivity function of the modeled system. The sensitivity function describes how sensitive a system is to small perturbations in the system; larger values indicate reduced robustness or decreased relative stability of the system. Alternative measures of robustness are the gain margin and phase margin, which relate to the overall gain and phase delay of the system, respectively. Recently, the robust space for a model of the postural control system based on a time-delayed proportional-derivative (PD) controller was outlined by using the individual gain and phase margins but not the combined sensitivity function (Masani et al. 2006). In our study, we use spectral analysis system identification (Peterka 2002) to fit experimental response data to a postural control model consisting of a single-link inverted pendulum (of length h, mass m, and moment of inertia J) modulated by a time-delayed PD controller (with gains $K_p$, $K_d$ and time delay $\tau$), passive torque generator (with stiffness $k$ and damping $b$), and negative feedback loop. For this system, the input signal was the applied tug force ($F$) and the output signal was body lean angle ($\theta$) relative to vertical. Lean angle was derived from AP COP data based on a procedure derived from the gravity-line projection method proposed by Zatsiorsky and Duarte (2000). Therefore, for this model, the transfer ($TF$) and sensitivity ($S$) functions can be determined using the superposition property for two independent inputs ($\theta, F$) such that:

$$\theta = TF(s) \theta + S(s) F$$

where,

$$TF(s) = \frac{(K_p + K_d s) e^{-\tau} + k + bs}{Js^2 + bs + k - mgh + (K_p + K_d s) e^{-\tau}}$$

(2)
Assessing quiet and perturbed balance

\[ S(\dot{z}) = \frac{K_i}{J_s^2 + bs + k - mgh + (K_o + K_i) e^{-\mu}} \quad (3) \]

the desired reference lean angle corresponds to upright stance (i.e., \( \theta = 0 \)). \( K_i \) represents the lever arm of the force around the ankle and converts \( R \) to a torque, and \( g \) is gravitational acceleration. The robustness of the system was determined from the maximum value of the identified sensitivity function.

To examine the efficacy of this model and measure, we tested thirty healthy subjects across three age groups: young (20-30 yrs), middle-aged (42-53 yrs), and older adults (71-79 yrs). Each subject performed 10 quiet stance and 10 perturbed stance (tug) trials, which were randomly presented and all 30 s. Supplemental examination of the quiet stance COP data found significant age-related differences among traditional and SDA measures of quiet stance sway, which agree with previous studies (e.g., (Collins et al. 1995; Prieto et al. 1996)). These results support that this older adult group of subjects has reduced static balance performance. From the perturbed stance data analysis, maximum sensitivity values were found to be significantly larger for older adults than young or middle-aged adults \( (p<0.05) \). These results suggest that the maximum sensitivity may be a good metric for assessing robustness or relative stability of the dynamic postural control system. Further studies are needed to assess the efficacy of this approach for detecting differences in dynamic system robustness in other populations.

4.2.2 Predicting dynamic postural response from quiet-stance behavior

As suggested above, the postural control system used during quiet stance has been represented as a stochastic process (Collins and De Luca 1993; Newell et al. 1997; Chiari et al. 2000). A statistical mechanics theorem that applies to many stochastic systems is the fluctuation-dissipation theorem (Kubo 1966). This theorem provides a relationship between the fluctuations of a quasi-static, stochastic system to the system’s response following a perturbation. A similar linear relationship was applied to the postural control system to explore whether AP fluctuations of the COP during quiet stance can be used to predict the AP response of the postural control system to a mild backward tug force in young adults (Lauk et al. 1998) and older adults (Hsiao-Weckslers et al. 2003). The decay of a correlation function derived from the AP COP during quiet-stance, \( dC(t)/dt \), was proposed to predict an individual's AP sway response to a weak backward tug, \( R(t) \). This relationship was defined as \( R(t) = a + b(dC(t)/dt) \), where \( a \) is typically small and reflects an arbitrary origin for the COP, and \( b \) has arbitrary units due to normalization of all trials. This relationship was evaluated using data collected from 10 quiet stance and 10 perturbed stance trials of 30 s length, which were randomly presented.

These studies found that the linear relationship between quiet- and perturbed-stance was applicable to the postural control system of 19 of 20 young (19-30 yrs) and 8 of 10 older (66-73 yrs) adult subjects (Lauk et al. 1998; Hsiao-Weckslers et al. 2003). The existence of this relationship with respect to the human postural control system further suggests that, for a given individual, the postural control system may use the same control mechanisms during quiet-stance and mild-perturbation conditions, regardless of age. These results suggest that it is possible to predict an individual's dynamic response to a mild perturbation using quiet-stance data. These findings may have promising clinical implications. They suggest that it may not be necessary to perform tests that involve mild perturbations to assess postural control capacity in healthy adults; rather, quiet stance data can be collected and used to predict dynamic balance performance.

Further studies are necessary to determine whether this technique is applicable to individuals with balance impairments and to tests using larger perturbations.

Some groups consider that perturbed upright stance, which results in no change in of the base of support, utilizes “static” balance since the body essentially stays in the same location; whereas walking or locomotion uses “dynamic” balance since the base of support changes and body position varies substantially.

Some equations to compute the 95% confidence ellipse are presented incorrectly in both Prieto et al. (1996) and Doyle et al. (2007). See Oliveira et al. (1996) for alternate expressions to calculate this measure.
Acknowledgements

The author would like to thank her collaborators and students that contributed to the presented work, especially Karl Rosengren, Jim Collins, Steve Robinovitch, Rich Doyle, John Jang, Pilwon Hur, and Brett Duiser.

References


Assessing quiet and perturbed balance


Profile

Elizabeth T. Hsiao-Wecksl

Dr. Elizabeth T. Hsiao-Wecksl received her PhD in Mechanical Engineering from the University of California-Berkley in 2000. She was a post-doctoral fellow in the Integrated Rehabilitation Engineering Program at Boston University and Harvard Medical School from 2000-2002. She is currently an Assistant Professor in the Department of Mechanical Science and Engineering, Director of the Human Dynamics and Controls Laboratory, and Affiliate in the Department of Bioengineering and Department of Industrial and Enterprise Systems Engineering at the University of Illinois at Urbana-Champaign. Her research focuses on posture and movement control with an emphasis on developing new measurement tools for quantifying changes in gait and balance.