*Manuscript

Revision submitted to 'Medical Engineering & Physics' – September 2009

Posturographic Measures in Healthy Young Adults during Quiet Sitting in Comparison with Quiet Standing

Running Title: Measures of Posturography during Quiet Sitting

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September 28, 2009

Number of words in Text: **5,700**

Number of words in Abstract: **254**

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ABSTRACT

Measures of postural steadiness – known as posturography – are commonly used for balance assessment during quiet standing. Although quiet sitting balance may be studied via posturography as well, this has not been done to date. As such, the purpose of this study was to characterize the posturography during quiet sitting in comparison with quiet standing and to provide a benchmark for future studies investigating differences in balance regulation and execution. Twelve young and healthy people agreed to quietly sit and stand on a force platform with their eyes open and closed. For each condition, one trial of two minutes was executed and the anterior-posterior, medial-lateral, and resultant distance fluctuations of the body's center of pressure (COP) were calculated. Finally, time-domain, frequency-domain, and stabilogram diffusion function (SDF) measures were identified and compared for all COP time series. The results consistently indicate that, for quiet sitting, the body sway size and velocity were smaller and the power-weighted average frequency larger than for quiet standing. Moreover, the SDF analysis revealed that quiet sitting shows fewer drifts over short time intervals, but also fewer controlled adjustments in the longer term to bring the system back to equilibrium. The observed differences can be partially explained by biomechanical and dynamic differences of the body portions that are in motion during quiet sitting and standing. The SDF analysis suggests, however, that also the balance control strategies are not identical. These findings may be especially useful for the assessment of sitting balance and the development of novel balance rehabilitation techniques and assistive devices.

INDEX TERMS – Center of pressure, human posture, posturography, sitting balance, standing balance.

I. INTRODUCTION

Posturography – the use of measures of postural steadiness – is a commonly applied and accepted technique for assessing postural stability and control during upright standing. A necessary condition of postural stability is that both the vertical projection of the whole-body center of mass (COM) and the center of pressure (COP) on the ground remain within the base of support [1-4]. Since it is believed that the COP summarizes the neuromuscular control response to the imbalances of the body's COM [5], COP measurements do not only contain valuable information on the overall COM fluctuation [6], but also on the utilized balance control strategy. In fact, a number of studies have used COP time series in an attempt to characterize the control mechanisms responsible for balance control during quiet stance [7-10].

The COP displacement during standing, as measured by a force platform, can be quantified in both dynamic and static conditions. During *dynamic* conditions, external forces (e.g., via platform translation) or sensory disturbances (e.g., visual or proprioceptive disturbances) are applied to the body to study postural responses and investigate underlying balance mechanisms. During *static* conditions, no disturbances are applied, implying that the COP displacement corresponds to spontaneous body sway and its control during quiet stance [7,11]. As a result, COP *parameters* – derived from the variations in COP displacement during quiet standing – have been used to characterize balance abilities and underlying control strategies among different subject populations, including healthy individuals [8-10,12-14] and individuals with spinal cord injury [15], Parkinson's disease [16], or haemophilic arthropathy [17].

Posturographic analyses commonly applied to quiet standing include time- and frequency-domain measures [12] as well as stabilogram diffusion function (SDF) measures [8-10]. On the one hand, time- and frequency-domain measures have been used to quantify the displacement, velocity, area, and frequency properties of the COP fluctuation [12]. On the other hand, SDF analysis was developed by Collins and de Luca [8] to characterize the nature of the applied balance control strategy by identifying the underlying stochastic activity from time-varying COP coordinates. Using this method, a two-part control behavior has been suggested during quiet standing: an *open-loop* control scheme implying no neural control over short time intervals, and a *closed-loop* control scheme implying the presence of neural feedback control over longer time intervals [8].

However, being able to characterize balance abilities and underlying control mechanisms may not only be important for standing, but also for sitting: the seated body needs to ensure a stable posture at all times and return to equilibrium position following internal and external perturbations [18-21]. The human spine is inherently unstable [22,23] and must be stabilized via activation of the abdominal and back muscles [24,25]. Various degrees of damage to the motor and sensory cortices (e.g., due to acquired brain injury or stroke) or the spinal cord (e.g., due to traumatic spinal cord injury) may lead to difficulties in maintaining trunk stability during sitting – as the impaired central nervous system might not be able to issue appropriate control commands to the muscles involved in balance control anymore. Consequently, the trunk muscles cannot provide adequate levels of stabilizing forces, which will decrease the person's ability to perform many functional tasks during sitting. In this context, posturography for sitting might have the potential to not only characterize and quantify the balance abilities of different populations during sitting, but also to identify neural control deficits that may be responsible for different degrees of sitting balance impairments. Similar to quiet standing, the posturography for quiet sitting could consequently be applied as an assessment tool for sitting balance and for the development of novel balance rehabilitation techniques and assistive devices.

The use of posturography to assess balance during sitting has mainly been reported for seated infants and adolescents [26-28], adults during a seated reaching task [18,29-31], and adults sitting on an unstable surface [32-34]. Moreover, sitting balance has been studied in dependence of respiratory perturbations [35,36] and sitting posture [36]. However, posturography in adults during quiet sitting – analogous *and* in comparison to quiet standing – has not been well documented to date. As such, there is a need to characterize the posturography during quiet sitting in comparison with quiet standing and to provide a benchmark for future studies investigating differences in balance regulation and execution for both tasks and various subject populations. Accordingly, the two main objectives of the present study were to: 1) provide a comprehensive set of posturographic measures for sitting and standing in healthy individuals that were identified under essentially identical experimental conditions; and 2) determine potential biomechanical, dynamic, and control differences between sitting and standing on the basis of the obtained measures.

II. METHODS

A. Subjects

Twelve healthy and young individuals (10 male and 2 female[1]; age 27.7 ± 6.6 years; height 176.2 ± 9.3 cm; weight 75.8 ± 13.6 kg) were invited to participate in this study. None of the subjects had any known history of neurological or musculoskeletal disorders or suffered from acute or chronic back pain. At the time of the experiments, the subjects were also taking no medications such as antihistamines that can cause drowsiness and, thus, interfere with their balance abilities. Each subject gave written informed consent to participate in the study, which was approved by the local ethics committee in accordance with the declaration of Helsinki on the use of human subjects in experiments.

B. Experimental Procedure and Data Acquisition

During the experiments, each subject was asked to quietly sit (SI) or quietly stand (ST) on a force plate with his or her eyes open (EO) or closed (EC). To cover all four task-eye condition combinations, four separate trials of 120 seconds each were recorded (4*120 seconds). Note that this sample duration was chosen based on the findings by Carpenter *et al.* [39] who recommended a trial length of 120 seconds to increase the reliability and low-frequency sensitivity of the identified COP measures. The order of the four trials per subject was randomized provided that the two sitting (standing) trials were executed back-to-

[1] One of the limitations of this study is the gender imbalance of the subject sample. Strong evidence has been presented, however, that there is no gender difference in measures of postural steadiness during quiet standing [37,38].

back (e.g., SI-EO, SI-EC, ST-EC, ST-EO or ST-EC, ST-EO, SI-EC, SI-EO). In between the four trials, the subject was able to rest for as much time as he/she needed.

The force plate was a split force plate (AccuSway$^{\text{Plus}}$ dual force platform, Advanced Mechanical Technology Inc., MA, USA) that was placed either on the floor (ST) or on top of a custom-made seating apparatus (SI) made of medium density fiberboard. During the sitting trials, each subject was asked to maintain a comfortable and natural quiet sitting posture, with his or her arms crossed in front of the chest. The lower extremities were not fixated to the seat whose height was chosen such that the subject's feet did not touch the floor, preventing the contribution of the lower extremities to the stabilization act. During the standing trials, each subject was asked to maintain a comfortable and natural quiet stance posture standing barefoot, with the arms hanging loosely along the sides of the body. Note that the subject was free to choose his/her foot position [12] as it has been shown that posturographic measures are affected by foot placement [11] and that 'unnatural' or 'uncomfortable' foot positions should therefore be avoided [40]. For both tasks, the force plate signals were collected at a sampling frequency of 100 Hz [12] using a 64-channel, 12-bit analog-to-digital converter (NI 6071E, National Instruments, TX, USA).

C. Data Processing and Analysis

The acquired force time series of each trial were used to calculate the COP fluctuation in the anterior-posterior (AP) and medial-lateral (ML) directions [41]. After subtracting respective mean values from both time series, the resultant distance (RD) time series [12] was then determined using matching points in time of the AP and ML data. Finally, a fourth-order, zero phase-lag, low-pass Butterworth filter with a cut-off frequency of 5 Hz [12,15] was applied to all COP displacement time series (AP, ML, and RD). Since kinematics and COP recordings during postural balance are generally dominated by very slow frequency components of up to 1 Hz [5,42], a cut-off frequency of 5 Hz [12,15] was applied to eliminate high-frequency noise artifacts that generally affect these recordings. To ensure an efficient decay of the power of higher (noise) frequencies and at the same time minimize the delay introduced during filtering, the fourth-order and zero phase-lag characteristics [12,15] were chosen, respectively.

To quantify and compare the postural steadiness of the subjects during sitting and standing, (1) *time-domain*, (2) *frequency-domain*, and (3) *stabilogram diffusion function* (SDF) analyses described by Maurer and Peterka [7] were performed with the three COP time series (120 seconds of AP, ML, and RD each). In the present study, two commonly used measures were identified for each of the three posturography categories above, resulting in a total of 18 measures (six for each of the AP, ML, and RD time series). Note that local divergence measures that have been used to compare local dynamic stability during walking and standing [13] were not included in our analysis. Kang and Dingwell revealed that, for standing, these measures correlate with traditional posturographic measures and at the same time agree with the two-part control behavior found in SDF analyses [13]. As such, it can be assumed that the posturographic measures identified from time-domain, frequency-domain, and SDF analyses provide sufficient information for the comparison between quiet sitting and standing.

1) Time- and Frequency-Domain Measures

The AP, ML, and RD time series were quantified by means of the following time- and frequency-domain measures: the mean distance (MD), the mean velocity (MV), the centroidal frequency (CFREQ), and the frequency dispersion (FREQD). CFREQ is the frequency at which the spectral mass is concentrated (power-weighted average frequency), whereas FREQD is a unit-less measure of the variability in the frequency content [7,12]. Following the procedure applied by Maurer and Peterka [7] and Prieto *et al.* [12], the two frequency-domain measures were calculated for the frequency range from 0.15 to 5.0 Hz. The required one-sided spectral density function *G(f)* was determined for each of the AP, ML, and RD time series using Welch's method. Note that the time- and frequency-domain measures were selected as to cover all three correlation groups found by Maurer and Peterka (Fig. 3 in [7]) using quiet stance simulations. In particular, MD originated from correlation group 1, MV and FREQD from correlation group 2, and CFREQ from correlation group 3 [7].

2) Stabilogram Diffusion Function Measures

SDF plots as described by Maurer and Peterka [7] and Collins and de Luca [8] were identified for all time series to derive the short- and long-term Hurst exponents H_S and H_L, respectively. Note that the Hurst exponent is a real number between 0 and 1 that measures the autocorrelation of the stochastic process. H_S and H_L are specifically calculated from the slopes of lines fitted to the short- and long-term regions of the log-log SDF plot that can be determined via the following scaling law [8]:

$$< \Delta x^2 > \sim \Delta t^{2H} \quad , \tag{1}$$

where $< \Delta x^2 >$[2] is the mean square COP displacement for a particular time interval Δt [7,8]. In an open-loop control scheme, H has been suggested to be larger than 0.5 (positive correlation of the stochastic process), whereas in a closed-loop control scheme, H is presumably smaller than 0.5 (negative correlation of the stochastic process) [8]. In accordance with the procedure applied by Maurer and Peterka [7] and Collins and de Luca [8], the short-term region of the SDF was defined as to include data for time intervals (Δt_S) ranging from 0 to 0.5 seconds, and the long-term region was defined as to include data for time intervals (Δt_L) ranging from 2 to 10 seconds.

3) Statistical Analysis

In order to identify potential differences between the two tasks (SI and ST) and eye conditions (EO and EC), the posturography results were analyzed using multiple statistical tests. In a first analysis, Wilcoxon tests of signed ranks were applied to all measures to identify significant differences in the group means between (1) quiet sitting and quiet standing (for each eye condition) and between

[2] The angled brackets <·> denote an average over time or an ensemble average over a large number of samples.

(2) eyes open and eyes closed (for each task). In a second analysis, Spearman's rho correlation coefficients were computed for all measures to reveal potential correlations between quiet sitting and standing for both eye conditions[3]. In all analyses, a significance threshold of $\alpha = 0.05(*)$ was used to prevent excessive false-positive results. Additional information on the strength of significance was provided by indicating $P < 0.01(**)$ [12].

[3] Non-parametric analyses were applied as we cannot be certain that the identified measures are normally distributed. Wilcoxon tests of signed ranks were chosen over a non-parametric ANOVA equivalent (Friedman's test) as the Wilcoxon tests are more descriptive, testing also the components of the interaction (while being mathematically identical for the case of two factors with two values each [43]).

III. RESULTS

In Fig. 1, examples of the COP time series during sitting and standing can be seen for one subject for EO condition. Figure 1A shows the AP (top) and ML (bottom) time series during sitting (left) and standing (right), whereas Fig. 1B depicts the planar phase plots of the COP fluctuation during sitting (top) and standing (bottom). Note that only the initial 15 seconds of data are presented for each task in order to reveal the fluctuation behavior. For example, a simple visual inspection suggests a smaller COP fluctuation of higher frequency for sitting when compared to standing (both AP and ML).

Examples of the log-log stabilogram diffusion functions (SDF) calculated from the COP data during sitting and standing (EO) are presented in Fig. 2 (AP and ML). Figure 2A depicts the log-log SDF plots for one subject during quiet sitting, whereas Fig. 2B shows the log-log SDF plots for the same subject during quiet standing. The dashed lines represent linear regression fits used to estimate the Hurst exponents H_S and H_L. A visual inspection suggests that the values of H_S and H_L – based on the slopes of lines fitted to the short- and long-term regions – were slightly different for the two tasks.

The first two columns of Table I list the means and standard deviations of all identified posturographic measures for both tasks and eye conditions. To generally compare body sway during sitting and standing, also the ratios between SI and ST for corresponding measures are given for each eye condition (third column of Table I). For the time-domain measures, the ratios ranged from 0.66 (MD_{AP}) to 0.81 (MV_{ML}) and from 0.33 (MD_{AP}) to 0.60 (MV_{ML}) for the EO and

EC conditions, respectively. As such, body sway was presumably smaller and slower during quiet sitting compared to quiet standing (which agrees with the visual inspection of Fig. 1). For the frequency-domain measures, the ratios of quiet sitting to quiet standing ranged from 1.06 ($FREQD_{RD}$) to 1.82 ($CFREQ_{AP}$) and from 1.08 ($FREQD_{RD}$) to 1.79 ($CFREQ_{AP}$) for the EO and EC conditions, respectively. These results indicate that spontaneous body sway during quiet sitting may be characterized by a higher centroidal frequency, but also a larger frequency variation compared to quiet standing (which again agrees with the visual inspection of Fig. 1). Finally, for the SDF measures, it was found that H_S was consistently larger than 0.5 for both tasks (positive correlation of the stochastic process), whereas H_L was consistently smaller than 0.5 for both tasks (negative correlation of the stochastic process). The ratios of quiet sitting to quiet standing ranged from 0.78 to 0.82 and from 1.30 to 2.74 for H_S and H_L, respectively (which agrees with the visual inspection of Fig. 2). These ratios in combination with the actual values of H_S and H_L imply a less positively correlated stochastic activity in the short-term region of quiet sitting and a less negatively correlated stochastic activity in the long-term region of quiet sitting when compared to quiet standing.

The last two columns of Table I summarize the results for the statistical analyses that were performed to identify significant differences and correlations between tasks and eye conditions. On the one hand, the Wilcoxon test between tasks (‘task’ in second last column of Table I) revealed significant differences in the magnitude of the measures between quiet sitting and quiet standing for most of the measures and both eye conditions. These findings support the aforementioned proposition that body sway during quiet sitting (1) is *smaller* and *slower*; (2) is characterized by a *higher* centroidal frequency and a *larger* frequency variation; and (3) exhibits a *less positively* (short-term region) and *less negatively* (long-term region) correlated stochastic activity in comparison to quiet standing. On the other hand, the Wilcoxon test between eye conditions (‘eye’ in second last column of Table I) indicated that none of the measures during quiet sitting exhibited significant differences between EO and EC, whereas the eye condition affected all time-domain measures during quiet standing. Finally, it was found using Spearman’s rho correlation coefficients (last column of Table I) that the mean COP velocity for the EO condition represented the only measure that was correlated between the two tasks, but this for all COP time series. The top panel of Fig. 3 shows the MV correlation for the AP fluctuation, the center panel the MV correlation for the ML fluctuation, and the bottom panel the MV correlation for the RD fluctuation (eyes open).

IV. DISCUSSION

A. Effects of Sitting versus Standing Posture

The performed posturography revealed that the time-domain measures during quiet sitting were consistently smaller and the frequency-domain measures consistently larger than during quiet standing (Table I). These differences in the COP fluctuation can be partially explained by the biomechanical and dynamic differences of the body portions that are in motion during the two tasks. In other words, the smaller size and moment of inertia of the moving trunk during sitting in comparison to the moving body during standing must affect the magnitude, velocity, and frequency of the body sway and, hence, of the COP displacement.

1) Biomechanical Effects on Time-Domain Measures

In the literature, the human body during quiet standing is often approximated by an inverted pendulum (IP) model consisting of a single rigid body segment that rotates about the ankle joints [7,41,44-47]. In fact, Gage *et al.* confirmed that the kinematics and kinetics of the IP model agree with those during quiet standing [48]. Following Winter's anthropometric approximations [5], the height of the IP model's COM above the ankle joint can then be calculated as x_{ST} = 0.547 h, where h is the body height. Similarly, the human body during quiet sitting can be modeled using an IP model consisting of a single body segment (head, arms, and trunk) that rotates about the hip joints [24,28,49]. In this case, the height of the COM above the hip joint can be estimated as x_{SI} = 0.273 h [5]. According to these estimations, the approximated ratio between the COM heights (sitting vs. standing) is x_{SI}/x_{ST} = 0.50. Thus, for the same body sway fluctuation (i.e., same ankle or hip joint angle changes), the magnitude and velocity of the horizontal COM fluctuation during quiet sitting can be assumed to be half the size as that during quiet standing. It has to be noted, however, that the result of the above calculations applies to COM, and not COP. Nevertheless, since the fluctuation characteristics of the low-pass filtered COP are similar to those of COM (e.g., most of the power is identical for both time series) [6], a similar ratio could be expected between the COP time series of quiet sitting versus quiet standing. Independent of the eye condition, the ratios of the COP time-domain measures in the present study ranged from 0.33 to 0.81 (Table I), which is in agreement with the simple estimate calculated above (0.50). As such, it can be concluded that biomechanical differences alone could generally account for the smaller time-domain measures in sitting compared to standing.

2) Dynamic Effects on Frequency-Domain Measures

The fact that the frequency-domain measures during quiet sitting were consistently larger than during quiet standing (up to 1.82 times; Table I) could be explained with the difference in the moments of inertia of the two systems. On the one hand, the moment of inertia of the IP model during quiet standing can be calculated as $I_{ST} = 0.299\ mh^2$, where m is the body mass and h the body height [5]. On the other hand, the moment of inertia of the IP model during quiet sitting can be estimated as $I_{SI} = 0.050\ mh^2$ [5]. Consequently, the modeled moment of inertia for standing is approximately six times larger than for sitting. Since systems with larger inertia are generally more sluggish and require more time to return to the (unstable) equilibrium position, their frequency of oscillation is – for a given control scheme – smaller than for systems with smaller inertia. More specifically, it has been reported that the natural frequency of the IP model is inversely proportional to the root of the moment of inertia [41]. Based on the dynamic properties alone, the frequency ratio estimated from the moments of inertia of the two IP models therefore suggests that sway frequencies during quiet sitting are more than two times ($\sqrt{6}$) larger than during quiet standing. Although the experimental centroidal frequency found in sitting was only up to two times larger than for standing, the discussed dynamic differences between the two systems must at least contribute to the observed differences in frequency.

B. Differences in Control Schemes

Although the time- and frequency-domain results could be to a large extent explained with biomechanical and dynamic differences between the two systems, also differences in the applied control strategies may affect the posturography in sitting and standing. Such proposition would be supported by the fact that most of the SDF measures, which can be understood as indicators for the underlying control strategy, showed significant differences between the two tasks (Table I). On the one hand, the smaller H_S in sitting indicates that the short-term, presumably open-loop postural control mechanism is less positively correlated compared to standing, and therefore perhaps less unstable. More specifically, the output of the overall system may have a lower tendency to move or drift away from a relative equilibrium point over short time intervals (as can be seen in quiet standing) [10]. On the other hand, the larger H_L in sitting indicates that the long-term, presumably closed-loop postural control mechanism is less negatively correlated compared to standing, and therefore perhaps less stable. In other words, there is a decreased probability over the longer term that movements away from a relative equilibrium point during sitting will be corrected by controlled adjustments back to the equilibrium position (as can be seen in quiet standing) [10]. Note that these two results actually complement each other: since, during quiet sitting, there are fewer drifts away from a relative equilibrium point over short time intervals (smaller H_S), fewer controlled adjustments are needed in the longer term to bring the system back to equilibrium (larger H_L).

In a comparison of stance control in the young and elderly, it was found that the control mechanism of the young is less positively correlated in the short term and less negatively correlated in the longer term [10]. The authors inferred from these results that young adults are more *mechanically* stable (e.g., due to higher joint stiffness), and therefore require less neural feedback control [10]. A similar conclusion could be drawn from the comparison between sitting and standing: sitting posture may be more mechanically stable than standing posture due to a lower center of mass as well as a higher joint stiffness, and may therefore require less neural feedback control. Note that this argumentation would also agree with the finding that quiet sitting is not sensitive to eye condition (Table I), implying a more secondary role of the visual feedback information in the control task when compared to quiet standing.

Our conclusion that concrete differences in the control schemes of the two tasks could be found is supported by the fact that solely the mean COP velocity for the eyes open condition was correlated between quiet sitting and standing

(Table I). If the underlying control strategies were the same for both tasks, more consistent correlations across all studied posturographic measures could be expected. Future work is needed, however, to fully understand the differences in the utilized control schemes.

V. CONCLUSIONS

In conclusion, the performed study provides strong evidence that biomechanical and dynamic differences between quiet sitting and standing affect the posturography and, hence, the body sway characteristics during those two tasks. In particular, the reported findings on quiet sitting reveal that body sway size and velocity are smaller and the centroidal frequency larger than for quiet standing. The performed SDF analysis and the lack of correlation between the measures of both tasks suggest, however, that also the applied balance control strategies may not be identical. Since, during quiet sitting, there are fewer drifts away from a relative equilibrium point over short time intervals (smaller H_S), fewer controlled adjustments are needed in the longer term to bring the system back to equilibrium (larger H_L).

The performed study is the first of its kind that directly compares the posturography during quiet sitting and quiet standing. As such, the obtained results can be used as a benchmark for future studies investigating specific control contributions and differences during both tasks (e.g., due to discrepancies in multi-body complexity) and for various subject populations (e.g., young versus elderly people or after traumatic spinal cord injury). The gained knowledge may be especially useful as an assessment tool for sitting balance (e.g., pre- and post-rehabilitation) and for the development of novel balance rehabilitation techniques and assistive devices.

ACKNOWLEDGMENT

This work was supported by the Canadian Institutes of Health Research (#129179), the Toronto Rehabilitation Institute, the Canadian Paraplegic Association Canada, and the Natural Sciences and Engineering Research Council of Canada.

CONFLICT OF INTEREST STATEMENT

There are no conflicts of interest for the authors of this study.

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TABLE LEGENDS

TABLE I. Analysis of the anterior-posterior (AP), medial-lateral (ML), and resultant distance (RD) COP fluctuation. Posturographic measures were computed for quiet sitting (SI) and standing (ST) under eyes open (EO) and eyes closed (EC) conditions. Shown are the mean and standard deviation for the mean distance (MD), mean velocity (MV), centroidal frequency (CFREQ), frequency dispersion (FREQD), short-term Hurst (H_S), and long-term Hurst exponent (H_L). Analyses included the Wilcoxon test of signed ranks between tasks and eye conditions and Spearman's rho correlation between tasks (* $P<0.05$; ** $P<0.01$).

FIGURE LEGENDS

Fig. 1. Examples of COP time series during sitting and standing (one subject, EO). A: The AP (top) and ML (bottom) time series during sitting (left) and standing (right). B: The planar phase plots of the COP fluctuation during sitting (top) and standing (bottom). Note that only the initial 15 seconds of data are presented for each task in order to reveal the fluctuation behavior.

Fig. 2. Examples of log-log stabilogram diffusion functions (SDF) calculated from the COP data during sitting and standing (one subject, EO). A: Log-log SDF plots during quiet sitting (AP and ML). B: Log-log SDF plots during quiet standing (AP and ML). The dashed lines represent linear regression fits used to estimate the Hurst exponents H_S and H_L.

Fig. 3. Correlation graphs for mean velocity (MV) measures, which exhibited a linear correlation between quiet sitting (SI) and standing (ST). Top: MV correlation for the anterior-posterior (AP) COP fluctuation during eyes open condition (EO). Center: MV correlation for the medial-lateral (ML) COP fluctuation during EO. Bottom: MV correlation for the resultant distance (RD) COP fluctuation during EO.

TABLE I. Analysis of the anterior-posterior (AP), medial-lateral (ML), and resultant distance (RD) COP fluctuation. Posturographic measures were computed for quiet sitting (SI) and standing (ST) under eyes open (EO) and eyes closed (EC) conditions. Shown are the mean and standard deviation for the mean distance (MD), mean velocity (MV), centroidal frequency (CFREQ), frequency dispersion (FREQD), short-term Hurst (H_S), and long-term Hurst exponent (H_L). Analyses included the Wilcoxon test of signed ranks between tasks and eye conditions and Spearman's rho correlation between tasks (* P<0.05; ** P<0.01).

Measures		**Quiet Sitting**		**Quiet Standing**		**Ratio of SI/ST**		**Wilcoxon Test of Signed Ranks**				**Spearman's rho**	
								Task		Eye		Task	
		EO	*EC*	*EO*	*EC*	*EO*	*EC*	*EO*	*EC*	*SI*	*ST*	*EO*	*EC*
MD (mm)	*AP*	**1.66** (1.02)	**1.40** (0.81)	**2.53** (0.65)	**4.26** (1.20)	0.66	0.33	–	**	–	**	–	–
	ML	**0.93** (0.46)	**1.00** (0.59)	**1.25** (0.41)	**1.89** (0.76)	0.75	0.53	–	*	–	*	–	–
	RD	**2.06** (1.04)	**1.87** (0.95)	**3.04** (0.59)	**5.03** (1.19)	0.68	0.37	*	**	–	**	–	–
MV (mm/s)	*AP*	**3.12** (0.76)	**3.18** (0.60)	**4.21** (0.96)	**7.90** (2.56)	0.74	0.40	**	**	–	**	0.7*	–
	ML	**2.46** (1.36)	**2.37** (1.08)	**3.04** (1.75)	**3.92** (1.94)	0.81	0.60	–	**	–	**	0.7*	–
	RD	**4.43** (1.55)	**4.40** (1.15)	**5.78** (2.10)	**9.61** (3.28)	0.77	0.46	*	**	–	**	0.7*	–
CFREQ (Hz)	*AP*	**0.65** (0.27)	**0.70** (0.26)	**0.36** (0.11)	**0.39** (0.14)	1.82	1.79	**	**	–	–	–	–
	ML	**0.65** (0.12)	**0.62** (0.21)	**0.57** (0.15)	**0.52** (0.25)	1.15	1.19	–	–	–	–	–	–
	RD	**0.56** (0.21)	**0.55** (0.17)	**0.38** (0.14)	**0.40** (0.17)	1.47	1.38	*	–	–	–	–	–
FREQD (–)	*AP*	**0.86** (0.08)	**0.85** (0.07)	**0.80** (0.06)	**0.77** (0.05)	1.08	1.11	*	*	–	*	–	–
	ML	**0.86** (0.04)	**0.89** (0.04)	**0.76** (0.09)	**0.77** (0.11)	1.14	1.16	**	**	–	–	–	–
	RD	**0.90** (0.03)	**0.91** (0.02)	**0.85** (0.03)	**0.84** (0.03)	1.06	1.08	**	**	–	–	–	–
H_S (–)	*AP*	**0.68** (0.05)	**0.68** (0.05)	**0.86** (0.04)	**0.86** (0.04)	0.78	0.78	**	**	–	–	–	–
	ML	**0.68** (0.07)	**0.66** (0.07)	**0.83** (0.03)	**0.83** (0.04)	0.82	0.79	**	**	–	–	–	–
	RD	**0.66** (0.05)	**0.65** (0.05)	**0.82** (0.06)	**0.83** (0.06)	0.80	0.79	**	**	–	–	–	–
H_L (–)	*AP*	**0.23** (0.12)	**0.23** (0.14)	**0.17** (0.12)	**0.18** (0.14)	1.35	1.30	–	–	–	–	–	–
	ML	**0.25** (0.12)	**0.28** (0.12)	**0.15** (0.04)	**0.20** (0.12)	1.65	1.39	*	–	–	–	–	–
	RD	**0.18** (0.11)	**0.17** (0.09)	**0.07** (0.10)	**0.07** (0.07)	2.74	2.51	*	*	–	–	–	–

Figure_1
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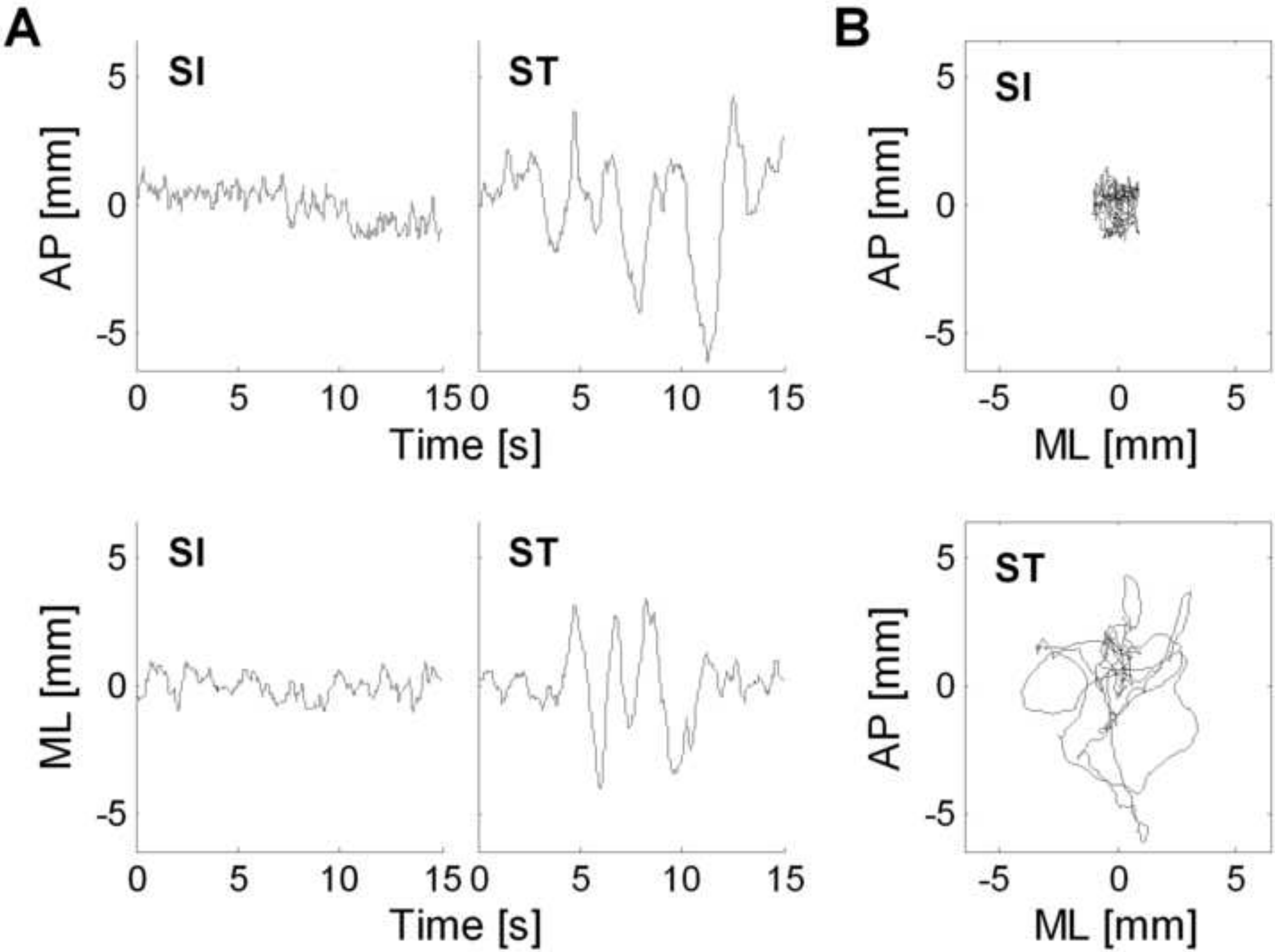

Figure_2

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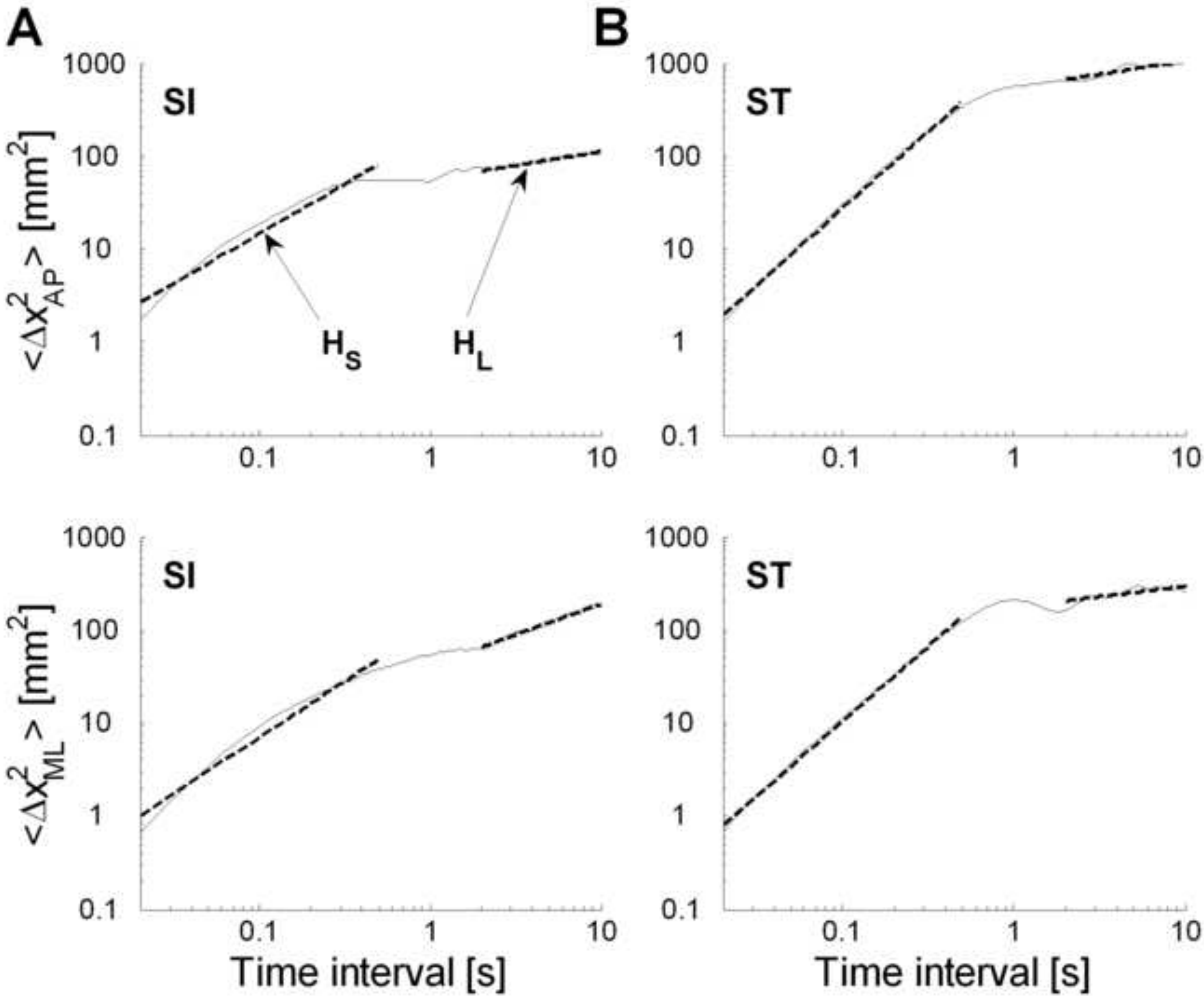

Figure_3

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