Trunk control impairment is responsible for postural instability during quiet sitting in individuals with cervical spinal cord injury

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Abstract

Background: Individuals with cervical spinal cord injury usually sustain impairments to the trunk and upper and lower limbs, resulting in compromised sitting balance. The objectives of this study were to: 1) compare postural control of individuals with cervical spinal cord injury and able-bodied individuals; and 2) investigate the effects of foot support and trunk fluctuations on postural control during sitting balance.

Methods: Ten able-bodied individuals and six individuals with cervical spinal cord injury were asked to sit quietly during two 60 s trials. The forces exerted on the seat and the foot support surfaces were measured separately using two force plates. The global center of pressure sway was obtained from the measurements on the two force plates, and the sway for each force plate was calculated individually.

Findings: Individuals with spinal cord injury had at least twice as large global and seat sways compared to able-bodied individuals, while foot support sway was not significantly different between the two groups. Comparison between global and seat sways showed that anterior-posterior velocity of global sway was larger compared to the seat sway in both groups.

Interpretation: Postural control of individuals with cervical spinal cord injury was worse than that of able-bodied individuals. The trunk swayed more in individuals with spinal cord injury, while the stabilization effect of the feet did not differ between the groups.

Foot support affected anterior-posterior fluctuations in both groups equally. Thus, trunk control is the dominant mechanism contributing to sitting balance in both able-bodied and spinal cord injury individuals.

Key words: sitting balance; postural sway; center of pressure (COP); spinal cord injury (SCI); foot support; trunk control.

1. Introduction

Individuals with spinal cord injury (SCI) often experience motor and/or sensory impairment below the level of injury. Cervical injuries lead to impairment in upper limb, trunk, and lower limb muscles, while thoracic injuries lead to impairment in trunk and lower limb muscles (Seelen et al., 1998). Individuals with cervical or thoracic injuries often have impaired sitting balance (Chen et al., 2003; Grangeon et al., 2012). Impaired sitting balance after SCI causes an individual to alter his/her sitting strategy (e.g. individuals with SCI tilt their pelvis in order to achieve greater stability during sitting) and results in compromised sitting posture. Instability during sitting may affect performance of activities of daily living such as reaching and object manipulation (Chen et al., 2003), and could result in secondary health complications such as pressure sores (Minkel, 2000). Continuous, tonic activation of trunk muscles is required to maintain upright sitting posture, and phasic, feedback-driven activations are required to respond to balance disturbances (Masani et al., 2009). Therefore, paralysis of trunk muscles is one of the main reasons for compromised sitting balance after SCI (Minkel, 2000). Individuals with SCI often use innervated, non-postural muscles (e.g. shoulder and neck muscles), to compensate for the sitting impairment by voluntarily contracting these muscles to regulate sitting balance (Seelen et al., 1998). They also use their arms to increase the base of support during sitting, which can help improve their stability (Grangeon et al., 2012). Despite the compensatory methods that an individual with SCI may use, their sitting balance still remains and suboptimal.

Center of pressure (COP) sway during quiet standing and quiet sitting has been utilized to assess postural stability during standing balance (Prieto et al., 1996; Vette et al.,

2010) and sitting balance among able-bodied individuals (Dean et al., 1999; Gilsdorf et al., 1990; Kerr and Eng, 2002; Vette et al., 2010). Such assessments are relatively easy to perform in the laboratory, as they do not threaten the stability of the participants, and as such, can be applicable to evaluate sitting balance of individuals with trunk instability resulting from SCI. A small number of studies evaluated sitting balance of individuals with SCI using COP recordings (Chen et al., 2003; Grangeon et al., 2012; Grangeon et al., 2013; Shirado et al., 2004). These studies showed that postural sway is larger among individuals with SCI compared to able-bodied individuals, indicating worse postural stability and compromised sitting balance after a SCI (Grangeon et al., 2012; Shirado et al., 2004). Interestingly, the research indicates here are no differences in postural sway between individuals with low and high thoracic SCI (Chen et al., 2003). However, in these studies the effect of foot support was either ignored (Chen et al., 2003; Shirado et al., 2004) or not analyzed in detail (Grangeon et al., 2012, 2013).

It is recognized that foot support affects sitting balance. For example, in able-bodied individuals both COP displacement and velocity increased by as much as 70% during forward reaching when the subjects were allowed to use foot support compared to reaching without foot support (Kerr and Eng, 2002). Footrests also increased trunk displacement during forward reaching among individuals with thoracic SCI (Potten et al., 2002). However, the effect of foot support on postural stability of individuals with SCI during quiet sitting has yet to be examined in the literature. In prior studies, Chen et al. (2003) and Shirado et al. (2004) used one force plate positioned under the buttocks with both feet supported on the ground, but they did not account for the effects of foot support on COP.

Postural control is the ability to maintain balance. Various factors, including foot support and trunk control contribute to postural control during sitting balance. Trunk control is the ability to control the trunk, which can be evaluated by analyzing the trunk fluctuations using COP measures obtained from a force plate placed under the buttocks. Similarly, foot support can be evaluated using postural sway fluctuations obtained using a force plate placed under the feet. Grangeon et al. (2012, 2013) used two force plates, one placed under the buttocks on the seat and the other one under the feet, to calculate the COP, but they did not analyze the separate contributions from the foot support and trunk fluctuations. This is likely because individuals with SCI typically do not have full voluntary control of lower limbs. Consequently, the utility of their foot support is often considered marginal despite the fact that it has been shown that foot support provides a significant contribution during transfers in individuals with SCI (Gagnon et al., 2008).

We hypothesized that individuals with cervical SCI will have worse sitting balance compared to able-bodied individuals, and that foot support will have a positive impact on postural stability. The objectives of this study were to: 1) compare postural control of individuals with SCI with able-bodied individuals; and 2) investigate the effects of foot support and trunk fluctuations on the postural control of the entire body during quiet sitting balance.

2. Methods

2.1. Participant recruitment

Able-bodied individuals and individuals with SCI were recruited to participate in this study. In order to participate, all participants had to have the ability to maintain unsupported sitting. Individuals were recruited in the able-bodied group if they had no history of neurological impairment or musculoskeletal injury that could affect their sitting balance. Individuals were recruited in the SCI group if they had either motor incomplete or complete, sensory incomplete or complete cervical SCI, and were minimum one year post injury. All participants gave written informed consent in accordance with the Declaration of Helsinki. The experimental procedures used in this study were approved by the local institutional research ethics board.

2.2. Experimental protocol

Participants were seated in upright sitting posture on a height-adjustable chair without back support and with their feet on the ground in all trials. The seating surface and the foot support surface were each instrumented with a force plate (AccuSway^{Plus}, Advanced Mechanical Technology Inc., Watertown, USA). A thin foam cover was placed over the seat surface to prevent risks of skin injury during data collection in both able-bodied and SCI groups. The height of the chair was adjusted such that the knee angle was at approximately 90^{o}. Each participant was asked to keep a steady sitting balance with his or her arms crossed over their chest and with their eyes open, as illustrated in Figure 1.

2.3. Measurements

Signals were recorded over two 60 s trials using two force plates (Figure 1). Seat COP: $COP_S(x,y)$ was calculated from the force plate on the seating surface. Foot support COP: $COP_F(x,y)$ was calculated from the force plate on the ground. Global COP: $COP_G(x,y)$ was then computed from $COP_S(x,y)$ and $COP_F(x,y)$ as described in the next section.

Moreover, force components measured using the seat force plate (F_{xS}, F_{yS}, F_{zS}) and the foot support force plate (F_{xF}, F_{yF}, F_{zF}) were collected. The x and y denote the respective coordinates of medial-lateral (ML) and anterior-posterior (AP) directions and z denotes the vertical direction, as shown in Figure 1. Radial distance (RD) time series was calculated as $RD = \sqrt{AP^2 + ML^2}$ to represent the combined AP and ML COP position according to Prieto et al. (1996). The AP and ML time series were used to examine the specific sway directions and the RD measure was used to examine the overall COP sway. All data were sampled at 500 Hz using a 12-bit data acquisition system (NI 6071E, National Instruments, Austin, USA). A low-pass filter with a cut-off frequency of 5 Hz was applied to all signals (Prieto et al., 1996; Vette et al., 2010).

2.4. Global COP calculation

The seat and the foot support force plate surface were aligned along the x and y axis and the seat surface was higher than the foot support surface along the z axis (i.e. distance h), as shown in Figure 1. The origin O(0,0,0) for all calculations was positioned on the seat surface in the middle of the seat and foot support force plates. The sum of all moments around the origin, M(x,0) and M(0,y), in the ML and AP directions was calculated in Equation 1:

$$\sum M(x,0) = (F_{zS} + F_{zF}) \cdot COP_G(0,y) = F_{zS} \cdot COP_S(0,y) + F_{zF} \cdot COP_F(0,y) + F_{yF} \cdot h\,,$$

$$\sum M(0,y) = (F_{zS} + F_{zF}) \cdot COP_G(x,0) = F_{zS} \cdot COP_S(x,0) + F_{zF} \cdot COP_F(x,0) + F_{xF} \cdot h\,. \qquad (1)$$

where h was the height difference between the seat and foot support plates Contributions of shear forces due to the height difference between the seat and foot support plates (i.e. $F_{yF} \cdot h$ and $F_{xF} \cdot h$) were included because they could be extensive during various sitting manoeuvres (Gilsdorf et al., 1990). Global COP for sitting was calculated by re-arranging Equation 1 to obtain $COP_G(x,y)$ as a function of measured parameters in Equation 2:

$$COP_G(x,0) = COP_S(x,0)\cdot\frac{F_{zS}}{F_{zS}+F_{zF}} + COP_F(x,0)\cdot\frac{F_{zF}}{F_{zS}+F_{zF}} + h\cdot\frac{F_{xF}}{F_{zS}+F_{zF}},$$

$$COP_G(0,y) = COP_S(0,y)\cdot\frac{F_{zS}}{F_{zS}+F_{zF}} + COP_F(0,y)\cdot\frac{F_{zF}}{F_{zS}+F_{zF}} + h\cdot\frac{F_{yF}}{F_{zS}+F_{zF}}. \quad (2)$$

In case when $h = 0$, $COP_G(x,y)$ gives a global COP for dual force plates, previously described by Winter et al., (1998).

2.5. Data analysis

The $COP_G(x,y)$, $COP_S(x,y)$ and $COP_F(x,y)$ postural sway fluctuations were analyzed separately. Each COP measure was also separately computed for the ML and AP fluctuations (i.e. x and y coordinate) as well as for the overall, RD fluctuations. Subsequent references to the global, seat and foot support COP fluctuations for RD, AP and ML directions will be made as COP_S, COP_F and COP_G, unless specifically stated. Time and frequency domain parameters that quantified all COP_G, COP_S and COP_F fluctuations were calculated and averaged over two trials to represent the value for each subject. Each COP measure was referenced by subtracting them from their mean (Preito et al., 1996). Time-domain measures included: a) the mean distance (MD) which represented the average distance traveled by the COP; b) the mean velocity (MV) which was the average velocity of the COP time series; c) the 95% confidence ellipse (AREA-CE) which estimated the elliptical area of best fit enclosed by 95% of COP points; and d) the mean frequency (MFREQ) which was a hybrid measure that represented the rotational frequency of the COP series by the ratio of the mean velocity to the mean distance in revolutions per second (Hz). Stability performance parameters (MD and AREA-CE) were used to evaluate postural stability, whereas the control demand parameter (MV) was used to evaluate the amount of postural activity that the system needed in order to achieve stability (Grangeon et al., 2013). The hybrid measure (MFREQ) described the relationship between the control demand and stability performance (Prieto et al., 1996).

The frequency-domain parameters characterized the area or shape of the power spectral density of the COP series, which was calculated in the frequency range from 0.15 to 5.0 Hz, and assessed postural stability regulation (Grangeon et al., 2013; Prieto et al., 1996). They included: a) centroidal frequency (CFREQ), which represented the central mass frequency; and b) the frequency dispersion (FREQD), which was a unit-less measure of the variability of the power spectral density. It has been reported that the natural frequency of an inverted pendulum is inversely proportional to the moment of inertia (Winter et al., 1998) and that CFREQ is also linked to the inertia of the trunk during sitting balance (Grangeon et al., 2013; Vette et al., 2010). Also, using computational simulations it was found that FREQD was correlated to the stiffness of the inverted pendulum, which was used to model the trunk (Maurer and Peterka, 2005).

2.6. Statistical analysis

Comparisons were performed using Mann-Whitney test for independent samples comparison and Wilcoxon signed-rank test for related samples comparison. Non-parametric tests were chosen over the equivalent parametric tests because the Shapiro-

Wilk test suggested that all identified measures were not normally distributed. Statistical analysis was performed with a significance level $P < 0.05$.

3. Results

3.1. Participants

Ten able-bodied individuals and six individuals with cervical SCI participated in this study. Participants' demographic information is summarized in Table 1. The mean age ($P = 0.428$), mean weight ($P = 0.792$), and mean height ($P = 0.445$) of participants in able-bodied and SCI groups were not significantly different, suggesting that the groups were matched.

3.2. Differences in able-bodied and SCI groups

Figure 2 illustrates a representative sample of 15 s of COP_G, COP_S, COP_F data during sitting balance for an able-bodied individual and an individual with SCI. The planar representation of AP and ML sway is shown on the left hand side of the figure and the separate AP and ML time series are shown on the right hand side of the figure. The plots illustrate that amount of postural sway. The sway area was considerably larger in an individual with SCI compared to the able-bodied individual.

Comparison of COP_G postural sway parameters between able-bodied and SCI groups, in the first two columns of Table 2, revealed that overall RD sway amount (MD and AREA-CE) was considerably larger in the SCI group for all measures. Mean frequency (MFREQ) of COP_G was smaller in SCI participants for AP and ML directions as well as the overall RD measure. Analysis of COP_S postural sway parameters between able-bodied and SCI groups, in the third and fourth column of Table 2, revealed that amount of sway (MD and AREA-CE) was significantly larger in the SCI group for ML direction, but not for AP direction, though the means were considerably different. Mean velocity (MV) of COP_S sway was larger in the SCI group only for the AP direction. Similarly, mean frequency (MFREQ) of SCI participants was smaller for the overall RD sway and for ML direction, but it was not different for AP direction. In the last two columns of Table 2, comparison of COP_F sway parameters did not show any significant differences between able-bodied and SCI groups.

3.3. Effect of foot support

Effects of foot support forces on the COP were analyzed by comparing parameters obtained for COP_G to those obtained for COP_S (i.e. COP_G vs. COP_S) in the able-bodied and SCI groups separately. Comparison between the first and third columns of Table 2 for the able-bodied group, and second and fourth columns of Table 2 for the SCI group, revealed that there were no significant differences in amount of postural sway (MD and AREA-CE) between COP_G and COP_S in either able-bodied or SCI groups. Mean velocity (MV) of COP_G sway, when compared to COP_S, was larger among the able-bodied and SCI group for the overall RD sway and for the AP direction. The sway frequency (CFREQ and MFREQ) was also larger for COP_G, when compared to COP_S, for all measures in able-bodied group and for the overall RD sway and AP direction among individuals with SCI.

4. Discussion

4.1. Sitting balance in individuals with SCI

Global postural sway (i.e. COP_G) parameters were used to assess the whole-body sitting balance. Results showed that individuals with SCI had much larger overall postural sway, and their mean frequency of fluctuations was smaller, compared to able-bodied individuals. There were no significant differences in velocity of sway between the two groups for the overall postural fluctuations, i.e. COP_G (Table 2). It has been suggested that the amount of COP sway (i.e. mean distance) is associated with postural stability performance, whereas COP velocity was associated with how much activity the postural control system needed to achieve stability (Grangeon et al., 2013; Maki et al., 1990; Prieto et al., 1996). Using this logic, our results suggest that individuals with cervical SCI had less postural stability (suggested by the sway mean distance) with the same amount of postural activity (suggested by the sway velocity results), for both the anterior-posterior and medial-lateral directions. They indicate that individuals with SCI were less stable despite having similar amount of postural control activity. These findings are consistent with results published by Grangeon et al. (2012), who reported that both the anterior-posterior and medial-lateral stability performance during unsupported sitting was lower among individuals with SCI compared to able-bodied controls.

Inability to voluntarily control postural trunk and lower limb muscles (Minkel, 2000) and reliance on higher, non-postural muscles such as shoulder and neck muscles for balance (Seelen et al., 1998) would most likely explain decreased postural performance observed in individuals with cervical SCI. Grangeon et al. (2012) reported similar findings even though their participant group was not homogenous and included some individuals with lower injuries (thoracic and high-lumbar injuries, in addition to cervical injuries). Our study examined only individuals with mid-cervical SCI (C4 to C6 level of injury) who generally exhibit higher levels of trunk muscle paralysis and as a consequence have limited ability, if any, to actively regulate sitting balance (Minkel, 2000). As expected, postural stability of SCI participants in our study was lower than that of the participants in Grangeon et al. (2012), when comparing group averages between the two studies (e.g. overall mean distance was 35% larger, and mean velocity 39% smaller among SCI participants in our study compared to Grangeon et al. (2012) study). These results suggest that global COP measures of postural stability can dependably evaluate sitting balance impairment after SCI. Moreover, based on these results, it is evident that individuals with cervical SCI have considerably compromised overall global (i.e. COP_G) postural control, which is calculated from the seat and foot support force plates, and indicates the performance of both the trunk control and foot support during sitting balance. In the next two sections the trunk control, which was analyzed using measurements on the seat, and the foot support, which was analyzed using measurements of the foot support, are discussed in more details.

4.2. Postural stability of the trunk

Postural sway parameters measured on the seat force plate (i.e. COP_S) showed that individuals with SCI, compared to able-bodied individuals, had considerably larger postural sway, smaller mean frequency, and larger anterior-posterior sway velocity (Table 2). These findings imply that individuals with SCI have lower postural stability of the trunk compared to able-bodied individuals. The observed trunk instability among individuals with SCI was prevalent in the medial-lateral direction. The observed medial-lateral instability of the trunk is consistent with Shirado et al. (2004) who also showed that individuals with complete thoracic SCI had greater medial-lateral COP postural movements of the trunk, compared to able-bodied individuals. This medial-lateral instability in individuals with SCI may be explained by the passive structures of the trunk and lower limbs which provide a larger base of support for the anterior-posterior direction and the frequently observed kyphotic posture in individuals with SCI which increases anterior-posterior support and may cause medial-lateral instability during sitting (Minkel, 2000). Results also showed that individuals with SCI required more postural activity (i.e. larger mean velocity) to stabilize anterior-posterior fluctuations recorded on the seat (i.e. COP_S). However, foot support may have influenced anterior-posterior fluctuations. Overall, our results for the seat fluctuations suggest that trunk control is considerably compromised in individuals with cervical SCI during sitting balance.

4.3. Effects of foot support on postural stability

Comparison of global and seat COP sway parameters (i.e. COP_G vs. COP_S) was used to examine the effects of foot support on postural stability in able-bodied and SCI groups separately, since foot support postural sway (COP_F) did not differ between the able-bodied and SCI groups. Results showed that anterior-posterior postural sway mean velocity with foot support (i.e. COP_G), compared to sway without the effect of foot support (i.e. COP_S) was larger in both the able-bodied group and SCI group. These findings revealed that foot support increased postural activity of the body by adding faster fluctuations in anterior-posterior direction. Grangeon et al. (2013) investigated the effects of upper limb support during sitting balance and showed that COP sway velocity increased when hands were placed on the thighs for support, compared to when hands were not rested on the thighs.

Results also showed that centroidal frequency as well as the mean frequency of postural sway with foot support (i.e. COP_G), compared to sway without foot support (i.e. COP_S), was larger in both able-bodied and SCI groups (COP_G was larger than COP_S for both able-bodied and SCI groups as shown in Table 2), which is consistent to what Grangeon et al. (2012) showed during upper limb supported sitting. Larger centroidal frequency of a swaying object is associated with smaller effective moment of inertia of the trunk (Grangeon et al., 2013; Vette et al., 2010). It has also been reported that the natural frequency of the swaying object and its inertia are inversely proportional (Winter et al., 1998). Systems with larger moment of inertial are generally more sluggish and require more time to return to equilibrium, which suggests that they are less stable (Vette et al., 2010). Similarly, it follows that that systems with smaller moment of inertial are less sluggish, require less time to return to equilibrium and are consequently more stable. Our findings suggest that effective moment of inertial during postural stability in both able-bodied and SCI individuals was smaller (i.e. larger centroidal frequency), and imply that both groups had more stable anterior-posterior postural stability due to the effects of foot support. Overall, foot support provided stability predominantly in anterior-posterior direction, which is undoubtedly because the passive mechanics of the foot support are aligned with anterior-posterior direction.

Previous research has shown that able-bodied individuals do not need to actively contract lower limb muscles during quiet sitting as long as they remain within the base of support (Dean et al., 1999). Similarly, individuals with thoracic SCI do not actively contract their hip and lower limb muscles during reaching tasks (Potten et al., 2002). Since both able-bodied and SCI individuals in our study used foot support in the same way during postural stability, it appears that foot support provided passive stability during quiet sitting in both groups. This is consistent with Gagnon et al. (2008) findings, which showed that individuals with SCI used passive foot support during transfers. Taken together, it seems that trunk control is the predominant mechanism contributing to differences in postural stability during sitting in able-bodied and SCI individuals. Foot support provides passive support and only passively influences how postural stability is attained during sitting (Potten et al., 2002).

4.4. Limitations and future work

Numerous factors such as the neurological injury level and sensory and motor completeness of injury can affect sitting balance performance after SCI (Grangeon et al., 2012; Seelen et al., 1998). It is possible that age, motor and sensory impairment, and time since injury played a role on the sitting balance of individuals with cervical SCI in our study. Therefore, further studies are warranted to systematically examine their effects on sitting balance after SCI. Moreover, future studies should consider full kinematic analysis of the upper body to understand the multi-segmented trunk control during sitting balance.

5. Conclusions

The results of our study indicate that individuals with cervical SCI have significantly compromised sitting postural stability as they swayed at least twice more than able-bodied individuals. The postural sway of trunk was larger in the SCI individuals than able-bodied individuals, while the use of foot support for postural stability improved the stability but the improvement in stability was not different between the groups. Thus, we conclude that trunk control is the dominant mechanism contributing to differences in sitting postural stability between able-bodied individuals and individuals with SCI, and that the overall instability in individuals with cervical SCI is due to postural instability of the trunk. These results emphasize the importance of trunk control in sitting balance and indicate the importance of recovering trunk function in rehabilitation of individuals with SCI as a way to improve their sitting balance required for functional activities.

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Conflict of interest

The authors declare that there are no conflicts of interests.

Tables

Table 1: Participants' demographic information, where AB represents able-bodied individuals and SCI represents SCI individuals.

Group		Gender	Age (years)	Height (cm)	Weight (kg)	Level of injury	Motor Impairment	Sensory Impairment	Time since injury (years)
AB	1	F	27	165.1	54.4				
	2	M	27	180.0	76.0				
	3	M	25	188.0	88.5				
	4	M	38	180.0	74.0				
	5	F	29	162.6	53.1				
	6	M	41	182.9	83.0				
	7	F	38	165.1	61.2				
	8	F	31	162.6	49.9				
	9	M	29	177.8	70.3				
	10	M	25	180.0	75.0				
Mean			***31.0***	***174.4***	***68.5***				
SD			***5.9***	***9.5***	***13.2***				
SCI	1	M	25	180.3	61.2	C5	Incomplete	Incomplete	3
	2	M	25	170.2	72.0	C5	Complete	Incomplete	3
	3	M	33	175.3	79.4	C4-6	Complete	Incomplete	16
	4	M	71	n/a	86.8	C4-6	Incomplete	Incomplete	6
	5	M	40	n/a	67.8	C6	Complete	Complete	19
	6	M	54	n/a	83.1	C5	Incomplete	Complete	27
Mean			***41.3***	***175.3***	***75.1***				
SD			***18.1***	***5.1***	***9.8***				

Table 2: Analysis of parameters for the global (COP_G), seat (COP_S) and foot support (COP_F) postural sway during quiet sitting were performed. AB represents able-bodied group and SCI represents individuals with SCI. Postural stability parameters for the radial distance (RD) measure and the anterior-posterior (AP) and medial-lateral (ML) direction were computed. Shown are the group mean (SD) for the mean distance (MD), mean velocity (MV), 95% confidence ellipse (AREA-CE), mean frequency (MFREQ), centroidal frequency (CFREQ) and frequency dispersion (FREQD). Analysis included the Mann-Whitney test to compare AB and SCI groups for each parameter (light grey) and Wilcoxon signed-rank test to compare COP_G and COP_S parameters in each group (dark grey).

Measure		**Global: COP_G**		**Seat: COP_S**		**Foot Support: COP_F**		**Statistics**				
								AB vs. SCI			**COP_G vs. COP_S**	
		AB	**SCI**	**AB**	**SCI**	**AB**	**SCI**	**COP_G**	**COP_S**	**COP_F**	**AB**	**SCI**
MD (cm)	RD	0.13(0.04)	0.40(0.39)	0.13(0.06)	0.48(0.43)	0.16(0.10)	0.20(0.19)	*	*			
	AP	0.10(0.04)	0.34(0.34)	0.11(0.05)	0.42(0.39)	0.10(0.06)	0.17(0.19)	*				
	ML	0.06(0.03)	0.15(0.14)	0.06(0.02)	0.17(0.16)	0.10(0.07)	0.07(0.04)	*	**			
MV (cm/s)	RD	0.46(0.18)	0.48(0.13)	0.31(0.10)	0.41(0.12)	0.52(0.24)	0.39(0.14)				**	*
	AP	0.35(0.15)	0.36(0.07)	0.17(0.05)	0.26(0.06)	0.30(0.09)	0.25(0.09)		*		**	*
	ML	0.22(0.10)	0.25(0.10)	0.22(0.09)	0.27(0.11)	0.35(0.26)	0.24(0.11)					*
AREA-CE (cm^2)	-	0.18(0.11)	3.14(5.54)	0.23(0.20)	3.57(5.81)	0.37(0.44)	0.62(0.68)	*	*			
MFREQ (Hz)	RD	0.66(0.35)	0.32(0.15)	0.42(0.15)	0.23(0.14)	0.68(0.33)	0.70(0.66)	*	*		**	*
	AP	0.77(0.47)	0.38(0.27)	0.37(0.17)	0.26(0.27)	0.77(0.44)	0.81(0.87)	*			**	*
	ML	0.77(0.28)	0.41(0.13)	0.65(0.20)	0.40(0.15)	0.84(0.43)	0.95(0.61)	**	*		*	
CFREQ (Hz)	RD	1.53(0.35)	1.27(0.30)	1.30(0.30)	1.03(0.25)	1.53(0.36)	1.48(0.60)				**	*
	AP	1.42(0.33)	1.27(0.36)	1.17(0.29)	1.05(0.34)	1.51(0.41)	1.51(0.60)				**	*
	ML	1.40(0.20)	1.10(0.33)	1.27(0.23)	1.05(0.31)	1.48(0.32)	1.55(0.38)				*	
FREQD (-)	RD	0.60(0.03)	0.58(0.05)	0.59(0.04)	0.57(0.06)	0.62(0.02)	0.60(0.04)					
	AP	0.60(0.04)	0.57(0.04)	0.60(0.04)	0.57(0.04)	0.60(0.05)	0.59(0.04)					
	ML	0.58(0.05)	0.59(0.04)	0.57(0.05)	0.58(0.04)	0.61(0.03)	0.59(0.04)				*	

* $P < 0.05$; ** $P < 0.01$

Figures Legend

Figure 1: Experimental setup for sitting balance utilizing two force plates. $COP_S(x,y)$ captured trunk sway on the seat surface and $COP_F(x,y)$ captured foot support sway on the ground. Vertical forces (F_{zS} and F_{zS}) and AP and ML forces (not shown) were also captured. The origin, O(0,0,0), of the global coordinate frame, which was used to calculate global COP was placed the middle of the force plates, between the seat and foot support surface on the seat surface, where the seat and foot support surface were aligned along the x and y axis and only separated by distance h which was the height difference between the seat and foot support surface along the z axis.

Figure 2: Example of the COP sway for global (COP_G), seat (COP_S) and foot support (COP_F) fluctuations during quiet sitting for: A) one able-bodied individual (AB); and B) one individual with SCI. AP represents anterior-posterior and ML medial-lateral sway direction. The planar representations (left) show spatial fluctuations of the combined AP and ML sway with respect to the origin of the global coordinate frame. Time series plots (right) show the corresponding AP and ML postural sway time series separately for COP_G, COP_S and COP_F. Note that only a representative 15 s of data is shown to reflect the postural sway behaviour.

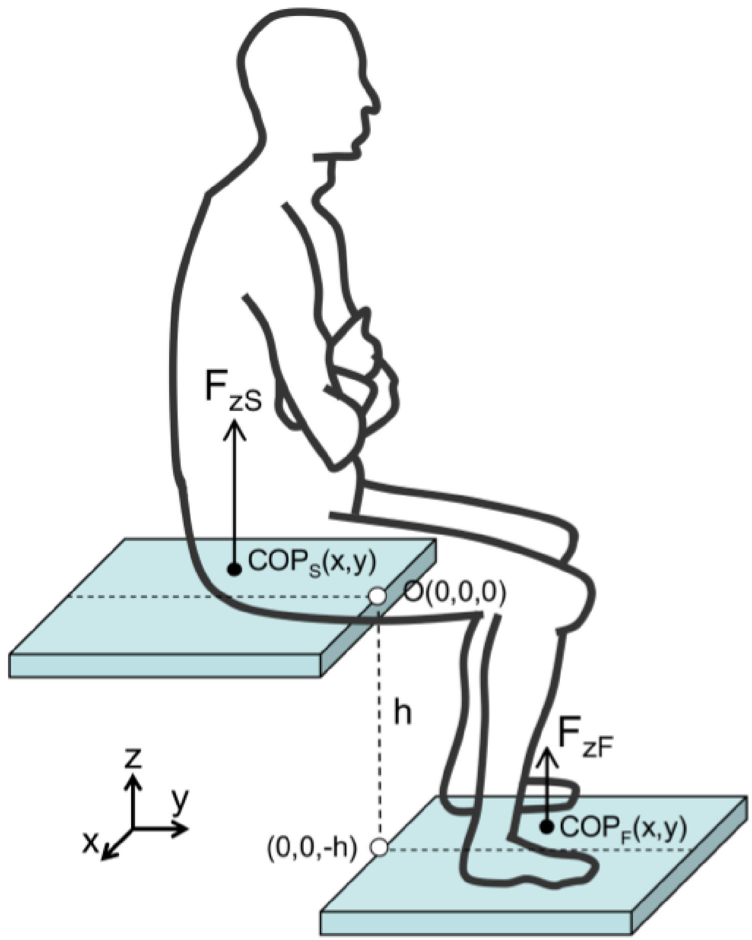

Figure 1

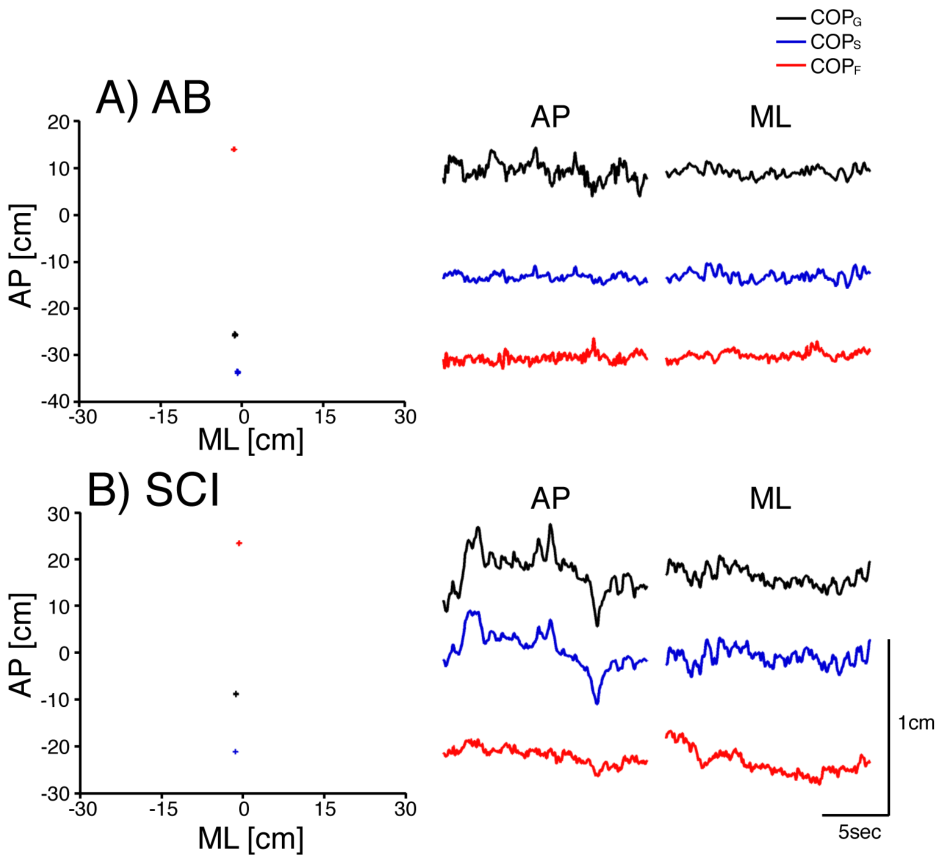
COP_G
COP_S
COP_F
A) AB
AP
ML
AP [cm]
ML [cm]
B) SCI
AP
ML
AP [cm]
ML [cm]
1cm
5sec

Figure 2