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## Quantitative analysis of static sitting posture in chronic stroke

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### ABSTRACT

Unsupported sitting requires postural stability of the trunk which is also necessary for almost all activities in daily living, yet there is a lack of research dealing with the persistence of trunk impairment post-stroke using quantitative methodologies. Therefore, the purpose of this study was to investigate unsupported sitting in individuals with chronic stroke by analyzing center of pressure (COP) signals from a force platform. Ten healthy control subjects and ten chronic stroke subjects sat on a chair without a footrest that was placed on top of a force platform. Trials consisted of eyes closed, staring at a target, and COP feedback. COP signals were analyzed using spatial and temporal techniques. Compared to controls, stroke group had larger sway area and larger displacements in all conditions ( $p < 0.05$ ) and less sample entropy ( $p < 0.05$ ) in eyes closed and target conditions. In feedback conditions, both groups had decreased sway area and maximum displacements along with stroke group having increased sample entropy ( $p < 0.05$ ). Our data suggest that trunk control, necessary for unsupported sitting, is impaired well into the chronic stage of stroke onset. Further investigations of sitting should be conducted for better understanding balance deficits under conditions localized to the trunk musculature.

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### 1. Introduction

In the early stages of stroke onset, the level of dysfunction in one's ability to maintain upright during unsupported sitting is one of the highest predictors of functional outcome [1,2]. This may be due to the fact that unsupported sitting requires postural stability of the trunk which is also important for rising from a seated position, standing, walking, bending over, leaning, reaching and resisting perturbation. Thus, poor postural stability of the trunk in individuals post-stroke disrupts recovery of overall functional performance [1–3] and needs to be better understood [1,2,4–6].

It was originally thought that impairment of the trunk observed after unilateral stroke may have a faster time-course of recovery compared to that of the extremities [7,8]. While the extremities are predominantly controlled by the contralateral hemisphere [9], there is evidence suggesting that the motoneuron pools of the trunk musculature are bilaterally driven from the brainstem as well as each hemisphere [9,10]. Because of these multiple projections that contribute to postural maintenance of the trunk,

it could be assumed that the non-lesioned hemisphere and the sub-cortical regions might provide adequate postural control post-stroke [10]. Yet, clinical measurements using the Trunk Impairment Scale show that the trunk does not provide proper postural control post-stroke and that the time-course of recovery during the six months post-stroke may not be different from that of the extremities [3,6,11]. Thus, it is necessary to study postural stability of the trunk during unsupported sitting using objective and quantitative tools to improve our understanding of this dysfunction [2,4,5].

Force platforms have been used to understand balance control in post-stroke individuals while standing [2,12–15] and recently, during unsupported sitting [4,5] by analyzing the trajectory and oscillations of the center of pressure (COP) location. More specifically, when a sample group has significantly larger spatial components (sway area, maximum displacements or average velocity) compared to another sample group, the former is thought to be less stable [4,16]. Temporal outcomes such as sample entropy provide information on the amount of randomness or regularity in the COP signal and are thought to be related to the neurophysiological integrity of the postural control system [13,17,18]. The more randomness, or higher sample entropy, is associated with the automaticity and efficiency that is typically seen in healthy postural control systems [19]. More regularity, or lower sample entropy, is associated with less automaticity and suggests that

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**Table 1**  
Demographic information of stroke and control group.

	Stroke (n=10)	Control (n=10)	Comparison
Gender	8m/2f	7m/3f	$p > 0.05$
Age (year)	56.7 ± 8.4	53.5 ± 7.0	$p > 0.05$
Mass (kg)	80.2 ± 9.5	72.1 ± 12.6	$p > 0.05$
Height (cm)	176.9 ± 21.0	172.1 ± 9.3	$p > 0.05$
Trunk impairment scale	14.2/23 ± 2.8		
Years since stroke	12.7 ± 10.6		

there may be excessive attention invested towards the task by the subject due to a malfunctioning postural control strategy [13,20].

In order to better understand how postural control of the trunk is affected in chronic stroke subjects compared to individuals who have not suffered a stroke, we tested our hypothesis that the stroke subjects would exhibit postural instability during sitting by comparing spatial and temporal outcomes on the COP trajectory.

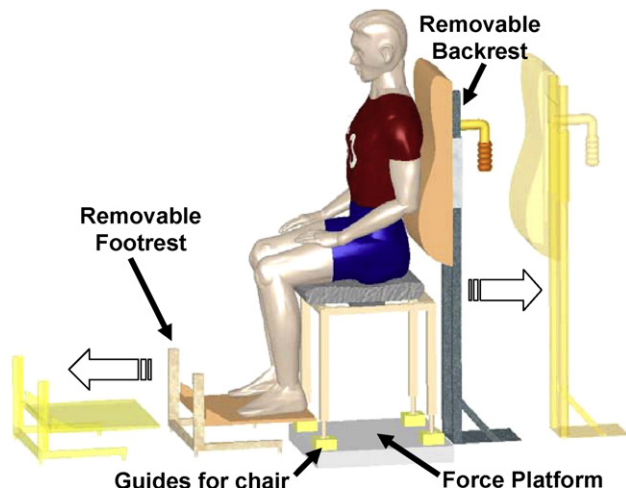
## 2. Methods and materials

Ten individuals with chronic unilateral stroke (seven left and three right hemisphere) and ten healthy controls with no history of neurological disorder participated in this study. The study protocol was approved by the Institutional Review Board (IRB) and all subjects gave informed consent to participate in the study. Stroke subjects were recruited from a non-public registry that is operated and maintained by licensed therapists. Inclusion criteria for stroke group were: first-time supratentorial stroke, moderate trunk impairment (TIS scores between nine and 17), and independent community ambulation. Exclusion criteria for stroke group were: less than a year from stroke onset, history of lower back pain or surgery, presence of hemi-neglect, bilateral stroke, visual deficit, or comprehension impairment. Inclusion and exclusion criteria for stroke subjects were enforced by screenings from a licensed therapist. Demographic descriptions of the participants are given in Table 1.

**Experimental chair (Fig. 1):** This chair had no armrest and was able to be adjusted in seat height and seat depth. The subject sat on a 45.7 cm × 45.7 cm × 7.6 cm viscoelastic cushion. A portable and adjustable backrest, along with a footrest, was used to provide two minutes of support between test trials.

**Force platforms (Fig. 1):** A force platform (1000 lb capacity, Advanced Mechanical Technology Inc., MA, USA) was placed under the four legs of the chair. Force components and the moments around these axes were recorded to calculate the COP location. In order to prevent that the chair slid or moved during or between experiments, wood guides were fixed to the force platform that allowed the contact points of the chair legs to slide through and stay at consistent locations.

The subject was asked to sit relaxed on the chair and to place hands on top of their thighs during trials in order to prevent grasping the front of the knee or seat as a support mechanism. Ten to fifteen seconds after giving these instructions, data acquisition began for duration of 60 s. Three conditions of quiet sitting were tested: Eyes Closed (EC), Target (T) and Feedback (FB). In EC trials, subjects sat still with their eyes closed for the duration of the whole trial. In T trials, subjects were asked to focus on an 11 cm circular target that was located at eye-level and 150 cm away



**Fig. 1.** Experimental setup. Subject sat on a chair that was placed through guides on top of a force platform. Also shown are the removable footrest and backrest that were used for support between trials and removed during testing.

in front of them. In FB trials, subjects attempted to keep a real-time COP indicator (6 mm × 6 mm) inside an outlined zone (50 mm × 50 mm) that was located at eye-level on a flat panel computer monitor that was 150 cm away in front of them. The size of the zone on the monitor was equivalent to a 5 mm × 5 mm area on the force platform surface. Leaning the trunk forward, backward, left or right would result in the indicator moving up, down, left or right, respectively, on the screen. Randomization was applied to EC and T trials only. FB trials were always performed last. Each subject performed three trials of each test condition.

### 2.1. Signal processing

Analog signals were low-pass filtered at a cut-off frequency of 40 Hz and then sampled at a rate of 120 Hz. In the post-processing stage, data was low-pass filtered (FIR) with a cut-off frequency of 5 Hz in order to remove noise.

### 2.2. Study outcomes

**Sway area (mm<sup>2</sup>):** Sway area was the total area covered by all the recorded COP locations within a trial. In a polar coordinate system, 2-dimensional (2D) COP locations were sorted based on their polar angles. The arc between each neighboring pair of the COP locations, and the radial distances of these two points formed a pie-shape area which could be approximated using the triangle formed by the two COP locations and the origin of the system. From 0 to 2 $\pi$  rad of the polar angle, the summation of all the triangles represented the total area that the recorded COP locations covered.

**Sway velocity (mm/s):** Sway velocity was defined as the average velocity of COP and was calculated as the total COP travel length over the time of the trial.

**Maximum displacement (mm):** The maximum COP displacement in both the coronal and sagittal planes were obtained for each trial.

**Sample entropy (SEn):** The COP location data for each of the mediolateral and anteroposterior directions was first normalized to the unit variance before SEn calculation was performed. SEn was computed using a maximum epochs length of three and a tolerance as 10% of standard deviation [13,21] and were obtained in both coronal and sagittal planes. A value close to zero indicates higher regularity while a value closer to one indicates higher entropy in the signal being analyzed.

SEn was calculated using the "sampern" algorithm from [www.physionet.com](http://www.physionet.com) [21]. All other calculations were performed using custom codes on Matlab platform (Matlab R2008a, The MathWorks, Inc., Natick, MA)

### 2.3. Statistical analysis

For continuous data regarding demographics of the participants, i.e. age, mass and height, average values were obtained for each group and a two-sample *t*-test was performed to detect significant difference between the groups. For gender distribution, a Chi-Square test was performed to detect the significant group difference.

All outcomes were first tested for normality using the Shapiro–Wilk method. If data was found to be not normally distributed, a logarithm transform was performed and data was re-tested for normality. Next, for all outcomes, a mean of the three trials from each condition (EC, T and FB) was obtained for each subject and then a group mean was calculated for each group for each condition. Therefore, group means and corresponding standard errors are reported.

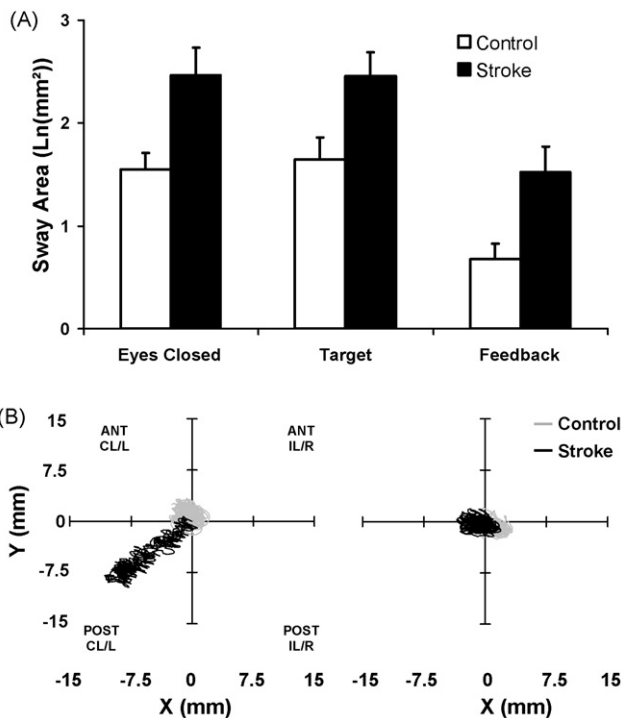
A 2 (Group: Control and Stroke) × 3 (Condition: EC, T, and FB) mix model analysis of variance (ANOVA) with repeated measures on the second factor was performed on Sway Area and Sway Velocity to detect significant differences between groups and among conditions. A 2 (Group) × 3 (Condition) × 2 (Plane: Coronal and Sagittal) mix model ANOVA with repeated measures on the second and third factors was performed for SEn and Maximum Displacement to detect significant effects of Group, Condition, and Plane. If there was a Condition effect, or any interactions, Tukey's HSD Post Hoc test was performed to obtain individual *p*-values.

All statistical analyses were performed using SPSS (SPSS version 17.0, SPSS Inc., Chicago, IL) with a significance level of 0.05.

## 3. Results

**Sway area (Fig. 2):** A significant Group effect ( $F(1,18)$ ,  $p < 0.005$ ,  $\eta_p^2 = .38$ ) was detected in that the average sway area (Fig. 2A) was found to be significantly larger in stroke group than that for the controls. A significant Condition effect ( $F(2,36)$ ,  $p < 0.001$ ,  $\eta_p^2 = .63$ ) was also detected in that subjects from both groups swayed significantly less during FB trials when compared to EC and T conditions ( $p < 0.05$ ). Representative sway data is shown in Fig. 2B for a control and a stroke subject during an EC and a FB trial.

**Maximum coronal and sagittal displacement (Fig. 3):** A significant Group effect ( $F(1,18)$ ,  $p < 0.005$ ,  $\eta_p^2 = .434$ ) was found revealing that stroke subjects had significantly larger coronal and sagittal displacements in all conditions. Along with this, there was a

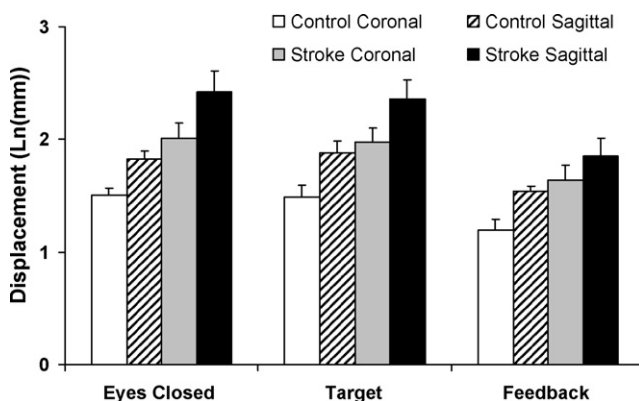


**Fig. 2.** Sway area results (mean  $\pm$  standard error, Ln(mm<sup>2</sup>)) and representative sway data. (A) Average sway area is shown for eyes closed, target and feedback conditions for both control and stroke subjects. (B) Representative sway x-y locations (mm) for a control and a stroke subject during eyes closed (left) and feedback (right) trial conditions. Also shown are coordinate systems placed at sway starting location that display the anterior (ANT) and posterior (POST) sides as well as left (L) and right (R) for control or ipsilateral (IL) and contralateral (CL) to lesion sides for stroke subject.

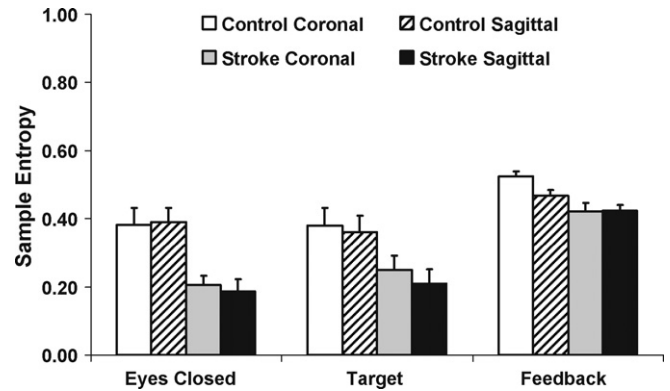
significant Plane effect ( $F(1,18)$ ,  $p < 0.001$ ,  $\eta_p^2 = .64$ ) showing that both groups had smaller coronal displacement when compared to their respective sagittal displacements. There was also a Condition effect ( $F(2,36)$ ,  $p < 0.001$ ,  $\eta_p^2 = .45$ ) showing that both groups had significantly smaller coronal and sagittal displacements in FB trials when compared to EC and T conditions ( $p < 0.05$ ).

**Sway velocity:** No significant Group ( $F(1,18)$ ,  $p > 0.05$ ,  $\eta_p^2 = .003$ ) or Condition ( $F(2,36)$ ,  $p > 0.05$ ,  $\eta_p^2 = .08$ ) effects were found.

**SEn (Fig. 4):** A significant Group x Condition interaction ( $F(2,36)$ ,  $p < 0.05$ ,  $\eta_p^2 = .16$ ) was found revealing that stroke survivors had significantly lower SEn values in EC and T but not FB conditions when compared to control group ( $p < 0.05$ ). Furthermore, the stroke group had significantly lower SEn during FB when compared



**Fig. 3.** Maximum displacements (mean  $\pm$  standard error, Ln(mm)). Results are shown for both coronal and sagittal planes for control and stroke subjects.



**Fig. 4.** SEn (mean  $\pm$  standard error). Results are shown for both coronal and sagittal planes for control and stroke subjects.

to EC and T conditions ( $p < 0.05$ ) while the control group had no differences between conditions ( $p > 0.05$ )

#### 4. Discussion

In summary, we aimed to use quantitative methodology to increase our understanding of how the ability to maintain upright during unsupported sitting is disturbed in chronic subjects of a unilateral stroke. To do this we used both spatial and temporal analyses to compare their COP trajectories to healthy controls during sitting.

Spatial analyses of the COP revealed that the chronic stroke individuals have less stable posture during sitting as shown by the significantly larger sway area than that from healthy controls. Few studies currently exist that examine sitting balance using force platforms [4,5]. A study by Genthon et al. looked at sitting balance of acute stroke subjects [4] sitting directly on a force platform without a footrest. Based on their results of significantly larger displacements and larger velocity for COP when compared to controls, they concluded that there is a postural disturbance in acute stroke subjects during sitting. Our results provide evidence for extension of this disturbance well into the chronic stage of stroke subjects.

Similar to Genthon et al., we did see larger sway area, but we did not find a significant increase of sway velocity in chronic stroke subjects when compared to controls. This was also seen by van Nes et al. [5] who only found an increase of velocity in chronic stroke subjects during unstable seating, but not when their subjects sat on a stable surface.

Similar to the study by van Nes, we did not see a difference in postural control whether subjects had their eyes open or closed while sitting on a stable surface. This was different than what is typically seen in standing wherein an absence of visual input causes significant disturbance to postural stability of stroke subjects [22]. This may be because the postural model for trunk control during sitting is simplified due to a lower center of gravity and decreased number of joints to control [4,23].

Similar to Genthon et al., who reported greater postural disturbance to be in the sagittal plane, the stroke subjects in our study had significantly larger maximum sagittal displacements [4]. While van Nes et al. reported greater postural disturbance in the coronal plane, it is possible that this difference is due to the fact that the subjects in their study sat on an unstable surface wherein the subjects in our study and Genthon et al.'s did not [5].

We must acknowledge that the experimental setups are different from each other in that the study by Genthon et al. did not use a footrest while that of van Nes et al. [4,5] did. For the current study, we chose not to use a footrest for two main reasons.

First, we wanted to isolate postural control to that of the trunk musculature and second, we wanted to remove lower limb contribution. The latter is necessary since any interaction with the footrest would affect the accuracy of the analyses because loads applied to the force platform would alter the COP trajectory and make it not completely related to trunk sway.

While the use of SEN as a method of quantifying the temporal characteristics of the COP signal is relatively new during standing [13,14,19], the current study is the first to use it for studying the sitting position. Similar to the lower SEN results in our seated stroke subjects, standing stroke subjects in a study by Roerdink et al. exhibited much lower SEN in the COP signal as well [14]. This lower SEN has been thought to be associated with excessive attention being placed on a typically automated task that subsequently interrupts and decreases the “automaticity” that is typically seen in a non-impaired nervous system [13,20]. Thus, this lower SEN suggested an improper and inefficient postural control strategy during sitting as well.

One thing that has not been reported before is the effect of visual COP feedback on sitting postural control. Previous studies have found that visual COP feedback in the standing position [24–26] significantly decreased body sway. Our results in sitting during feedback conditions also showed a decrease in trunk sway and maximum displacements in both groups, as well as a significant increase of SEN only for the stroke group. This suggests that the stroke group were able to improve postural efficiency to a level similar to controls, when externally provided with an accurate indicator of trunk position in real-time. This may indicate that the stability problem in sitting post-stroke may be related to sensory organization of postural trunk control as seen in standing balance [22,27–30] but cannot be made certain at this point since our study only attenuated vision. Future studies that test the effects of tactile and vestibular attenuation would provide a more direct conclusion on whether or not various sensory components are in fact re-weighted post-stroke.

A major limitation that should be noted in this study is young age of our stroke group along with the wide range of time from stroke onset across subjects. This may be due to the fact that the non-public registry used for subject recruitment has a significant amount (~50%) of post-stroke individuals under the age of 55.

## 5. Conclusion

The results in our study show that balance deficits seen post-stroke during standing are also present in sitting well into the chronic stage of stroke onset. Since sitting balance is primarily controlled by trunk musculature, we believe that this dysfunction is due to poor postural control of the trunk—an area that has not received a significant amount of research. Since unsupported sitting is a simplified postural model of standing [23], the disturbed sitting balance seen post-stroke may strongly predict the dysfunction of one's postural control system when task complexity increases (i.e. standing and walking). Furthermore, the complexity of standing balance sometimes limits the type of cohort that can be investigated. Thus, studying sitting balance may allow researchers to investigate postural control strategies of stroke subjects who are severely impaired, closer to stroke onset, or both. This is significant for it may quantitatively predict future standing balance deficits or more importantly, allow early balance training during sitting that might possibly improve initial standing balance and expedite further interventions of balance and gait.

## Conflict of interest

No author claims to have any conflicts of interest for the data reported in this manuscript.

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