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Preliminary investigation of the lateral postural stability of spinal cord-injured individuals subjected to dynamic perturbations

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Study Design: A study of the impact of spinal cord injury (SCI) on seated balance was conducted by comparing the results obtained from experiments with able-bodied and SCI subjects.
Objectives: The purpose of this preliminary study was to examine the lateral postural stability of seated individuals with SCI in a dynamic environment.
Setting: Experiments were conducted at the Cleveland Clinic Foundation in Cleveland, Ohio.
Methods: Controlled perturbations were applied to each subject, seated in a wheelchair, through the use of a servo-controlled tilt platform. The platform was rotated so as to create disturbances similar in nature to those experienced in the frontal plane during left turns in a vehicle. Four quadriplegic, four paraplegic, and five able-bodied subjects participated in this study. Kinematic information and center of pressure (COP) movement were recorded.
Results: None of the spinal cord-injured subjects was able to maintain his stability when exposed to the stronger perturbations, while all of the able-bodied subjects stayed upright for all of the trials. On an individual basis, injury level was not always indicative of balance. However, regression results suggest a correlation between ability to perform static leaning and dynamic balance ($P<0.001$).
Conclusions: SCI subjects lost stability under dynamic conditions even though they were stable in the static situation. Initial results also raise some questions about where and when external support may be needed. Information of this nature could help to guide the design of new lateral supports with improved client acceptance.

Keywords: spinal cord injury; posture; equilibrium; wheelchairs

Introduction

Traditionally, rehabilitative practice with regard to seated posture has focused on the static situation. Emphasis has been placed on clinical areas such as the reduction of pressure sores, control of abnormal tone, and skeletal alignment.[1] With respect to wheelchair users with spinal cord injury (SCI), prevention of or compensation for kyphosis and pelvic obliquity are typical concerns. Supports are used to counterbalance the effects of gravitational forces.

More recently, researchers have begun to examine postural control under conditions in which the magnitude and direction of the perturbing force change with time. This situation arises during task performance; acceleration of objects or the person's own body creates forces that must be resisted by the trunk in order for the individual to maintain balance. In three separate studies, distances that could be reached,[2] time to complete the tasks,[3] and muscular effort required[4] were all found to be worse in individuals with SCI as compared to those with no disability. Postural instability may hinder the ability to perform transfers as well.[5]

Seated balance may also be challenged when these dynamic forces are externally applied. This is the focus of this study. External perturbations may occur, for example, when riding in a wheelchair over uneven terrain. One survey of falls from wheelchairs and wheelchair tipping found that the majority of these incidents occurred either on sloping ground or at sudden changes in ground elevation.[6] Significant moments, forces, and accelerations were recorded in experiments simulating wheelchairs falling off or running into curbs.[7]

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Another common environment in which these perturbations occur is inside a moving vehicle. Turns and brakes generate moments that must be counteracted in order to maintain equilibrium. Loss of postural control can result in misalignment with respect to restraining shoulder belts, thereby greatly reducing their efficacy.[8] With disabled drivers, loss of stability leading to accidents is a definite concern.[9,10] Just as importantly, difficulty in maintaining balance may prevent an individual from driving.

However, relatively little research has analyzed the impact of external forces on the posture of individuals with SCI. Bernard *et al* imposed accelerations on paraplegic athletes.[11] However, they employed only sinusoidal inputs and the direction of the accelerations was limited to the sagittal plane of the subject. Sprigle and Linden examined the balance of tetraplegic subjects during controlled driving maneuvers.[12] They had difficulties, though in generating repeatable inputs.

The goal of the present work was to provide a preliminary investigation of the effects of SCI on dynamic stability. Emphasis was placed on lateral balance. Characteristics of the subjects and their responses were examined for correlation with stability under dynamic conditions, in the hope of providing insights for the future development of lateral supports. Several studies have described improved performance with the addition of trunk orthoses or belts.[2,5,7] However, use of these items is limited due to unwanted restrictiveness as well as difficulty in donning and doffing the current designs.

Sustained perturbations were applied in the frontal plane of the subject. Responses were quantified by estimation of both the displacement and velocity of the subject's center of pressure (COP) with respect to the seat. Estimates of joint torques were made from kinematic body segment data, employing inverse dynamics for a multi-segment model of the upper body. Comparisons were made among the data from tetraplegic, paraplegic, and able-bodied subjects.

Methods

Equipment

To analyze dynamic stability, external perturbations were applied to the subject through the use of a servo-controlled tilt platform. The subject sat in a standard manual wheelchair, which was rigidly secured to the platform. The device arrangement is shown in Figure 1.

The authors chose to impose a controlled disturbance similar in nature to what might be experienced during daily living. Namely, they decided to examine lateral balance during sustained perturbations resembling those experienced during turning maneuvers in a vehicle. Rotation of the tilt platform with respect to the gravitational field generates moments in the trunk of the subject, in a manner akin to moment production by inertial forces incurred during turning.[13] The magnitude and timing for the platform rotational profiles used for this study were modeled after vehicle accelerations recorded along the lateral axis of the vehicle during controlled driving maneuvers. The constant-radius left turns were derived from the Canadian guidelines for testing wheelchair securement devices.[14]

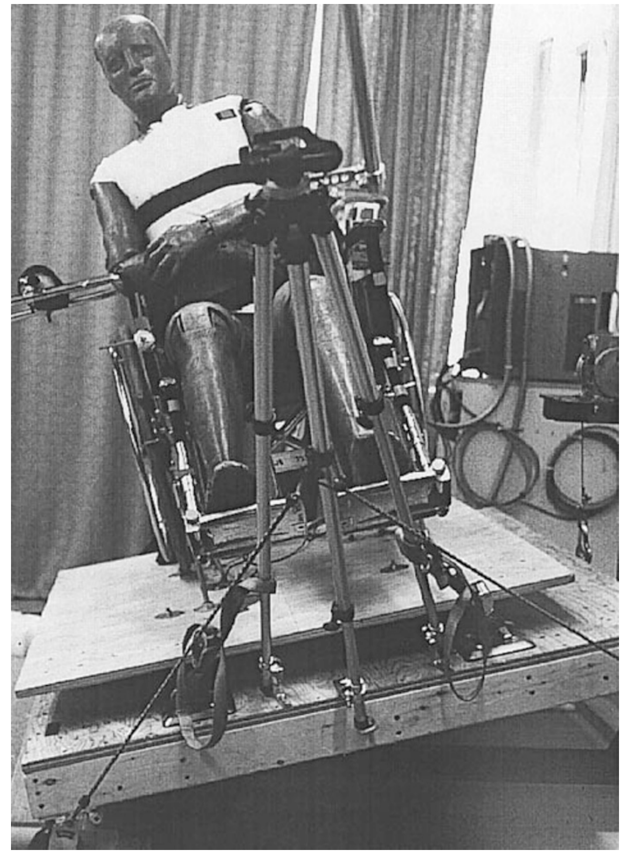

Figure 1 Tilt platform is rotated about right castors to create a dynamic disturbance with a Hybrid II anthropomorphic test dummy as the wheelchair occupant

The steady-state portions of the platform rotations correspond to the targeted steady-state vehicle acceleration levels, 0.2 *g* and 0.4 *g*. The chosen vehicle acceleration levels correspond to a lower and an upper value, respectively, for lateral accelerations experienced during controlled driving maneuvers.[15] For each of the two steady-state levels of perturbation, two different profiles were created, each with a different rate of disturbance application. In driving, different rates can arise from different initial velocities heading into the turn. With the tilt platform, the disturbance application rates translate into rates of platform rise. In this study, one rate was slow, corresponding to a quasi-static test, while the other rate was faster. For the faster rate, the acceleration curves from the controlled driving maneuvers served as guidelines. The profiles for the four different combinations, referenced from hereon as LS, LF, HS, and HF, are shown in Figure 2.

A standard manual wheelchair frame (Everest and Jennings) was bolted to the tilt platform through load cells. The frame was reinforced to make it rigid. The original sling seat of the wheelchair was replaced with a foam cushion (5 cm of HR70 topped by 2.5 cm of HR32) affixed to a dropped seat pan. The new seat measured 40.6 cm in width. The wheelchair had a 3° seat angle[16] and a 90° seat-to-back angle with a 7° cane angle in the backrest. The right armrest was replaced with a custom unit that diminished the chance of accidental contact of the arm with the armrest (Figure 1). A lap belt secured the subject to the wheelchair.

Four load cells were used to measure axial forces between the tilt platform and the subject-wheelchair system. Motion of the subject's COP with respect to the seat was computed from these vertical ground reaction forces.[17] Trials were recorded on videotape by a camera that rotated with the platform.

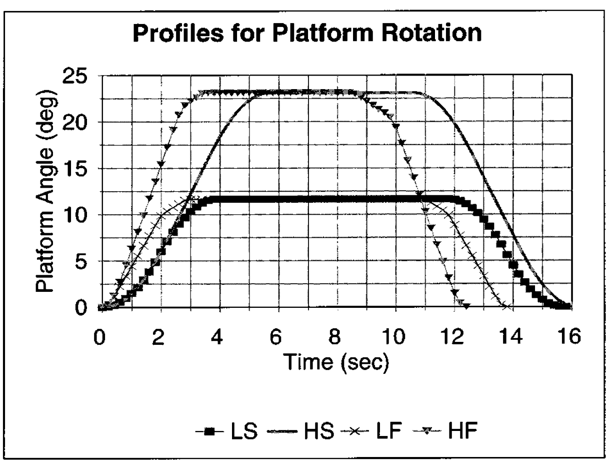

Figure 2 Platform rotational disturbance profiles used in the experiments. L: lower steady-state angle; H: higher steady-state angle; S: slower rate of platform rotational velocity; F: faster rate of platform rotational velocity

Subjects

Thirteen male subjects participated in this study. Permission to utilize human participants was obtained from the Institutional Review Board and the participants indicated their informed consent. Subjects were divided into three categories based on injury level: tetraplegic (TP), (complete lesion between C5–C7); paraplegic (PP), (complete lesion between T2–T9); and able-bodied (AB), (no injury). All SCI was of traumatic origin which had occurred at least 3 years prior to this study. Potential subjects exhibiting the pelvic obliquity or scoliosis were excluded. Each subject could transfer independently. Table 1 describes the subject characteristics.

The height of the subject's center of mass (COM) was determined for the individual seated in the wheelchair with arms folded across the chest. Moment balances for data at different platform rotation angles were solved simultaneously to find the height. The COM height was referenced to the base of the seat.

Procedure

Each subject participated in two trials for each disturbance profile. During the tests, the subject was instructed to look at a target, directly ahead of him, which remained fixed with respect to the tilt platform. The individual was told to attempt to keep his arms crossed against his chest throughout the maneuver. This posture was chosen in order to focus on the performance of the trunk musculature in providing postural control.[18] Use of the upper extremities to stabilize oneself signified a loss of stability. The onset of instability was defined as the time at which an upper extremity contacted either the wheelchair or the legs.

Table 1 Physical description of the subjects who participated in the trials

Category	*Subject*	*Age*	*Height (cm)*	*Weight (kg)*	*Injury level*	*Years post-injury*	*Seated COM height (cm)*
TP	DL	27	188	64.3	C5–C6	3	29.5
	MB	28	185	60.9	C5–C6	6	30.0
	KR	35	183	82.7	C6–C7	17	29.5
	TT	44	180	76.3	C7	20	27.0
PP	JB	35	177	85.7	T2–T3	29	28.5
	JW	38	179	79.9	T4–T5	16	27.5
	SH	35	191	76.4	T7	12	29.0
	AC	28	183	83.9	T9	6	28.0
AB	CH	29	160	58.2	NA	NA	24.5
	DK	29	191	86.0	NA	NA	29.0
	GN	30	175	64.7	NA	NA	24.0
	MP	33	178	55.2	NA	NA	23.0
	RS	29	175	67.9	NA	NA	26.5

TP: tetraplegic; PP: paraplegic; AB: able-bodied; NA: not applicable

Onset of instability was determined from examination of the videotape. Synchronization of the videotape with electronic data enabled the determination of the tilt platform angle at this point.

COP movement was recorded throughout the trial. Absolute movement of the COP, however, does not properly describe the stability of the subject. COP displacement may be significant without the subject losing postural stability as long as the excursion limit, which may vary among individuals, is not surpassed. COP movement should be normalized with respect to the capability of each subject to better gauge postural control.[13] Thus, COP data was normalized according to the limit of the COP motion that the subject could maintain without use of the upper extremities. Maximum volitional COP displacement was found by having the subject, with arms crossed against the chest, lean as far as possible to his right, up to the balance point. The measured displacement of the COP was divided by this maximum value to attain the performance index termed FLCOP.[13] The 'FL' acronym refers to the fraction of the theoretical limit that the measured movement represents. In practice, FLCOP values greater than one were seen without loss of stability because the limits were based on voluntary performance.

Research in standing balance has suggested that the state of the COP (its velocity as well as its position), plays an important role in determining the stability of an individual.[19] Thus, an index related to COP velocity, DFLCOP, was developed to examine this concept with respect to seated balance.

$$\mathrm{DFLCOP} = \frac{d(FLCOP)}{dt} \qquad (1)$$

To compute the DFLCOP index, the FLCOP signal was numerically differentiated.

Kinematic data were also recorded. Reflective markers were placed on the snugly fitting T-shirt of the subject to facilitate measurement of body segment movements. The markers were located on the left and right anterior superior iliac spine and on the beltline halfway in-between, on the xiphoid process, on the sternoclavicular notch, on the chin, and on the forehead. These positions enabled estimation of the rotation of the pelvis, lower torso, upper torso, neck, and head in the frontal plane. The points of rotation between these segments were arbitrarily chosen to roughly correspond to the spinal levels L5/S1, T12/L1, and C7/T1. Figure 3 shows the marker locations along with the corresponding angular displacement measurements. All angles were measured with respect to the vertical from manual selection of marker location on images digitized from the videotape. The data describing body segment rotation was smoothed by low-pass filtering at 2 Hz.

Inverse dynamics was employed to compute the joint torques for the model from the joint positions, velocities, and accelerations.[20] In these estimations of joint torques, the effects of the seat back were not included. Joint torques were estimated using a three-link two-dimensional model of the upper body in the frontal plane. The links were assumed to be connected by revolute joints. The links represented the lower torso, the upper torso and arms together, and the neck and head together. The parameters for the links were calculated from body segment measurements through the use of an anthropomorphic software package[21] (Generator of Body Data, Wright-Patterson Air Force Base, OH, USA).

Figure 3 Location of reflective markers in the frontal plane for use in obtaining body segment angles from video

Analyses

The impact of subject category on the tilt platform angle at instability onset was tested using an ANOVA (SAS®). For the cases in which instability did not occur, the maximum tilt platform angle for the trial was considered as the angle at instability. Differences in angle among the various categories were analyzed for statistical significance through multiple comparisons.

The relationship between the peak value of FLCOP and DFLCOP for each trial and the loss or maintenance of balance was analyzed through logistic regression. From the regression models, the values of these indices corresponding to the 0.5 probability of loss of stability were computed. The resulting values were then tested as thresholds for the prediction of instability on the same data sets. For all cases in which a fall occurred, the peak value for each index up to the point of instability was obtained.

Torque was also analyzed by normalization with respect to a maximum voluntary level. The maximum for the torque about the joint between the pelvis and lower torso (T_{LT}) was estimated using the limit of COP displacement at the balance point. The peak value for T_{LT} was found for each trial. For the trials in which the subject fell, the peak was attained for the portion of the trial up to the onset of instability. Peak T_{LT} was normalized by dividing its value by the approximation for the maximum for the given subject.

Results

Table 2 lists the platform angles at which stability was first lost for the SCI subjects. Only data for the SCI subjects is given because all of the AB subjects maintained stability for all of the trials. The ANOVA performed to examine the impact of subject category on platform angle at the onset of instability verified that subject category was significant ($P<0.05$). Multiple contrasts revealed that across all of the disturbance profiles the angle was highest for AB subjects and lowest for tetraplegic subjects ($P<0.001$).

Table 3 lists the limit that each subject could voluntarily move his COP, in the absence of applied perturbations, before loss of balance. The relationship of this static measure to dynamic stability was assessed using regression. A linear regression was performed between the natural log of the ΔCOP_{max} and the tilt platform angle at instability onset for the HS disturbance. The slope and the regression were significant ($P<0.001$, $R^2=0.65$).

Peak FLCOP and DFLCOP were examined for a correspondence with instability. From the logistic regressions, threshold values were calculated for predicting whether a fall occurred given the peak FLCOP or DFLCOP for a trial. Use of either the FLCOP or DFLCOP threshold resulted in two falsely negative and one falsely positive predictions of instability ($n=99$). In cases where imbalance occurred, the elapsed time to reach the DFLCOP threshold was highly correlated with elapsed time to the onset of instability (Pearson correlation coefficient, $r=0.95$). The temporal relationship between FLCOP threshold and instability onset was not as strong ($r=0.90$).

The kinematic response, as characterized by body segment trajectories, illustrated differences between the SCI and able-bodied subjects. The segment angles for the AB subjects typically stayed within $\pm 5°$. Rotation of the lower torso and pelvis tended to be greater in the SCI subjects, while residual control of trunk musculature permitted them to maintain an upper torso posture more like that of their AB counterparts (Figure 3). Relative rotation of the lower torso with respect to the upper torso was significantly greater in the SCI subjects ($P<0.05$). In cases where balance was lost, rotation of the pelvis and lower torso in the direction of the fall usually preceded that of the rest of the body.

Table 2 Platform angles at which subjects with disabilities first became unstable

		Tilt platform angle at onset of instability (deg)			
	SCI	*Platform rotational profile*			
Subject	*Level*	*LS*	*LF*	*HS*	*HF*
DL	C5–6	7.2	9.6	11.0	14.8
		8.1	9.9	12.7	–
MB	C5–6	6.2	10.6	6.7	11.1
		7.3	9.6	6.1	9.0
KR	C6–7	S	S	20.2	21.8
		S	S	16.6	22.1
TT	C7	S	S	15.4	7.5
		S	S	11.8	13.0
JB	T2–3	S	S	21.4	22.0
		S	S	23.1	21.3
JW	T4–5	S	S	21.6	17.8
		S	S	18.6	19.8
SH	T7	7.1	7.4	6.9	12.6
		6.1	9.2	7.9	14.2
AC	T9	S	S	15.4	17.6
		S	S	18.0	17.3

'S' signifies that stability was maintained. '–' indicates that datum from a trial was not used and the entry was treated as missing datum in the statistical analyses. Maximum platform angles during each type of disturbance were: LS and LF: 11.7°; HS and HF: 23.1°

Table 3 Peak voluntary movement of COP to subject's right up to limits of stability

Subject	*SCI level*	*Max ΔCOP (cm)*
DL	C5–C6	4.09
MB	C5–C6	3.81
KR	C6–C7	3.66
TT	C7	4.78
JB	T2–T3	5.16
JW	T4–T5	6.35
SH	T7	2.39
AC	T9	7.32
CH	AB	18.95
DK	AB	18.77
GN	AB	17.81
MP	AB	20.45
RS	AB	17.70

Table 4 Peal normalized joint torque for T_{LT}, calculated from kinematic data. Italicized and bolded entries indicate that stability was maintained throughout a trial. The other entries were computed for the duration of the trial up to the point of instability

	Estimated peak normalized torque for T_{LT}			
	Platform rotational profile			
Subject	*LS*	*LF*	*HS*	*HF*
DL	2.06	1.84	2.10	2.46
MB	0.03	0.87	0.22	0.49
KR	***0.42***	***0.50***	0.85	1.96
TT	***1.02***	***0.83***	1.02	0.64
JB	***0.66***	***0.75***	1.18	1.22
JW	***0.60***	***0.59***	0.82	1.12
SH	1.13	0.31	0.12	0.32
AC	***0.15***	***0.10***	0.08	0.13

T_{LT}: torque about joint between lower torso and pelvis

Table 4 lists the normalized score for peak torque around the joint between the pelvis and lower torso, T_{LT}, for each input disturbance for each SCI subject. Numbers that are italicized and bolded indicate trials in which the subject maintained balance. Normalized values greater than one are possible because the subject may have begun to fall before his arm contacted the wheelchair.

Discussion

While all of the SCI subjects were stable under static conditions, they all became unstable in a dynamic environment. Sustained, laterally applied perturbations of magnitudes associated with normal driving caused instability in the SCI subjects. One can see from Table 2 that the majority of the SCI subjects were stable for the platform rotations with the lower steady-state angle (LS and LF), but no SCI subject kept his balance during the trials with the higher level of rotation (HS and HF). It should be noted that Table 2 might be misleading in regard to comparing trials with different rates of platform rise. Greater platform rotation during the reaction time required for subjects to reach with their arms during falling sometimes led to larger angles for the 'F' trials. Across all subjects, the peak FLCOP values were statistically greater for the LF than the LS and for the HF as compared to the HS trials.

As expected, paraplegic subjects, as a group, exhibited greater stability than tetraplegic subjects. However, on an individual basis, the level of injury was not always a good prognosticator of stability. For example, the subject with one of the lowest injury levels, T7, was one of the least stable subjects of all. Interestingly, Table 3 shows that he had the smallest maximum distance that he could voluntarily move his COP by leaning right, without losing balance. For the small sample size of this study, the maximal COP displacement tolerated was highly related to dynamic stability, as demonstrated by the regression analysis. Use of a simple test of a patient's ability to lean could help clinicians make initial estimates of stability under dynamic conditions without having to expose the subject to large disturbances. Another study has suggested some relationship between the degree to which seated individuals can lean and their functional capabilities.[22]

The results show that all of the SCI participants could have benefited from some type of lateral support. Of course, use of the arms would have improved stabilization for some of the subjects, but in practice this interferes with task performance or places the joints of the arm in potentially injurious postures in a dynamic environment. Certainly, some of the subjects may have performed better in their own wheelchairs, sized and contoured specifically for them. To avoid possibly confounding postural control with the adequacy of the prescribed wheelchair, however, a single testing configuration was used. The size of wheelchair chosen (40.6 cm) provided a reasonable match to the range of the distance across the hips measured on the SCI subjects (35.6–43.2 cm). It should be noted, though, that the three least stable subjects had measurements toward the lower end of this range. As far as support, in some cases the full sling back and uprights of the testing wheelchair seemed to actually afford greater restraint of lateral motion than the individual's wheelchair. The higher back of the test wheelchair provided more frictional resistance than the lower back favored by many subjects. The greater obtrusiveness of the back supports in the test wheelchair also impeded movement to an increased extent.

In general, the participants in this experiment wanted as little restriction to desired movement as feasible. This underscores the challenge for the design of lateral supports. Currently, lateral supports are not widely accepted because they interfere with functional tasks in a static environment. Acceptance would probably grow if the supports could be engineered to be minimally restrictive until actually required for support.

One question to address is where support is really needed. Analysis of body segment rotation suggests that the majority of the SCI subjects exhibited reasonable control of the upper torso and head. Instability seemed to result from an inability to prevent rotation of the pelvis and lower torso. The greater rotation led to requirements for greater torque, on average for the SCI as compared to the AB subjects, in order to maintain balance. Restriction of the rotation of the pelvis and lower torso without limitation of upper torso movement may provide sufficient stabilization for a number of SCI individuals.

The data from this study also provide some insights on when SCI subjects may feel the need for external support. While trunk strength undoubtedly plays a large role in sitting balance,[18] evaluation of Table 4 reveals that there were cases in which a subject reached to stabilize himself even though the joint torque necessary to maintain stability was below his theoretical limit. In fact, in some instances the peak torque preceding a fall was less than the value for the same subject for a trial in which he stayed upright. COP displacement up to the onset of instability did not fully explain the loss of stability, either. The average of the peak FLCOP taken up to the onset of instability, for all trials in which it occurred, was significantly less than the threshold value. However, the average peak DFLCOP prior to the onset of instability was greater than the threshold value. Time to reach DFLCOP threshold was highly correlated with the time of instability onset. This suggests that, as with standing balance,[19] velocity of COP motion is important in examining overall stability.

Thus, future studies of the seated dynamic balance of SCI individuals seem warranted. The potential relationship between static leaning and dynamic balance could prove to be clinically useful. The minimal height requirement for a lateral support to provide stabilization is another direction for research. Further examination of the association between rate and stability may be beneficial for the design of lateral supports. For example, future devices may be able to increase the amount of support provided dependent on the velocity of their displacement or the rate of increase in contact force encountered.

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