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Postural stability of wheelchair users exposed to sustained, external perturbations

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Abstract--The postural stability of wheelchair users experiencing external perturbations was examined. Rotation of a tilt platform generated moments in the trunks of subjects seated in a manual wheelchair on the platform. The magnitude and duration of the moments were on the order of those that might be encountered in the sagittal plane during controlled braking maneuvers in a vehicle. Four subjects with tetraplegia, four with paraplegia, and five controls participated in experimental trials on the platform. As input, four different Disturbance profiles with either a 0.2g (gravitational acceleration) or 0.4g maximal level were imposed. The majority of the subjects with spinal cord injury lost balance at Disturbance levels below 0.2g. The results suggest that the rate of change of the applied perturbation may also affect stability. The use of a stability index based on normalized motion of the center of pressure with respect to the seat showed efficacy in characterizing the response.

Key words: *automotive safety, postural stability, wheelchair users.*

INTRODUCTION

Transportation is vital to full integration into modern society. Mobility is an especially important issue for people with disabilities. Twenty-eight percent of disabled individuals who are unemployed cite lack of accessible transportation as a major obstacle to finding work (1).

For the over 1 million wheelchair users living in the U.S. (2), traveling in motor vehicles can lead to significant challenges to their postural control. The inertial forces inherent to braking and turning generate moments about the spinal and pelvic joints. The diminished muscular control that dictates utilization of a mobility aid may affect not only the extremities, but the postural muscles of the trunk as well.

While considerable research has focused on improving accessibility to, and crash protection in, motor vehicles for disabled individuals (3-7), relatively little attention has been given to the response of the rider at acceleration levels normally seen during driving. In addition to the comfort and fatigue issues faced by disabled passengers in maintaining seated balance, safety concerns also exist. Loss of an upright posture could cause an individual to become improperly aligned with the vehicle restraint devices. One group studying the crash simulation of wheelchair riders in vehicles found excessive trunk motion resulted from an initial posture of sideways lean away from the shoulder belt (8). Over 2,000 injuries involving wheelchair users traveling in motor vehicles were reported over a 5-year period according to one study (9); the accidents resulted from either the wheelchair occupant falling out of the chair or both chair and occupant tipping together.

With disabled drivers, loss of stability leading to accidents is a definite concern (10), and documented cases of this situation exist (11). Just as importantly, difficulty in preserving balance may prevent an individual from driving. One pilot survey indicated that 40 percent of the respondents who did not drive experienced stability problems at least some of the time (12).

Wesson et al. (13) looked at the movement of an anthropomorphic test dummy seated on a wheelchair inside a vehicle during normal driving conditions. Lateral displacements of its chest exceeded 18 cm for a driving maneuver that produced a lateral acceleration of 0.5g (g, gravitational acceleration). Linden and Sprigle (14,15) examined the response of tetraplegic passengers in vans during controlled driving maneuvers. They found evidence of destabilization of some subjects at acceleration levels below 0.2g. However, in controlled driving experiments, vehicle acceleration is difficult to control and to repeat from one trial to the next (14).

The aim of this study was to gain further insight into the effects of spinal cord injury (SCI) on postural stability when the individual is exposed to perturbations on the order of magnitude of those seen in normal driving. To better control the input, experiments were conducted in a laboratory environment using a tilt platform. Both the magnitude and the rate of change of the perturbation were varied. Persons with tetraplegia or paraplegia and control subjects participated. Methods were developed to characterize the postural response of each individual in terms of normalized movement of the center of pressure of the subject with respect to the seat. Rotation of body segments was also examined. The effects of SCI and of the type of perturbation on postural response were tested using analyses of variance.

METHODS

Disturbance Input

Perturbations were created through the use of a servo-controlled tilt platform, details of which are described elsewhere (16,17). With the platform, rotation of the subject with respect to the gravitational field is used in place of the translational acceleration that occurs during driving. Simulation of driving maneuvers using strictly translation is not generally feasible, due to excessive expense and space requirements. For example, simulation of a 0.2g-deceleration from 35 km/hr would require over 20 m of controlled travel. Hence, for this study, the platform and wheelchair were pitched forward, the direction the subject faced, rather than translated backwards.

Typically, the perturbation created by a given driving maneuver is quantified in terms of accelerations recorded in the vehicle (13,14). **Figure 1** shows an example of a recording of vehicle deceleration, along the long axis of the vehicle, during a controlled braking maneuver. Inertial forces, characterized by this deceleration, generate moments in the trunk of the rider. These moments must be resisted in order for the individual to remain upright.

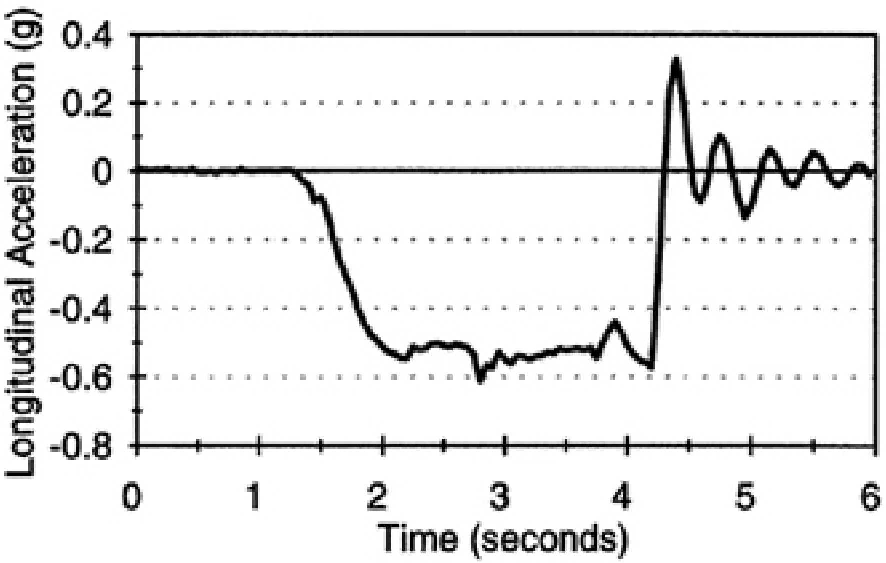

Figure 1.
Linear acceleration recorded with an accelerometer in the wheelchair bay of a vehicle undergoing a controlled 0.5g braking maneuver. Acceleration direction is aligned with the long axis of the vehicle.

Similarly, with rotation, the components of gravity parallel with the seat of the wheelchair create moments about the joints of the subject. To quantify the perturbation imposed by rotation of the platform, the concept of a restoring torque is introduced (16,17). The restoring torque is defined by the authors as the amount of torque that would have to be generated about the L5/S1 joint of the spine in order to resist the perturbation and remain upright. Since the restoring torque is a measure of the input and not of the response, for its computation the subject does not move with respect to the wheelchair. As described by Kamper et al. (16,17), the restoring torque is estimated by a two-dimensional model of the upper body as a single link connected to the lower body through a pin joint (see **Figure 2** for depiction of forces):

[1]

$$\tau_y \approx d * R_x$$

[2]

$$\tau_y \approx d * m_{HAT}\left(\ddot{\theta}_y r \sin\alpha + \dot{\theta}_y^2 r \cos\alpha - g \sin\theta_y\right)$$ **[2]**

$\theta_y \equiv$ angle of platform pitch; $\alpha \equiv$ angle between subject CM_{HAT} and the point of platform rotation with respect to the platform plane; $CM_{HAT} \equiv$ location of the center of mass for the head, arms, trunk; r $\equiv$ distance between the subject CM_{HAT} and the point of platform rotation; d $\equiv$ distance between the L5-S1 joint and subject CM_{HAT}; **M**HAT $\equiv$ mass of the head, arms, trunk; g $\equiv$ gravitational acceleration; $R_x \equiv$ reaction force in x-direction at CM_{HAT}.

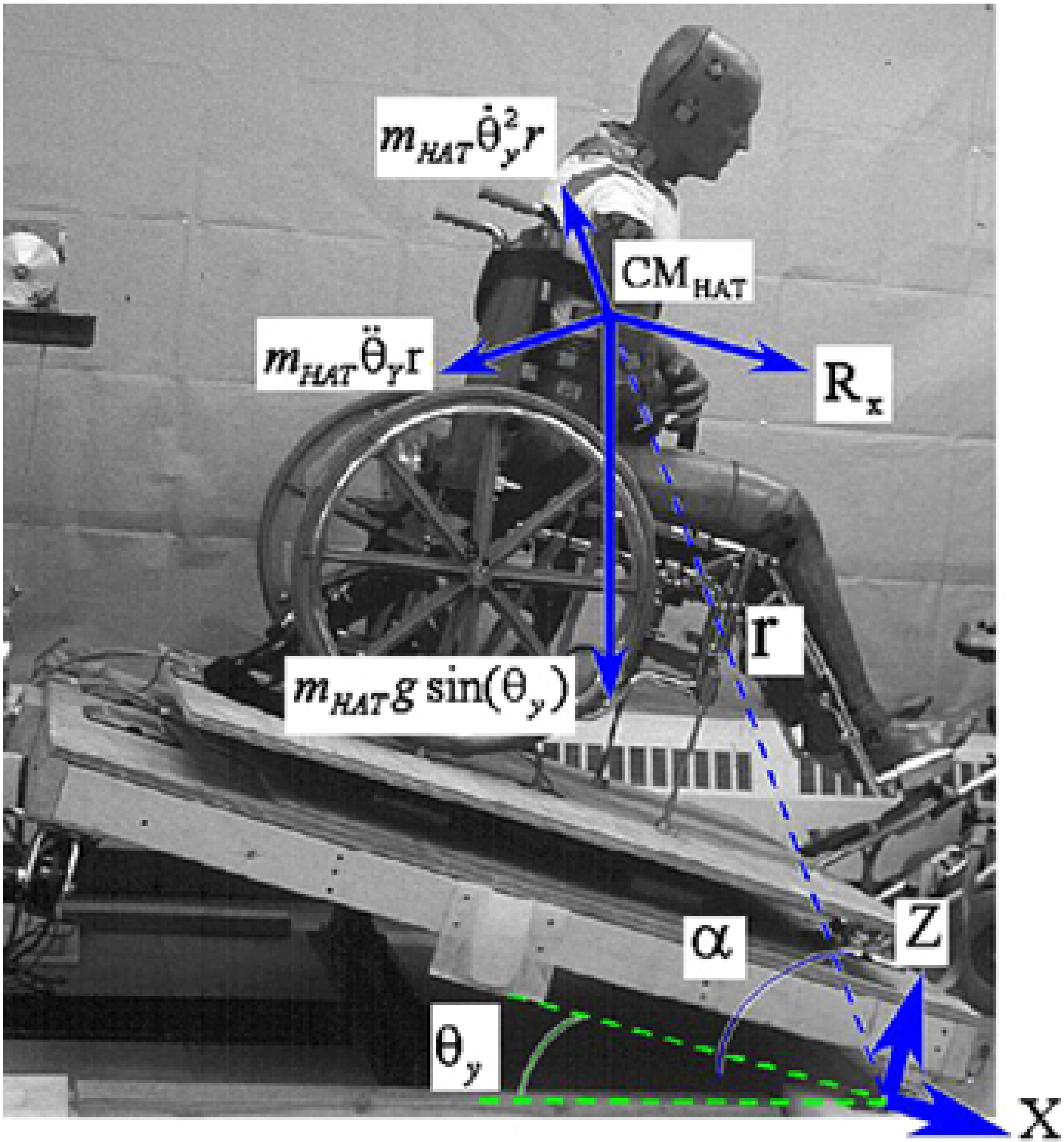

Figure 2.
Platform is pitched forward to simulate a "brake" with a Hybrid II anthropomorphic test dummy as the wheelchair occupant. Axis of rotation was chosen to reside at the front edge rather than the center of the platform in order to minimize starting platform height to facilitate transfers. Forces generated by platform rotation are shown.

The term in parentheses in Equation 2 plays a role similar to that of the inertial acceleration during braking in determining the magnitude of the perturbation. This term was kept the same for every subject for a given perturbation profile. Hence, the controlled input variable for each set of experiments was defined by:

[3]

$$Disturbance = \left(\ddot{\theta}_y r \sin\alpha + \dot{\theta}_y^2 r \cos\alpha - g \sin\theta_y\right)$$

[3]

As in the vehicle for a given driving maneuver, the magnitude of the restoring torque varied according to the body parameters of the subject. Disturbance as a function of time, however, was the same for all of the subjects just as the same maneuver is used for all subjects in controlled driving experiments. The differences in r and α, resulting from varying heights of the center of mass in the subjects, had negligible effect on Disturbance; therefore r and α were assumed to be fixed, and platform rotation was invariant across subjects.

Maximal Disturbances corresponding to two steady-state vehicle acceleration levels, 0.2g and 0.4g, were chosen for this study to represent a lower and an upper value, respectively, for accelerations experienced during controlled driving maneuvers (13,14,16). The durations of the steady-state portion of the Disturbances were selected to match the average of those observed during controlled testing in a van (18). The braking maneuvers used for that study were derived from the Canadian guidelines (19).

For each of the steady-state levels of Disturbance, two different profiles were created, each with a different rate of change of the Disturbance. The slower rate corresponded to a quasi-static test, while the faster followed acceleration curves from the controlled driving maneuvers (18).

Nomenclature was developed to describe the different input profiles with two-letter acronyms. The first letter, "L" indicates the lower level of steady-state Disturbance, while "H" represents the higher level. The second letter signifies the rate of change: "S" stands for slow, "F" for fast. For example, the label identifying the profile with the higher level of Disturbance and the faster rate of change is HF. The profiles of the four different combinations are shown in **Figure 3.** Each subject was to be exposed to all of the profiles.

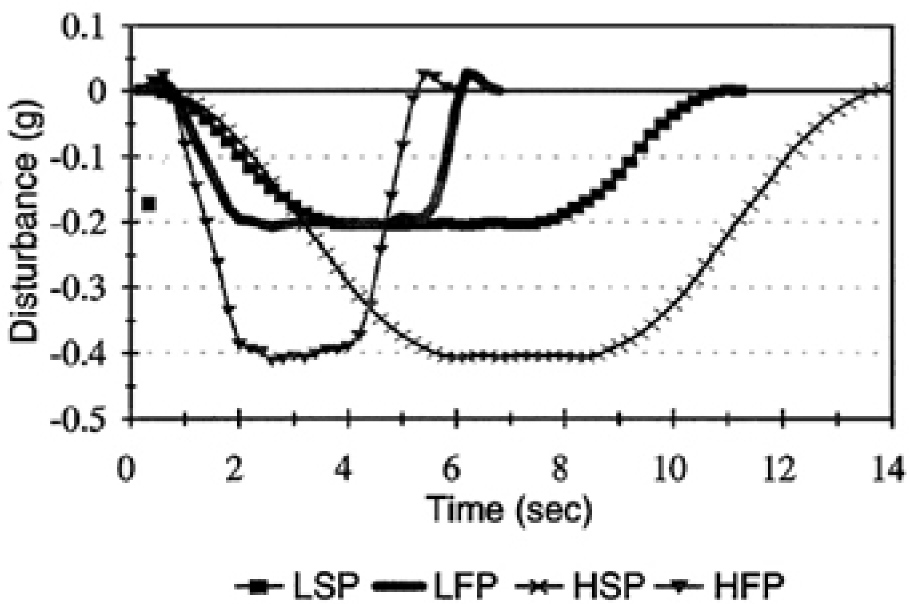

Figure 3.
The platform rotational profiles used in the experiments. L: lower steady-state Disturbance level; H: higher steady-state level; S: slower rate of change of Disturbance; F: faster rate of change.

A standard manual wheelchair (Everest & Jennings, Earth City, MO) was set on the platform with its frame rigidly fixed to the support surface in a manner that precluded movement (16,17). This reduced the potential for wheelchair motion to confound subject response. The formerly collapsible frame was made rigid. The original sling seat was replaced with a 7.5-cm foam cushion attached to a plywood base. The width of the wheelchair seat was 40.6 cm. The seat was angled 3° backward with respect to the horizontal, a value within the suggested range of 3-10° (19).

Subjects

Thirteen male subjects participated in this study, with the subjects divided into three categories based on injury level: persons with tetraplegia, TP, (complete lesion between C5-C7), those with paraplegia, PP, (complete lesion between T2-T9), and controls with no injury, AB. All SCI was of traumatic origin, occurring at least 3 years prior to this study. Individuals presenting with pelvic obliquity or scoliosis were excluded. All but one subject drove. Permission to utilize human participants was obtained from the Institutional Review Board and the participants indicated their informed consent. **Table 1** describes the subject characteristics.

Table 1.

Physical description of the subjects who participated in the trials.

Category	Subject	Age	Height (cm)	Weight (kg)	Injury Level	Post-Injury
TP	DL	27	188	64.3	C5-C6	3
	MB	28	185	60.9	C5-C6	6
	KR	35	183	82.7	C6-C7	17
	TT	44	180	76.3	C7	20
PP	JB	35	177	85.7	T2-T3	29
	JW	38	179	79.9	T4-T5	16
	SH	35	191	76.4	T7	12
	AC	28	183	83.9	T9	6
AB	CH	29	160	58.2	NA	NA
	DK	29	191	86.0	NA	NA
	GN	30	175	64.7	NA	NA
	MP	33	178	55.2	NA	NA
	RS	29	175	67.9	NA	NA

TP=tetraplegic; PP=paraplegic; AB=able-bodied; NA=not applicable; post injury in years.

Wearing cotton sweatpants and a T-shirt, each subject was instructed to remain facing forward while focusing on a target fixed with respect to the platform. He attempted to keep his arms crossed against his chest throughout the duration of the maneuver. This was done to eliminate the use of the hands and arms in maintaining balance, in order to concentrate on the performance of the trunk musculature. Use of the upper limbs to stabilize oneself signified a loss of stability. A lap belt, connected to the platform, was used to secure the subject. Prior to the start of each trial, the subject was positioned with his trunk centered in the chair back and the posterior of the buttocks to the back of the seat. Footrests were used to support the legs, but the footplates were rotated out of the way in the tests with controls to prevent possible voluntary torque generation about the ankles.

Quantification of Subject Response

An 8-mm video camera, focused on the right side of the subject, captured movement in the sagittal plane. Since it was connected to the platform itself, the camera did not move relative to the platform during the testing. This greatly improved resolution by reducing the required size of the field of view. Video data were examined to determine if a loss of stability, defined as contact of the upper limbs with either the wheelchair or the legs, occurred during a trial. Synchronization with the input permitted the detection of the Disturbance level corresponding with the onset of instability.

Center-of-pressure (CP) movement has been widely used in quantifying standing balance (21), but to provide a continuous measure of seated postural stability, a new index was developed, based on motion of the subject's CP with respect to the seat (CPS). CPS was computed from vertical ground reaction forces measured by load cells placed between the wheelchair and the platform; compensation was performed for the presence of the wheelchair and the rotation. Detailed formulation of its computation is provided elsewhere (16).

Absolute movement of the CPS, however, does not describe the stability of the subject. CPS displacement may be significant without the subject losing postural stability, so long as the excursion limit, which may vary among individuals, is not surpassed. CPS movement should be normalized with respect to the capability of each subject to better gauge postural control. Koozekanani et al. (21), advanced the concept of "stability margin" with respect to standing balance. The stability margin represented the minimum distance from the location of the CP to the edge of the base of support. If the CP were to go outside the base of support, the individual would fall.

The concept of a stability margin, however, must be adjusted for the seated situation. Koozekanani calculated his standing stability margin with respect to biomechanical limits, the outline of the contact area between the foot and the support surface. The seated analogue is the perimeter of the contact area between the legs and pelvis of the subject and the seat: when the CP moves beyond these biomechanical limits, the subject falls out of the chair. In a vehicle, though, the subject is supposed to be wearing a lap belt that precludes falling out of the chair. The belt makes the biomechanical limits almost infinite. However, limits still exist with regard to CP motion before the subject loses the ability to maintain a functional posture. Unchecked torso rotation (possibly leaving the individual draped over an armrest of a wheelchair) can occur with little movement of the legs and pelvis. The individual is still in the seat, but certainly not in a functional position.

For this research, the functional limits of CP motion, which have also been employed in standing balance (22), were used rather than biomechanical limits. Thus, the limits of stability were measured not according to the area of the base of support, but rather by the maximum distance that the subject could voluntarily move his CP. The subject, seated with arms crossed, leaned as far forward as possible, up to the balance point. The limit of stability was defined as the maximum possible displacement of the CPS that the subject could maintain without use of the upper limbs. **Table 2** illustrates the difference between the biomechanical and functional limits in the sagittal plane for the SCI subjects who participated in this study. The biomechanical limit was the distance from the CPS location at rest to the front limit of lower torso contact with the wheelchair. For the controls, the limit of functional CPS excursion was taken to be the front edge of the seat, (the biomechanical limit), since the functional limit was not reached even with the trunk flexed onto the legs.

Table 2.
Biomechanical and functional stability limits in the sagittal plane for the spinal cord-injured subjects.

		Maximum Displacement of CPS (cm)	
		Biomechanical Limits	Functional Limits
Subject	SCI Level	$[\Delta CPS]_{max}$	$[\Delta CPS]_{max}$
DL	C5-6	68.1	2.4
MB	C5-6	64.0	3.4
KR	C6-7	66.3	5.4
TT	C7	64.3	3.3
JB	T2-3	67.6	5.7
JW	T4-5	66.3	5.0
SH	T7	65.5	2.6
AC	T9	64.3	14.1

The ΔCPS limits were used to normalize the CPS displacements recorded during testing. The variable CPS displacements were divided by the constant ΔCPS limits, $(\Delta CPS)_{max}$, to create a fraction-of-limit term, denoted FLCP. Theoretically, the closer the FLCP becomes to one, the less the adjusted margin of stability. In reality, values greater than one were seen because the maximums were based on the limits of volitional movement of CP. The FLCP can be used to compare the stability of different individuals or the same individual being tested under different conditions. Equation 4 defines FLCP:

$$FLCP = \frac{(\Delta CPS)}{(\Delta CPS)_{max}} \quad [4]$$

Kinematic information was also recorded for each subject. Reflective markers placed on the snugly fitting T-shirts facilitated measurement of body segment motions. All angles were measured with respect to the vertical axis perpendicular to the platform. Markers were set external to the femur, the greater trochanter, the iliac crest, the lateral extent of the 10th rib, the axilla, the lateral base of the neck, and the ear. These locations enabled estimation of rotation of the pelvis, lower torso, upper torso, neck, and head. **Figure 4** displays these markers. In this study, segment rotation was quantified in terms of the change in angle from the starting posture.

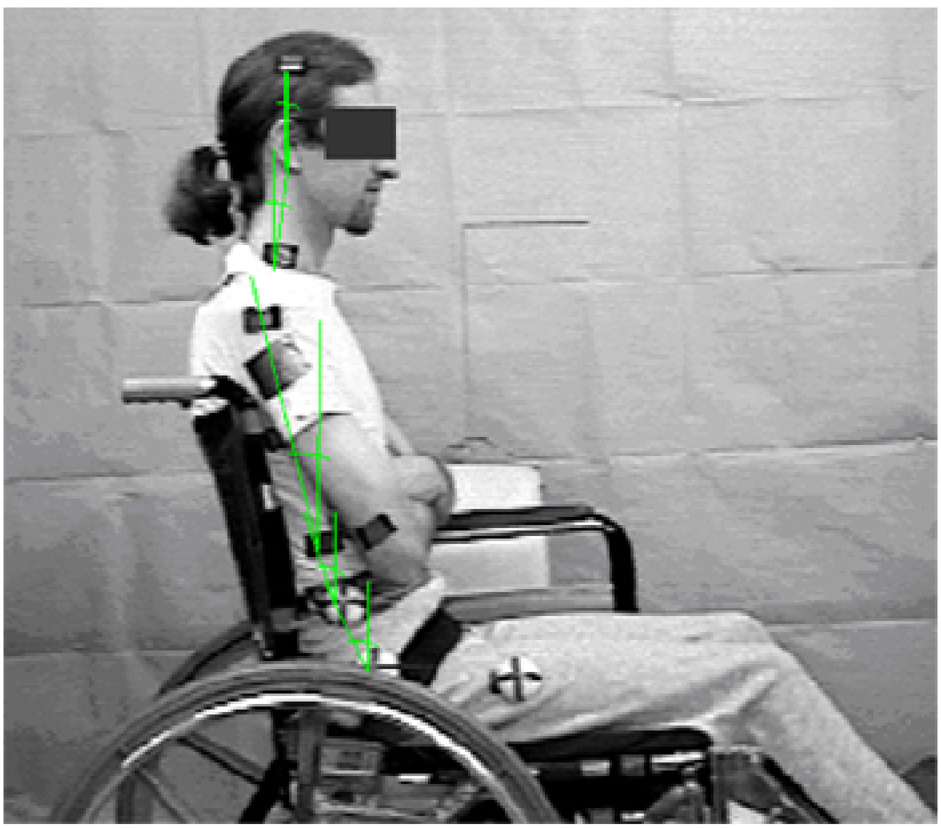

Figure 4.
Location of reflective markers in the sagittal plane for use in obtaining body segment angles from video.

For each trial, the video signal was digitized once every six frames for a sampling rate of 5 Hz. Marker position in the images was selected manually through the use of a pointer. The body segment angles were computed from the marker positions for each frame sampled. Of course, this implies that a straight-line approximation of the curvature of a number of vertebrae was employed in order to obtain the rotation angle for each body segment. The segment rotations were low-pass filtered at 2 Hz with a FIR filter to remove high-frequency noise introduced by the data processing. Precision of the angle measurements was approximately ±0.5°; however, confidence in the accuracy of the measurements was lower due to possible marker motion and difficulty in locating proper marker attachment points. Focusing on changes in segment angle from the initial posture, instead of absolute angle, lessened the impact of positioning errors.

Data Analysis

Each participant was to be subjected to each of the four Disturbance profiles. One subject, DL, did not experience the HS and HF profiles due to concerns about safety, as he had difficulty in stopping torso rotation even with his arms. Two trials were conducted for each experimental condition. If stability was maintained for one trial and lost for the other, a third trial was conducted and the two in agreement were analyzed. This occurred twice.

The Disturbance at which a subject lost balance was examined for possible impact of subject group. An ANOVA with repeated measures was performed using the statistical package SAS (SAS Institute, Cary, NC). The maximum Disturbance level (0.2g or 0.4g) for a given profile was assigned to those trials in which stability was maintained.

Peak FLCP values were determined for each trial. The main effects of subject group and input profile were assessed using an ANOVA with repeated measures. When a main effect was significant, multiple contrasts, involving comparisons of the least-squares means of the levels of the factor, were performed using SAS.

The kinematic data were statistically analyzed through examination of the difference between rotation of the lower and upper torsos. Rotation was measured as a change from the initial posture. Visual observations had suggested that the rotations might describe the postural control strategies used by different subject groups. The relative rotation (symbolized LT-UT) was analyzed in an ANOVA with repeated measures design. A peak (LT-UT) was determined for each trial. For cases in which the subject lost balance, the peak for the period from the start of the trial up to the point of instability was chosen.

RESULTS

Effects of Spinal Cord Injury on Seated Stability

Data for the three different subject groups were analyzed to examine the effects of SCI on postural control. All of the control subjects maintained stability for every trial. **Table 3** summarizes the stability results for the SCI subjects. Only one SCI subject, the one with the lowest lesion, remained stable for all four Disturbance profiles. The repeated measures ANOVA confirmed that subject category had a significant effect across all trials on the Disturbance level at instability onset ($p<0.002$). Multiple comparisons yielded a statistically higher absolute level for control as compared to SCI, and for PP as compared to TP subjects. The levels for each group were highly distinct ($p<0.0001$).

Table 3.
Disturbance levels at which subjects with disabilities first became unstable.

		Disturbance Level at Onset of Instability (g)			
	SCI	Disturbance Profile			
Subject	Level	LS	LF	HS	HF
DL	C5-6	-0.146	-0.120	*	*
		-0.048	-0.143	*	*
MB	C5-6	-0.143	-0.156	-0.071	-0.164
		-0.136	-0.162	-0.164	-0.262
KR	C6-7	-0.191	-0.167	-0.150	-0.126
		-0.173	-0.190	-0.174	-0.145
TT	C7	S	-0.187	-0.197	-0.191
		S	-0.190	-0.210	-0.182
JB	T2-3	S	S	-0.339	-0.394
		S	S	-0.313	-0.357
JW	T4-5	-0.165	-0.178	-0.171	-0.126
		-0.153	-0.156	-0.150	-0.173
SH	T7	-0.146	-0.187	-0.128	-0.375
		-0.159	-0.173	-0.197	-0.145
AC	T9	S	S	S	S
		S	S	S	S

S=stability maintained; *=datum from a trial was not available: entry treated as missing datum in statistical analysis; Maximum Disturbance levels: LS and LF-0.2 g; HS and HF=0.4 g.

The normalized CPS motion was also analyzed to provide a more complete description of stability throughout a test. A sample plot of FLCP for a trial in which the subject lost his balance is shown in **Figure 5.** Across Disturbance profiles, the amount of FLCP movement was distinct for each subject group category, being greatest in the TP group and least in the controls (**Table 4**).

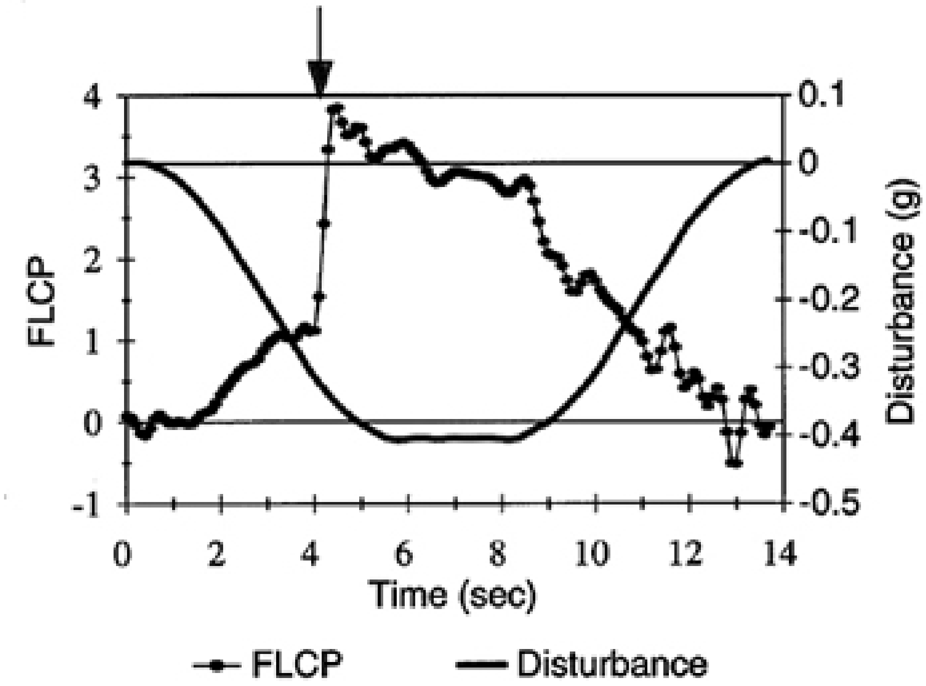

Figure 5.
FLCP curve for subject JB (T2-3) given the HS input. The arrow marks the point at which JB lost stability.

Table 4.
Results from analysis of the effect of subject group on the peak fraction-of-limit data. Least-square means and standard deviations of each subject category are shown.

Peak FLCP

Subject Group Category*				
TP Mean (SD)	**PP** Mean (SD)	**AB** Mean (SD)	**Multiple Contrasts+**	**P-Value**
5.00 (0.17)	4.02 (0.14)	0.47 (0.13)	TP>AB	0.0001
			TP>PP	0.0001
			PP>AB	0.0001

Output obtained from SAS®; *: Main effect of subject group was statistically significant ($p<0.05$); +: Statistically significant differences between group categories are shown along with the p-values; SD=standard deviation.

Subject group was also found to be significant in explaining changes in rotation of the lower torso with respect to the upper torso ($p<0.05$). The SCI subjects exhibited greater relative rotation than did the controls. The (LT-UT) values were statistically greater for PPs than for TPs. Rotation values and subject group categories with significant differences are displayed in **Table 5.**

Table 5.

Main effects of subject group for the ANOVA of peak relative rotation of the lower torso with respect to the upper torso.

Peak (LT-UT) (deg)				
Subject Group Category*				
TP Mean (SD)	**PP** Mean (SD)	**AB** Mean (SD)	**Multiple Contrasts+**	**P-Value**
2.0 (1.4)	12.2 (1.2)	-2.3 (1.1)	PP>AB	0.0001
			PP>TP	0.0001
			TP>AB	0.0189

*: Main effect of subject group category significant in ANOVA ($p<0.05$); +: Differences between subject group categories tested at $\alpha=0.05$ significance level; TP=tetraplegic; PP=paraplegic; AB=able-bodied; SD=standard error.

Effects of Disturbance Type on Stability

The main effect of the shape of the Disturbance on subject response was also analyzed. Synchronizing the loss of stability with the Disturbance revealed that in every case in which a subject fell, the fall occurred as the platform was being raised and, therefore, before the maximum Disturbance was attained.

Across all subjects, the main effect of the Disturbance profile on peak FLCP was statistically significant according to ANOVA results ($p<0.0001$), as shown in **Table 6.** The profiles that produced distinct amounts of motion are listed.

Table 6.

Results from analysis of the effect of Disturbance profile on the peak FLCP data. Least-square means and standard deviations of each subject category are shown.

Peak FLCP values

Disturbance Profile*

LS Mean (SD)	**LF** Mean (SD)	**HS** Mean (SD)	**HF** Mean (SD)	**Multiple Contrasts** +	**P-Value**
2.24 (0.16)	2.95 (0.17)	3.15 (0.17)	4.31 (0.18)	HF>LS	0.0001
				HF>LF	0.0001
				HF>HS	0.0001
				HS>LS	0.0002
				LF>LS	0.0026

*: Main effect of disturbance profile was significant in ANOVA ($p<0.0001$);
+: Statistically distinct types of Disturbance profile are enumerated under Multiple contrasts. SD=standard deviation.

Figure 6 provides an illustration of the impact of the rate of change of the Disturbance on the kinematic response. Body segment rotation is shown for the same subject subjected to profiles HS and HF, two inputs with the same peak Disturbance, but with different rates of change.

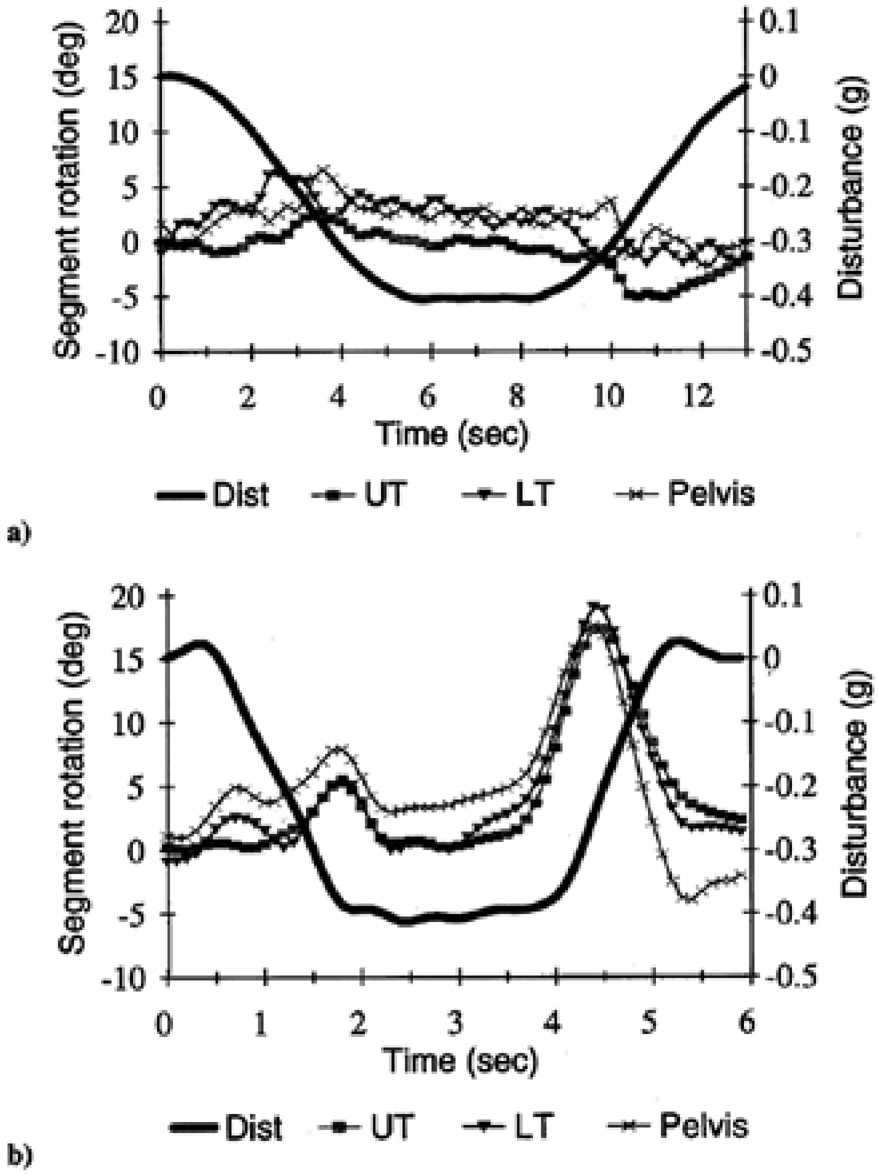

Figure 6.
Responses of control subject CH to two profiles with the same maximum Disturbance level, but with b) having a rate of Disturbance change three times that of a).

DISCUSSION

Overall, SCI had a very discernible effect on seated postural stability, as expected. **Table 3** shows that the majority of SCI subjects became unstable for all of the Disturbance inputs. None of the controls lost stability during any of the trials. Analysis of Disturbance level at instability onset verified that, as a group, TPs were indeed less stable than PPs. The significantly greater peak FLCP values in **Table 4** show that on average the TPs also fell relatively further when they did lose stability. However, it should be noted that injury level alone was not always a good predictor of stability, even though all subjects were active.

The SCI subjects seemed to attempt to compensate for diminished control of the pelvis and lower trunk by actively rotating the upper torso in a direction opposite that of the lower torso and platform. This movement reduces CPS displacement. Greater upper torso control in the PP group enabled greater relative rotation and, thus, greater success with this strategy (**Table 5**). In general, when instability occurred, forward rotation of the pelvis and lower torso preceded forward rotation of the upper torso. Certainly, the addition of chest belts could have prevented instability. However, the belts may be difficult to don, and they may also prevent desired torso motion. The results of this study suggest that restraints that limit pelvic and lower torso rotation, but not that of the upper torso, might provide sufficient stabilization in some cases.

The shape of the Disturbance profile also affected the subject response. While the number of subjects in this study is small, the results do clearly demonstrate that perturbations of a similar magnitude to those encountered during even normal driving conditions can pose significant challenges to the postural stability of individuals with SCI. The majority of the SCI subjects exhibited signs of instability below the 0.2g Disturbance level. This is in agreement with the results of Sprigle and Linden (24). These perturbations are below the levels for which the inertial-locking mechanisms for the lap and shoulder belts in automobiles are designed to engage (25). Hence, even the addition of a shoulder belt would not provide much support. Perhaps investigations should be made into the use of a modified locking mechanism for disabled travelers.

Of course, for the experiments in this study, the subjects were not allowed to use their upper limbs. Utilization of the arms would undoubtedly increase stability (24), but at the same time would reduce functional use of the arms for other tasks. Also, the opportunity for injury would increase. A common stabilizing technique in the tetraplegic population is to hook an arm around the push bar of the wheelchair, thereby assuming a potentially dangerous posture in the event of a crash.

It should be noted carefully that the tilt platform does not simulate the vehicle environment. The vehicle represents, potentially, a six-dimensional space of applied perturbations. The tilt platform examines operating points along a single axis of this space. However, for pure braking maneuvers, the primary perturbation is along this same axis. Also, the platform provides a more repeatable perturbation than the vehicle. A cursory examination of the correlation of results from the two different testing modalities was performed. The same wheelchair configuration used on the tilt platform was inserted into the wheelchair bay of a van. Three of the SCI subjects (MB, TT, and JW) participated in 0.2g and 0.4g controlled braking maneuvers. The stability outcomes matched those for the LF and HF Disturbance profiles on the tilt platform.

To concentrate on the subject response, the wheelchair was bolted to either the tilt platform or the vehicle floor. The use of securement systems and pneumatic wheelchair tires may allow translation and tipping of the chair with respect to the vehicle. This motion could further destabilize the rider. Current studies are using the tilt platform to examine the performance of securement systems.

While not an independent factor, the rate at which the Disturbance level changed seemed to impact the subject response. For example, a TP subject, TT, was stable for the slower rate of Disturbance change (LS), but fell during trials with the same peak Disturbance at a faster rate of change (LF). Peak FLCP across all subjects was greater for LF than LS trials and for HF than HS trials. The data in **Table 3** are misleading in regard to comparing the different Disturbance profiles in terms of the Disturbance level at instability onset. The Disturbance level changed more during the reaction time for the arm to make contact for the LF and HF trials than for LS and HS, so onset levels were often higher for LF and HF. Unlike this outcome measure, the FLCP signal contains information not just about the loss of stability, but also about the periods preceding and following. For example, the peak FLCP values for the LS trials for subjects TT and JB were greater than one, while those for the others who maintained their balance for the LS trials, AC and the controls, were less than 0.5. This helps to explain why TT and JB lost balance during the HS trials, while AC and the others remained stable.

Rate of change seemed to affect the transient response of even the controls. This was most easily seen by examining the kinematic data (**Figure 6**). Rotation of the upper and lower torsos and pelvis shows a transient rise for the HF profile, but not for HS. Significant torso rotation was also reported in vehicle testing for maneuvers with high levels of jerk (18,24). This large trunk rotation could cause instability. Thus, the rate at which the perturbation is changed may be an important factor to consider when testing seated balance, especially in individuals with SCI.

The results suggest future studies of disabled drivers would be of particular interest, leading one to surmise that disabled drivers may push on or pull against the hand controls during driving to help stabilize themselves. The addition of a rigid structure within arm's reach in front of a subject in a vehicle improved the stability of TP individuals in one study (24). It can be assumed that disabled drivers are using driving controls in a similar fashion to form a closed link. Determination of the magnitude and direction of the forces on the hand controls may help in the design of adaptive controls.

CONCLUSIONS

Examination of seated postural stability through the use of a tilt platform revealed that SCI subjects lost balance at perturbation levels seen during normal braking maneuvers. In fact, the majority of SCI subjects became unstable at levels below which inertial belts are designed to lock. Expectedly, even for the small sample sizes, as a group TPs were less stable than their PP counterparts. However, on an individual basis, injury level was not always a good predictor of stability.

While not tested independently, the rate of change of the perturbation seemed to influence response. This suggests that the rate should be controlled in experiments of this nature. Finally, normalized CP motion with respect to the seat proved a reliable, continuous index of stability. It should be especially useful in comparing trials for which balance is maintained.

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