Braking Electric-Powered Wheelchairs: Effect of Braking Method, Seatbelt, and Legrests

Rory A. Cooper, PhD, Michael J. Dvorznak, BS, Thomas J. O'Connor, MS, Michael L. Boninger, MD, Daniel K. Jones, PhD

ABSTRACT. Cooper RA, Dvorznak MJ, O'Connor TJ, Boninger ML, Jones DK. Braking electric-powered wheelchairs: effect of braking method, seatbelt, and legrests. Arch Phys Med Rehabil 1998;79:1244-9.

Objective: To examine the influence of three electric-powered wheelchair braking conditions and four wheelchair seating conditions on electric-powered wheelchair motion and Hybrid II test dummy motion. This study provides quantitative information related to assessing the safety of electric-powered wheelchair driving.

Design: Rehabilitation engineering comparison and ANSI/RESNA standards testing. Convenience sample of eight different electric-powered wheelchairs. Within-chair comparisons were conducted.

Intervention: Electric-powered wheelchairs were compared under three braking scenarios (joystick release, joystick reverse, power-off) and four seating conditions (seatbelt and legrests, seatbelt and no legrests, no seatbelt but legrests, no seatbelt and no legrests).

Setting: A rehabilitation engineering center.

Main Outcome Measures: The braking distance, braking time, and braking accelerations for electric-powered wheelchairs during three braking scenarios; trunk motion, head motion, and trunk angular acceleration during three braking scenarios and four seating conditions; and number of falls from the wheelchairs for three braking scenarios and four seating conditions.

Results: Significant differences ($p < .05$) were found in braking distance, braking time, and braking acceleration when comparing the joystick release and joystick reverse scenarios with the power-off scenario. The mean braking distance was shortest with the power-off braking scenario (.89m), whereas it was longest when the joystick was released (1.66m). Significant differences ($p < .05$) in head displacement and trunk angular displacement were observed among braking conditions and between seating conditions. There were also significant differences ($p = .0011$) among braking conditions for maximum trunk angular acceleration. The Hybrid II test dummy fell from the wheelchairs with highest frequency when there were no legrests and no seatbelt used.

Conclusion: The results of this study indicate that use of a seatbelt when driving an electric-powered wheelchair reduces the risk of falling from a wheelchair. Furthermore, the use of legrests can reduce the risk of injury to the wheelchair driver. This study shows that the most abrupt braking occurs when deactivating the power switch.

MANY USERS OF POWER wheelchairs have difficulty maintaining an unsupported upright seating position or a supported seating posture when the upper body is subjected to external forces.[1,2] Data collected by Unmat and Kirby[3] report that approximately 36,000 serious wheelchair accidents occur annually.[3] Between 1973 and 1987, there were 770 wheelchair-related deaths reported to the United States Consumer Product Safety Commission (USCPSC), 68.5% of which were attributed to falls and tips.[4] Of the 2,066 nonfatal accidents reported between 1986 and 1990 to the USCPSC, falls and tips were the cause 73.2% of the time.[3] It is important that these numbers be reduced and that the use of electric-powered wheelchairs (EPWs) be made safer. This can only be accomplished when the causes of the falls are known.[5] Many safety devices such as seatbelts, anti-tip casters, and legrests are not used or may be improperly adjusted. Many EPWs are manufactured without seatbelts, and it is common for users to remove the legrests for a tighter turning radius.

Improvements in motors, transmissions, and controllers have made EPW increasingly more powerful and versatile.[2,6,7] Modern EPWs are capable of speeds of up to about 3 m/sec, and rapid, controlled braking is needed if the driver suddenly encounters an obstacle.[8,9] Applying the brakes rapidly exposes the driver to potentially dangerous conditions, such as tipping over or being thrown from the wheelchair. The most common device for controlling an EPW is a hand-actuated joystick.[1,2,6] Once an EPW is moving, there are three common ways to brake it. The driver may release the joystick, allowing the electronic controller to bring the EPW to a stop. For more rapid braking, the driver may pull backward on the joystick; full-reverse of the joystick provides maximum controlled braking. The third alternative is to hit the power switch, which removes power from the motor and in many cases applies electromechanical parking brakes.

Seatbelts are available on most EPWs to secure and position the driver in the chair. As in automobiles, wheelchair seatbelts are easily removed to allow the occupant to transfer to or from the chair. Footrests are also used on EPWs to maintain posture and to keep the feet from dragging on the ground. Some people remove the footrests to make the chairs more maneuverable indoors.

There is an upward trend in the number of serious accidents attributed to wheelchair use.[3] In recent years, most EPW driver

From Human Engineering Research Laboratories: Department of Rehabilitation Science and Technology, School of Health and Rehabilitation Sciences, University of Pittsburgh (Drs. Cooper, Dvorznak, O'Connor, Boninger); the Division of Physical Medicine and Rehabilitation, Department of Orthopaedic Surgery, University of Pittsburgh Medical Center Systems (Drs. Cooper, Dvorznak, O'Connor, Boninger, Jones); and the Research Service, VA Pittsburgh Healthcare System (Drs. Cooper, Dvorznak, O'Connor, Boninger, Jones), Pittsburgh, PA.

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Reprint requests to Rory A. Cooper, PhD, Human Engineering Research Laboratories (151-R1), VA Pittsburgh Healthcare System & University of Pittsburgh, 7180 Highland Drive, Pittsburgh, PA 15206.

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injuries have been the result of falling out of the chair or tipping.[3-5] Investigators have indicated that the stability of wheelchairs is an area that warrants further research.[10-16] The International Standards Organization has begun to define tests to evaluate EPW braking.[2,8,9] One test addresses braking EPWs on horizontal, incline, and decline surfaces; the emphasis is on the distance required for braking. Another test pertains to deceleration during braking by cutting the power or full-reverse of the joystick. These standards are still under development, and further studies are needed for their practical implementation. It is anticipated that the standards will help to improve the safety of EPWs.[2,9]

In 1993, the National Rehabilitation Hospital measured the braking performance of 10 EPWs from seven manufacturers.[8] That study presented performance characteristics based on the American National Standards Institute and Rehabilitation Engineering and Assistive Technology Society of North America (ANSI/RESNA) standards, including parking brake and running brake tests, with an emphasis on maximum speed and braking distance, did not consider occupant motion. Gaal et al[16] assessed the forward dynamic stability of manual wheelchairs, using a test dummy, rolling down a hill and hitting a low obstacle. Different obstacles were investigated to see when the chair would tip over or when the dummy would fall from the chair. They found that seatbelts did not have a significant effect on holding the dummy in the seat. They also found that the height of the seat affected dynamic stability; higher seats tended to tip over more readily. Fast and colleagues[15] studied the effects of two restraining belts (lap belt and a four-point belt) on an instrumented dummy during three test scenarios: manual wheelchair hitting straight into a curb; manual wheelchair falling straight off a curb; manual wheelchair falling diagonally from a curb.[15] They monitored forces, moments, and accelerations via sensors placed in the head, spine, and limbs of a 50th percentile Hybrid III test dummy. Their results suggest that a lap belt reduced the risk of injury.

Research is needed to better understand the effects of EPW driving on the EPW user. Data are lacking on safe acceleration rates, the effectiveness of safety systems, and the influence of wheelchair and seating setup on safety. The purpose of this study was to examine the influence of three EPW braking conditions and four wheelchair seating conditions on EPW motion and Hybrid II test dummy (HTD) motion. The study tested the following hypotheses: (1) changing braking conditions (ie, joystick release, joystick reverse, and power-off) would lead to different braking distances, braking times, and braking accelerations; (2) selected seating conditions (ie, with and without legrests, and with and without a seatbelt) in conjunction with changing braking conditions would yield different test dummy motions and accelerations.

METHODS

Test Dummy

A 50th percentile anthropometric HTD[a] was used to simulate a person driving a power wheelchair.[17] This type of HTD has been used in previous studies of wheelchair stability.[12,13] The HTD was outfitted with black clothing. Passive reflective markers were glued onto areas representing nine anatomic positions on the left side of the HTD's body: back of the ear, shoulder, elbow, wrist, joint between hand and index finger, hip joint, knee joint, ankle joint, and fifth toe. The HTD was tuned just before testing began. The same HTD was used for all tests. It was positioned in the test wheelchair before each trial according to the procedure described in ISO 7176-08 draft standard.[2]

We performed a simple test to determine whether the HTD's response would be similar to that of an individual with a spinal cord injury. A single subject with T8 paraplegia caused by traumatic spinal cord injury served as a subject for this test. The subject was 180cm tall with a mass of approximately 60kg. A harness was placed around the shoulders of the subject, and a force gauge was attached to a D-ring at the center of the chest. Each of three test technicians pulled on a force gauge until the subject fell forward onto his knees. The maximum force was recorded. The same procedure was repeated for the HTD. Analysis of variance with Scheffe's post-hoc analysis showed that there were no significant interrater differences and no significant differences in maximum force required to cause forward folding for the person with a spinal cord injury or the HTD. This gives some indication that the HTD's responses may be similar to those of an actual wheelchair user.

Test Wheelchairs

Data were collected from eight different wheelchairs (table 1[b-e]). The wheelchairs were selected based on availability at our medical center and research laboratories. Major manufacturers were also contacted to provide products to increase diversification. All wheelchairs had similar seating dimensions (eg, seat depth, seat width, backrest height) and were equipped by the manufacturer with digital controllers. The factory default settings for each controller on each wheelchair were used throughout all testing. All wheelchair batteries were fully charged before testing. Before testing, the caster and drive wheel tires of each wheelchair were inflated to manufacturer specifications, if applicable.

All highly reflective areas (eg, chromed parts, nuts, bolts) on each EPW were blackened to facilitate videotaping. Reflective markers were placed on six areas of the left side of each wheelchair: top corner of the backrest, corner of the backrest and seat frame, front end of the seat corner, rear wheel sidewall, rear wheel hub, and the hub of the front caster (fig 1). When legrests were used, there was a reflective marker placed on the left side in the center of the foot plate edge. The left armrest was removed from each wheelchair during videotaping to expose the anatomic markers on the HTD.

Experimental Protocol

Test protocol included three braking scenarios: power-off (ie, the power switch was deactivated), joystick release, and full reverse of the joystick. There were four wheelchair seating conditions tested: seatbelt and footrest, seatbelt and no footrest, no seatbelt with the footrest, and no seatbelt and no footrest.

Table 1: Models and Manufacturers of Wheelchairs Tested

Wheelchair Manufacturer/Models
Everest & Jennings[b]
Tempest
Hoveround[c]
MPV4
Technique
Levo[d]
LC
Sunrise Medical Incorporated[e]
Quickie P100
Quickie P110
Quickie P200
Quickie P300

Complete names and addresses of the four wheelchair manufacturers are listed in the Suppliers list after References.

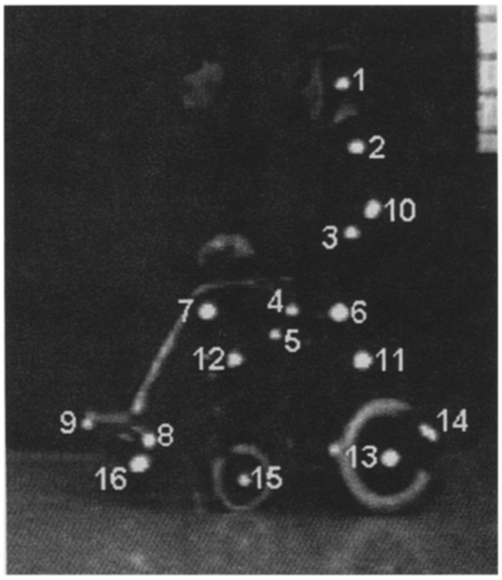

Fig 1. Image of markers placed on HTD during testing.

The order of the testing was randomized. The test operator dressed entirely in black clothing. Each test wheelchair was operated at maximum speed from the right side, as viewed from the back. The wheelchairs were driven 10 meters before reaching the filming area to obtain maximum speed, which was determined for using the methods described in ISO 7176-03.[2,9] A line on the test surface was used to ensure that the wheelchair was being driven straight. A braking line was marked on the floor, perpendicular to the line of motion, where the operator initiated the braking condition when the front wheels crossed the braking line. Two observers determined when braking was initiated. Trials were repeated until the observers agreed that braking had initiated at the line for a total of three trials. The videotape later corroborated their judgment. Data were collected for trials when the observers were in agreement, and all markers were visible on the videotape throughout the entire braking distance. A total of 270 trials were analyzed (eight chairs, three braking scenarios, four seating conditions, and three trials of each) were included in this study. The results from all 270 of the trials were included in the analysis. The Hoveround MPV4 has non-removable legrests; hence, only three seating conditions were tested with this chair.

Motion Analysis

Black curtains were used as a backdrop in the filming to ensure a high contrast between the reflective markers and the environment. The filming was done in two locations: the test laboratory and a recreational room, both of which had identical tile surfaces. After the testing started, it was discovered that two of the wheelchairs needed to be tested in the recreation room because they needed more space to complete the testing. The camera was 5 and 8 meters from the line of action during the laboratory and the recreation room testing, respectively. A halogen spotlight illuminated the filming areas. A Panasonic Digital 5100 camera and a Panasonic AG-7400 VCR were used to film and record the testing. Data were collected at 30 frames/sec with a shutter speed of 1/500 second. The videotape was digitized at 60 fields per second and conditioned with a low-pass Butterworth filter at a cutoff frequency of 6Hz.[f] The motion analysis system was calibrated before each data collection session.

Statistical Analysis

Repeated trials of the conditions were compared using analysis of variance (ANOVA).[g] Statistical differences in maximum speed, speed just before braking, braking methods, and seating conditions were investigated. A significance level of $p < .05$ was set *a priori*. Scheffe's post hoc analysis was used when ANOVA showed the existence of significant differences.

RESULTS

The maximum speed of each wheelchair just before braking was calculated for each of three trials for each condition (table 2). ANOVA showed significant differences ($p < .05$) between maximum speeds among the eight EPWs tested.

The results from the tests with the three braking conditions for the four seating conditions are presented in table 3. A two-factor ANOVA showed that there were significant differences ($p = .001$) in braking distance among the three braking conditions. However, there were no significant differences in braking distance across the four seating conditions. This analysis suggests that the change in center of gravity due to the different dynamic seating conditions did not significantly affect braking. Furthermore, there were no significant interactions. Scheffe's post hoc analysis with the EPW showed significant differences ($p < .05$) in braking distance among each of the braking conditions. The mean braking distance was shortest with the power off (.89m), whereas it was longest when the joystick was released (1.66m). A two-factor ANOVA of braking time also showed a significant difference ($p = .001$) in braking

Table 2: Results From Scheffe Analysis of Maximum Speed (m/sec) Just Before Braking for Test Chairs

	Wheelchair Model							
	Tempest 2.06 ± .08 (n = 36)	MPV4 1.97 ± .08 (n = 18)	Technique 2.03 ± .09 (n = 36)	Levo LC 1.79 ± .06 (n = 36)	P100 1.78 ± .12 (n = 36)	P110 1.93 ± .07 (n = 36)	P200 3.22 ± .12 (n = 36)	P300 2.91 ± .13 (n = 36)
P300	*	*	*	*	*	*	*	
P200	*	*	*	*	*	*		
P110	*		*	*	*			
P100	*	*	*					
Levo LC	*	*	*					
Technique								
MPV4								
Tempest								

The mean ± one standard deviation of the maximum speed for the number of trials of each wheelchair is listed in the column heading below the wheelchair model name.
*Significant difference, $p < .05$.

Table 3: Critical Power Wheelchair Braking Parameters for Selected Seating Conditions

	Braking Condition								
	Joystick Release			Joystick Reverse			Power-Off		
Seating Condition	Braking Distance	Braking Time	Braking Acceleration	Braking Distance	Braking Time	Braking Acceleration	Braking Distance	Braking Time	Braking Acceleration
Legrests & seatbelt	1.63 (.44)	1.56 (.18)	−3.18 (1.14)	1.33 (.42)	1.10 (.19)	−3.55 (1.07)	.86 (.50)	.79 (.27)	−5.62 (2.30)
Legrests & no seatbelt	1.74 (.58)	1.56 (.09)	−3.06 (.88)	1.34 (.44)	1.10 (.20)	−3.46 (1.03)	.90 (.58)	.82 (.34)	−5.89 (3.18)
Seatbelt and no legrests	1.59 (.46)	1.52 (.19)	−3.04 (.82)	1.31 (.55)	1.09 (.22)	−3.55 (.98)	.94 (.56)	.84 (.23)	−6.29 (4.15)
No seatbelt or legrests	1.64 (.55)	1.54 (.28)	−3.30 (1.08)	1.31 (.45)	1.09 (.20)	−4.11 (2.66)	.85 (.45)	.76 (.24)	−5.96 (3.41)
Combined data	1.66 (.52)	1.56 (.18)	−3.16 (1.30)	1.32 (.46)	1.10 (.20)	−3.66 (1.56)	.89 (.52)	.80 (.27)	−5.93 (3.25)

The mean (± one standard deviation) are reported. There are 24 trials for each cell with legrests, and 21 trials in each cell without legrests. This is because the MPV4 has nonremovable legrests. (Braking distance is in meters; braking time is in seconds; and braking acceleration is in m/sec^2.)

time among the three braking conditions, but no significant differences ($p > .05$) due to the four seating conditions. There were no significant interactions for the braking time analyses. From the Scheffe's analysis, all three braking conditions resulted in significantly different braking times. Our analysis also showed significant differences in braking acceleration. The two-factor ANOVA of braking acceleration suggests that there are significant differences ($p = .001$) among the three braking conditions, but that there were no significant differences ($p > .05$) between the four seating conditions. Again, there were no significant interactions. A Scheffe's post hoc analysis of the braking acceleration data showed significant differences ($p < .05$) for the joystick reverse and joystick release when compared with the power-off condition, but no differences were found between joystick reverse and joystick release. A simple ANOVA showed that there were no significant differences ($p > .05$) between the speeds just before braking among the three braking conditions.

The testing with the four seating conditions showed several significant differences in the dummy motion. The results from the braking tests with the selected seating conditions are presented in table 4. These data include test trials in which the HTD fell from the chair. All of the movement recorded was in a single direction (ie, there was no rebound). A two-factor ANOVA showed that there were significant differences ($p < .05$) in head displacement among braking conditions and between seating conditions. Head displacement was the distance moved by the head of the HTD from the start of braking until the HTD came to a complete stop. There was also a significant interaction ($p = .006$) for head displacement between braking conditions and the presence of the legrests. Scheffe post hoc analyses of the head displacement data showed significant differences ($p < .05$) between braking by turning the power off and joystick reverse or joystick release. The Scheffe post hoc analysis also showed significant differences ($p < .05$) in head displacement when the legrests were in place versus when they were removed, and when a seatbelt was used versus without a seatbelt.

We also examined the effects of braking and seating conditions on trunk angular displacement (ie, range of trunk motion from just before braking to when the HTD came to a complete stop). A two-factor ANOVA showed significant differences in trunk angular displacement among braking conditions ($p = .001$) and seating conditions ($p = .001$). There were no significant interactions for the trunk angular displacement data. Scheffe's post hoc analysis of the trunk angular displacement data showed that there were significant differences ($p < .05$) between braking by releasing the joystick and braking by reversing the joystick or turning the power off. Furthermore, significant differences were seen in the trunk angular displacement data when the legrests were in place versus when removed, and for the use of a seatbelt versus braking without a seatbelt.

Recordings of the maximum trunk angular acceleration while braking for the selected seating and braking conditions were examined. A two-factor ANOVA only showed a significant difference ($p = .0011$) among braking conditions for maximum trunk angular acceleration. Scheffe's post hoc analysis revealed that there were significant differences ($p < .05$) in maximum trunk acceleration among braking conditions. The maximum trunk acceleration was significantly higher for the power-off braking condition than either the joystick release or joystick reverse condition.

In some cases, the HTD was ejected from the wheelchair because of the rapid acceleration during braking. We recorded the number of times during each of the three trials for each seating and braking condition (ie, each wheelchair was tested a total of 36 times, except for the MPV4 that has nonremovable legrests) that the HTD fell from the wheelchair (table 5). The four EPWs with the highest maximum speed accounted for 86% of the falls. The HTD fell from the EPW most during the power-off braking condition. The HTD was positioned in the test wheelchair with legrests and seatbelt removed seating

Table 4: HTD Motion for Common Braking Conditions and Selected Seating Conditions

	Braking Condition								
	Joystick Release			Joystick Reverse			Power-Off		
Seating Condition	Head Distance	Trunk Rotation	Trunk Angular Acceleration	Head Distance	Trunk Rotation	Trunk Angular Acceleration	Head Distance	Trunk Rotation	Trunk Angular Acceleration
Legrests and seatbelt	.48 (.12)	45.9 (12.3)	746.3 (577.0)	.52 (.11)	51.4 (13.9)	1318.4 (307.0)	.59 (.14)	55.0 (12.5)	1585.8 (1300.8)
Legrests but no seat belt	.54 (.13)	53.7 (13.4)	767.0 (372.6)	.57 (.09)	56.3 (11.8)	934.1 (404.6)	.94 (.64)	63.2 (11.4)	1522.7 (1009.7)
Seatbelt but no legrests	.57 (.15)	55.2 (16.2)	957.0 (1114.0)	.60 (.13)	60.9 (13.1)	836.2 (347.3)	.66 (.11)	64.3 (11.9)	1476.5 (905.8)
No seatbelt or legrests	.71 (.30)	54.7 (16.6)	1404.9 (1729.6)	.75 (.22)	69.5 (10.7)	1081.6 (362.0)	1.11 (.60)	73.2 (11.4)	1883.3 (989.9)

The mean (± one standard deviation) are reported. There are 24 trials for each cell with legrests, and 21 trials in each cell without legrests. This is because the MPV4 has nonremovable legrests. (Head distance is in meters; trunk rotation is in degrees; and trunk angular acceleration is in degrees/sec^2.)

Table 5: Trials When HTD Fell From the Electric-Powered Wheelchairs While Braking

Braking Condition	Seating Condition			
	Legrests and Seatbelt	Legrests But No Seatbelt	Seatbelt But No Legrests	No Legrests or Seatbelt
Joystick release	0	0	0	4
Joystick reverse	0	0	0	3
Power-off	0	6	0	9
Total (falls/trials)	0/72	6/72	0/63	16/63

condition most often when it fell from the EPW during braking. Figure 2 shows the HTD kinematics for a single wheelchair during the power-off braking condition without legrests and without a seatbelt.

DISCUSSION

The EPWs used in this study have varying purposes. Some are targeted toward indoor use with light outdoor use, whereas others are designed for active indoor/outdoor use. The differences in the intended use of the EPW accounts for the maximum speeds recorded. During this study, the Quickie P200 recorded the highest maximum speed at nearly twice the maximum speed of the Quickie P100. These results are not surprising because these two products are designed for different use profiles. All of the EPWs use microprocessor-based controllers, and maximum speed is a variable that can be tuned for the individual. Driving at higher speeds may place wheelchair users at greater risk for injury. The HTD tended to fall from the wheelchair with greater frequency from the faster wheelchairs when a seat-belt was not used. The results show that the four fastest EPWs (Tempest, Technique, P200, P300[e]) account for a significantly higher percentage (86%) of the falls than the four slowest wheelchairs, despite the fact that no falls were recorded for the HTD while testing the Tempest. Candidates and their clinical team must be aware of the potential for greater risk of injury and driving skill that are associated with high-speed EPW mobility. Our results show that this risk can be mitigated through the use of a properly adjusted seatbelt and correctly adjusted legrests.

The method of braking affects the braking distance, braking time, and braking acceleration. Our results indicate that the braking distance and braking time are shortest when the power is turned off; braking acceleration is greatest during this braking method. Releasing the joystick resulted in the longest braking distance and slowest braking time; joystick release also provided the lowest braking acceleration. When the electric power is removed from the EPW, the most common mode of braking is for the EPW controller to activate the electromechanical parking brakes. The electromechanical parking brakes are spring activated and attempt to bring the wheel/motor to an abrupt stop. This braking condition resulted in the highest number of HTD falls. Inadvertant deactivation of the power switch may be an important contributing factor to EPW-related falls and tips. The use of a soft power-off switch or controlled operation of the electromechanical brakes may reduce the number of accidents associated with unintentional braking due to deactivating the power.[2] A simple switch guard or careful placement of the power switch could also minimize unintentional deactivation of power. A soft power-off switch would send a signal to the power wheelchair controller that the user wishes to turn the power off. The power wheelchair controller receives data about speed and heading. Based on these data, the power wheelchair controller could begin turning the power off when it is safe based on the wheelchair's dynamics. Another solution would be to design the wheelchair so that when the power was deactivated, the electromechanical brakes would chatter (ie, switch on and off) bringing the EPW to a rapid controlled stop much in the same way that automatic braking systems do on automobiles. It is important for EPW users and clinicians who prescribe and assess EPWs to be aware of the rate at which they brake when the power is removed. Releasing the joystick may be the best way for people with poor trunk control to brake rapidly.

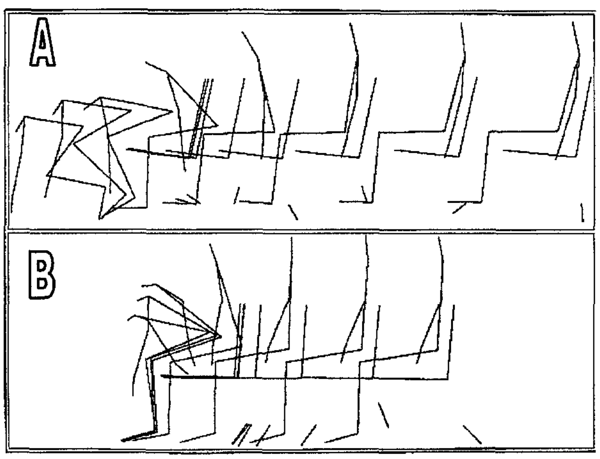

Fig 2. Stick figure plot from digitized image of markers during sudden braking. (A) Kinematics of the HTD when it falls from an EPW. (B) Kinematics when the dummy remains in the EPW.

Wheelchair legrests are used for positioning and to keep the user's feet from dragging.[1] Some people, especially people with lower limb function, remove the legrests to provide greater mobility indoors. Without legrests, the overall length of the wheelchair is decreased, and hence the turning space required is decreased. Our results show that removing the legrests without using a seatbelt increases the likelihood of falling from the EPW. Furthermore, the HTD experienced greater trunk rotation and trunk acceleration, which could lead to injury when the legrests were removed. Injury may result from striking a part of the wheelchair (eg, armrest) or further loss of control by moving the joystick. The HTD never fell when a seatbelt was used without a legrest, indicating that using a seatbelt mitigates the risk of falling when legrests are not used. Our results also showed that HTD had less of a tendency to fall from or experience high trunk range of motion with the slower wheelchairs. This suggests that lowering the speed while driving without legrests may also decrease risk of injury.

There are some limitations to this study. An individual would normally drive an EPW with their arms on the armrests. This was not possible during our testing without obscuring the camera's view of the markers. In some cases, a fall may have been prevented if the left armrest had been in place. During some trials, the HTD came in contact with the joystick because of the forces imposed during braking. These trials were discarded because the EPW would begin to accelerate into a turn. During the testing, the arms of the HTD were held at its side to prevent interaction with the joystick and provide freedom of movement for the trunk. This may have exaggerated trunk motion.

Future studies should examine the effect of varying the speed at the initiation of braking for a given wheelchair on both braking dynamics and HTD kinetics. Studies also need to be done on pediatric products using a smaller HTD.[18,19] We investigated the limited case of braking scenarios with selected seating conditions. It is also important to examine the dynamics of the electric-powered wheelchair and HTD when negotiating obstacles that are encountered daily.

CONCLUSIONS

The results of this study indicate that use of a seatbelt when driving an EPW reduces the risk of falling from a wheelchair. Furthermore, properly adjusted legrests can reduce the risk of injury to the wheelchair driver. These conclusions correspond with the findings of Fast and associates[15] in their testing of manual wheelchairs. Our results indicate that severe accelerations may be experienced by the occupant and wheelchair when braking by deactivating the power switch. The results of this study support the need for standards to ensure the safety of EPW drivers and to assist in the assessment of potential of EPWs. The desire of many drivers of EPWs for high speed must be considered in the development of such standards.[20]

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Suppliers

a. First Technology Safety Systems, Inc., 47460 Galleon Drive, Plymouth, MI 48170.
b. Everest & Jennings, 4203 Earth City Expressway, Earth City, MO 63045.
c. Hoveround Corporation, 2151 Whitfield Industrial Way, Sarasota FL 34243.
d. LEVO AG Dottikon, Bleicheweg 5, CH-5605 Dottikon, Switzerland.
e. Sunrise Medical, 2842 Business Park Avenue, Fesno, CA 93727-1328.
f. Peak Performance Technologies, Inc., 7388 South Revere Parkway, Suite 603, Englewood, CO 80112.
g. Mathworks, Inc., 24 Prime Park Way, Natick, MA 01760-1500.