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ORIGINAL REPORT

INFLUENCE OF HAND-RIM WHEELCHAIRS WITH REAR SUSPENSION ON SEAT FORCES AND HEAD ACCELERATION DURING CURB DESCENT LANDINGS

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Objective: **Shocks and vibrations experienced while using a hand-rim wheelchair can contribute to discomfort, fatigue and injury. The aim of this study was to compare the seat forces and head accelerations experienced by manual wheelchair users during independent curb descent landings in a standard and 3 suspension-type rigid-frame wheelchairs.**
Design: **Experimental: repeated measures analysis of variance.**
Participants: **Eight men with paraplegia due to spinal cord injury.**
Methods: **Participants performed independently-controlled curb descent maneuvers with 4 wheelchairs. The seat force and head accelerations were compared across wheelchairs.**
Results: **The suspension-type wheelchairs decreased the seat force and head accelerations by significantly ($p<0.05$) extending the force rise time. Also, the seat force and head accelerations were inversely related to the seat force at initial contact. The monoshock-based suspension wheelchairs showed the least seat force and longest force rise time.**
Conclusion: **Suspension systems result in softer landings by attenuating the magnitude and time duration of the force and reducing head accelerations. Hand-rim wheelchair users can also soften landings by utilizing a "pull-up" strategy that reduces the force and head accelerations. Softer landings can contribute to improved ride quality.**

Key words: spinal cord injury, wheelchair, biomechanics, shock and vibration, suspension.

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INTRODUCTION

Almost 11,000 new spinal cord injury (SCI) survivors are added each year to the total population of approximately 256,000 people now living with SCI in the USA (1). A substantial proportion of this population relies on the use of hand-rim wheelchairs as a daily means of ambulation (2). Over the past 30 years there have been many improvements in the design and function of wheelchairs and seating systems that are better fitting and more functional (3). While individuals with SCI are leading longer and more active lives, pain and discomfort during wheelchair riding has been a growing concern. In particular, exposure to whole-body vibration exceeding the standards set for industrial occupations (4, 5) has been documented during wheelchair use (6–9). Exposure to shock (infrequent high loads) and vibration (low-magnitude repeated loads) has been linked to muscle fatigue (10), back injury (11, 12) and neck pain (13). Consequently, shock and vibration experienced during daily wheelchair riding can decrease an individual's comfort (14), increase their rate of fatigue (6) and limit their functional activity and community participation (15).

To improve ride quality, manufacturers have developed innovative frames, seating, and suspension systems designed to reduce the shock and vibration during hand-rim wheelchair use. There are different frame types (e.g. rigid and folding) using different materials (e.g. aluminum, titanium, etc.) as well as components of suspension systems (e.g. rear wheels, front wheels) available commercially. Moreover, there are different rear wheel suspension elements, including independent suspensions using coil springs attached to the wheelchair frame, monoshock-based suspensions supporting the wheelchair seat frame, and polymer-based suspensions placed underneath the hubs. To examine the fatigue life of manual wheelchair frames, curb drop testing using a test dummy was included in the American National Standard Institute/Rehabilitation Engineering and Assistive Technology Society of North America (ANSI/RESNA) wheelchair fatigue testing standards (16–18). These evaluations provided vital information about the durability and cost-effectiveness of these suspension (and non-suspension) wheelchairs.

Managing a curb is one of the most important skills for evaluating the effectiveness and safety of manual wheelchairs in the user's own environment (19, 20). To successfully perform the curb descent maneuver, the wheelchair users must balance themselves and the wheelchair using only the rear wheels (i.e. a wheelie). During curb descent landings, large reaction forces are applied at the wheel-ground interface. These reaction forces are transmitted to the wheels and frame and are experienced by the user through contact with the seat and cushion. Consequently, users are exposed to large loads that may lead to falls resulting in serious injuries (21) or that may contribute to the development of pain, discomfort, and injury to the back and neck (13). Wheelchair design and components

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that can effectively reduce the shock and vibration experienced during daily wheelchair use have the potential to improve ride quality and wheeled mobility function.

Assessment of human response to shocks and vibration is generally achieved using psychophysical techniques combining subjective measures (22–24) with stiffness and vibration characteristics (25). Time and frequency domain analysis are often used to quantify the seated human's accelerations (26, 27). These quantities are then related to subjective measures of discomfort (24), pain (12), and fatigue (10, 28, 29). A variety of measures has been proposed and used to assess whole-body vibration exposure (30). Most commonly used is the ISO-2631 (4), which sets guidelines for how to take the measurements and calculates exposure statistics derived from power spectral densities of accelerations taken from the seat cushion interface. Frequency weighting functions are then applied to the measured whole-body accelerations on the assumption that they represent the dependency of human response on vibration frequency (27). The ISO-2631 specifies that seated humans are most sensitive to those whole-body vibrations in the frequency range 4–12 Hz. While these analysis techniques provide valuable information regarding the level of exposure and response to vibration, the values are more specific to automobile occupants and industrial workers (e.g. truck drivers, machine operators), rather than active manual wheelchair users with SCI. In addition, the ISO-2631 and other standards assume that the subjects are exposed to a constant or near-constant level of vibration exposure over the time-frame of the test. It cannot distinguish vibration that contains mechanical shocks. In the case of wheelchair users, large but infrequent impulsive forces are experienced in addition to the constant level of vibrations during the course of the day. Therefore, the actual forces acting on the user at the seat interface and corresponding accelerations of the body through the head during realistic conditions (i.e. wheelchair curb descent) can provide objective information regarding the wheelchair user's response and their reactions to mechanical shocks.

To evaluate the whole-body vibration suppression performance of different hand-rim wheelchairs during actual use, Kwarciak et al. (31) quantified the wheelchair seat accelerations during curb descent landings with suspension-, rigid-, and folding-type frames from various heights (0.05, 0.10 and 0.15 m) with a hand-rim wheelchair user. They determined that suspension-type wheelchairs varied in the vibration suppression performance (via frequency domain analysis) as a function of suspension type and orientation of the suspension element at impact (31). To determine how the shocks and vibration can contribute to discomfort, fatigue, and onset of injury; and to find ways to prevent such injuries, the magnitude and direction of the reaction forces experienced by the user and their subsequent body responses during real-world activities require further evaluation.

The aim of this study was to determine the seat force and head acceleration experienced by manual wheelchair users performing curb descent landings with rigid and rear suspension-type frames. We hypothesized that users would experience less seat force and head acceleration when using suspensions compared with a rigid hand-rim wheelchair. We anticipated that the wheelchair frame and suspension design; and conditions at impact (instant of time of ground contact) would influence the magnitudes of the seat forces, rise time, and accelerations experienced by the user. This information could be used to develop an objective means to assess the ride quality of hand-rim wheelchairs. Ultimately, the findings from this work may lead to the development of interventions and advanced wheelchair frame designs that maximize activities and community participation (15) among manual wheelchair users with SCI.

METHODS

Participants

We invited 8 men with complete (American Spinal Injury Association A or B) paraplegia (T12) to volunteer. Participants had a mean age of 28.0 years (age range 23–35 years), average time since injury 10 years (range 8–15 years) and mean body mass 80 kg (range 67–113 kg). We recruited all participants from the outpatient services of Rancho Los Amigos National Rehabilitation Center (RLANRC), Downey, CA, USA. All participants reported using manual push-rim wheelchair propulsion as their only means of community mobility, including negotiating curbs. We excluded individuals from participation if they reported a history of back and shoulder pain that altered performance of daily function or required medical treatment. Prior to data collection, we asked all volunteers to read and sign an informed consent form that had been approved by the RLANRC Institutional Review Board. We performed all testing at the rehabilitation engineering department at RLANRC.

Instrumentation

We instrumented one standard rigid-frame (SR) (16-inch; 0.406 m) Quickie GPV II, Sunrise Medical, Longmont, CO, USA) (Fig. 1A) and 3 suspension-type (ShS, SpS and ElS) wheelchairs (Fig. 1B–D) with load cells. All rear suspension systems were mounted on a rigid type aluminum frame: independent spring-based suspension (SpS) (16-inch Colours Boing, Colours In Motion, Inc., Corona, CA, USA) (Fig. 1B), elastomer-based suspension (ElS) (16-inch Invacare A4, Invacare, Elyria, OH, USA) (Fig. 1C), monoshock suspension (ShS) (16-inch Quickie XTR, Sunrise Medical) (Fig. 1D). The ElS (elastomer-based) suspension utilizes elastomer disks placed between scissor assemblies that couple each wheel axle with the frame (Fig. 1C). The SpS (spring-based) suspension utilizes 2 metal springs that independently regulate compression of the A-arm assembly that couples the frame and axle (Fig. 1B). The ShS (monoshock-based) suspension utilizes a single Rock Shox® (SRAM, Chicago, IL, USA) mountain bike suspension system that couples the seat with the axle and lower frame section (Fig. 1D).

We fabricated a lightweight aluminum seat frame and strategically placed 7 load cells (MLP Series, Transducer Techniques, Temecula, CA, USA) below the seat frame and backrest to record the vertical (Z), fore-aft (X), and lateral (Y) reaction forces acting on the user (Fig. 2). Four load cells were oriented vertically to measure the seat vertical reaction forces, as well as to constrain the vertical translation, pitch, and roll of the seat. Two load cells were mounted horizontally, over the wheels, to measure horizontal (fore–aft) forces as well as constrain the backward and forward translation and rotation of the seat. One load cell was placed horizontally, along the axle, to measure side to side (lateral, Y) forces. All load cells were placed perpendicular to each other to prevent signal cross-contamination. The weights (excluding the cushion) of each instrumented wheelchairs were: SR = 15.6 kg, SpS = 19.3 kg, ShS = 18.4 kg, ElS = 16.8 kg.

We used accelerometers on the wheel hub, instrumented seat, and user to measure the attenuation of acceleration through the user-chair system. For each wheelchair, we mounted 1-axis accelerometer modules (range: ± 10 g, Model 2210, Silicon Designs, Issaquah, WA,

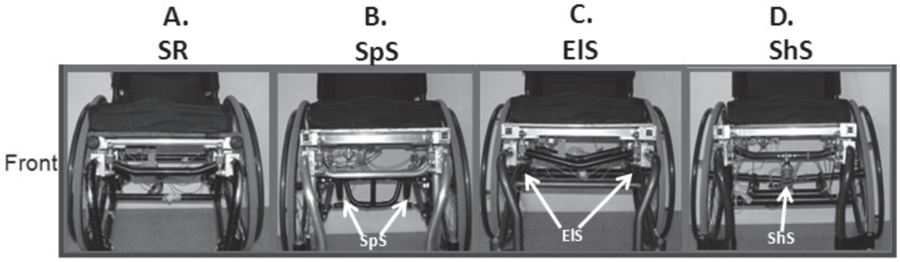

Fig. 1. One rigid-type frame (SR) and 3 suspension-type frame (SpS (spring), ElS (elastomer), ShS (monoshock)) wheelchairs were instrumented to measure seat reaction forces and accelerations. Arrows indicate the location of the suspension system.

USA) in the following locations: one on the posterior left corner of the instrumented seat frame and 2 mounted on the left hub of the wheel perpendicular to each other. We also mounted 2 perpendicularly-arranged accelerometers (range: ± 5 g) on a bicycle helmet worn by the test subject to measure vertical and horizontal head acceleration (Fig. 3). The accelerometer modules were chosen because of their rugged construction, low power consumption, fully calibrated, and simple 4-wire connection. The high-drive, low-impedance output allowed longer cable connection with minimal noise; suitable for evaluating under more dynamic conditions (e.g. traversing a course, curb descents, etc.). The portable data acquisition system we used to collect the data consisted of a personal computer (PC) laptop, 2 PCMCIA cards (DAQP-12H and DAQP-16, Quatech Inc., Hudson, OH, USA), a battery pack, and a voltage regulator. The DAQP-12H contains 8 differential channels with up to 1000 × gain and was used to collect load cell data. The DAQP-16 contains 8 differential channels with 1× gain and was used to collect accelerometer data. Seat reaction forces and acceleration data (2000 Hz) were simultaneously collected and synchronized by the data collection PC for each trial and saved onto a hard drive for subsequent processing and analysis.

Data collection procedures

Each subject performed 5 curb descent trials, using each instrumented wheelchair. All subjects used the same type of seat cushion (Nexus Spirit, ROHO, Belleville, IL, USA) and adjusted the height of the footrest and backrest to match their personal wheelchair. The subject's sitting posture and balance was confirmed by the research physical therapist with extensive experience in wheelchair seating and positioning. For all wheelchairs, the same make and model tires (Model 23-540, Kenda Tires, Reynoldsburg, OH, USA) inflated to 80 psi were used with the same make and model spokes wheels (SunRims CR20, Sun Metal Products, Warsaw, IN, USA). To accommodate to the test environment, we allowed the subjects to practice the curb descents landings 2–3 times in each wheelchair. Prior to collection of force and acceleration data, we performed a 5-sec baseline trial (while the subject was off the wheelchair) and a seated weight trial (while the subject sat on the wheelchair motionless looking forward). We then asked the subjects to perform at least 5 curb descents landings from a height of 0.10 m. The order of wheelchairs tested was chosen randomly for each subject. During all trials, a spotter remained near the subject to prevent falls. We allowed the subjects to rest for at least 5 min between using each wheelchair.

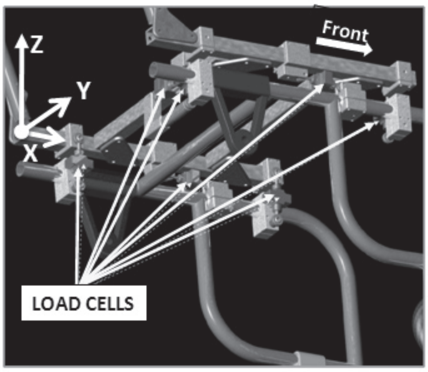

Fig. 2. Seven load cells were placed below the seat and backrest frame of all test wheelchairs for recording the vertical (Z), fore-aft (X), and lateral (Y) seat reaction forces during curb descent landings. The photograph show the wheelchair frame viewed from underneath.

Data processing and analysis

The digitized load cell and accelerometer data were filtered using a zero-phase fourth-order digital Butterworth low-pass filter (Matlab, Mathworks Inc., Natick, MA, USA) with 114 Hz cut-off frequency and then scaled to determine the seat reaction forces (in Newtons) and head vertical and horizontal accelerations (in g = 9.81 m/s^2). Seat force was expressed relative to the 3 axes (XYZ) of the wheelchair coordinate system (Fig. 2). Furthermore, we normalized the seat force to body weight (BW).

The recorded seat forces and head accelerations were repeatable and consistent between curb descents performed by the same user in the same wheelchair. Therefore, we used a time domain analysis by extracting key metrics from the resultant seat force (F) and vertical (Av) and horizontal (Ah) head accelerations (Fig. 3). The peak resultant force (Fmax), seat force at impact or initial ground contact (Fic), rise

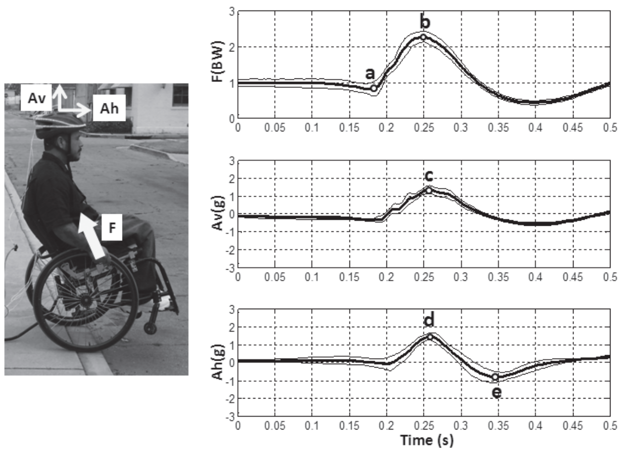

Fig. 3. Peak analysis of the resultant seat force (F), vertical head acceleration (Av) and horizontal head acceleration (Ah). The graph shows a mean (thick line) and standard deviation (thin line) from 5 curve descent trials by one subject with the standard rigid-frame wheelchair. From the seat force signals, the following points were identified: (a) Fic = Seat force at initial contact, (b) Fmax = Peak resultant force, (c) Avmax = peak vertical head acceleration, (d) Ahmax = peak forward horizontal head acceleration, and (e) Ahmin = peak backward horizontal head acceleration. In addition the following were calculated: dt = rise time is the time interval from Fic to Fmax, and dF = change in seat force is the difference between Fmax and Fic. The identified and calculated values are used to examine the shock absorbing performance of the wheelchair (and user).

time (dt) from Fic to Fmax, and change in seat force (dF = Fmax-Fic) were determined for each trial. We also identified the peak vertical head acceleration (Avmax), peak forward (positive) horizontal head acceleration (Ahmax), and peak backward (negative) horizontal head acceleration (Ahmin) (Fig. 3). We also considered the negative peak oscillations of the seat force and vertical head accelerations, but excluded them from our analysis. We defined landing as the time interval from Fic to Ahmin. We analyzed 5 trials performed in each wheelchair.

Mechanically, we can describe the behavior of the wheelchair-human interaction as represented by 2 energy-absorbing elements that is experiencing an input perturbation (seat force) which results in an output motion (head accelerations). Conceptually, this behavior can be described as equivalent to the impulse response of a damped system (27). If the magnitudes of the seat force (Fmax) and head accelerations (Av, Ah) are high, then the shock-absorbing performance of the chair (and user) would be poor. If the magnitudes of the seat force and head accelerations are small, then the shock-absorbing performance of the wheelchair plus the user is good. On the other hand, if the magnitude of the seat force is high while the head acceleration is low, then the shock absorbing performance of the user is good. In terms of how "soft" or "hard" is the landing performance of the wheelchair, we can examine the relationship between the change in seat force (dF) and rise time (dt). Large magnitude dF and short dt is indicative of a hard or impact type of landings, while low magnitude dF with long duration dt is indicative of a softer, less impact type of landings. In addition, we included a measure of the head horizontal accelerations (Ahmax and Ahmin). Ahmax is the amount of forward head acceleration in response to the applied seat force; akin to the forward motion that passengers feel when a car brakes. Ahmin is the subsequent backward head acceleration after the occurrence of Ahmax; akin to the whiplash effects. Overall, lower seat force, low head accelerations, and longer rise time would presumably translate to better ride quality, improved comfort, and reduce likelihood of musculoskeletal injuries.

In addition to the analysis of the magnitudes of force and accelerations, we used Fic (seat force at initial contact) as an indicator of the amount of "pull-up" exerted by the user in preparation for impact. The seat force at initial contact (Fic) can be seen as amount of contact (or net pressure) at the user/seat interface. A Fic of body weight (BW) indicated that the amount of contact between the user and wheelchair was equivalent to the seat fully supporting the weight of the user. A Fic less than BW indicated that the user reduced the amount of contact with the seat prior to impact. During curb descents, the only way to maintain a Fic near BW is to actively pull-up against the hand rims in preparation for landings. A Fic near BW describes greater pull-up, while those lower than BW describe less pull-up.

Statistics

The Shapiro-Wilk statistic determined that the seat force and acceleration metrics were normally distributed. Therefore, we utilized parametric statistics. To determine if the seat forces, vertical and horizontal head accelerations were significantly different within types of wheelchair frames, we applied repeated measures analyses of variance (ANOVA). We determined that the difference in the seat force and head accelerations were consistent within subjects. Consequently, we used the normalized group mean data for further comparison. We used a Bonferroni *post-hoc* analysis to identify significant differences between the individual wheelchair frame types. We calculated the Pearson's product moment correlation coefficient to determine the linear association between the seat force at impact and the seat force and head accelerations during landing. We analyzed all data using SPSS 12.0 software (SPSS, Inc., Chicago, IL, USA), setting the significance level to 0.05.

RESULTS

Wheelchair users showed lower seat force and head accelerations during landing in the suspension-type wheelchairs compared with a standard rigid-frame wheelchair. Rise time (dt) was significantly ($p < 0.05$) greater in the monoshock suspension (ShS) and spring-based suspension (SpS) compared with the standard rigid (SR) and elastomer (ElS) wheelchairs (Table I). Overall, change in seat force (dF), peak upward head accelerations (Avmax), peak forward head accelerations (Ahmax), and peak backward head accelerations (Ahmin) were lower in the wheelchairs with suspension (ElS, SpS, ShS) compared with the (non-suspension) SR wheelchair. However, only Ahmin was significantly lower in ShS (–0.23 ± 0.43 g) compared with SR (–1.10 ± 0.41 g) and ElS (–0.96 ± 0.53 g). When comparing across all the wheelchairs, the monoshock-based suspension (ShS) had the lowest impact landings. Specifically, the dt of ShS was 43% longer than SR, 34% longer

Table I. *Total Seat Force is the resultant of 3D seat force at impact or initial contact (Fic) to peak force (Fmax) to change in seat force (dF = Fmax-Fic) in body weight (BW). Rise time (Rt) from Fic to Fmax in msec. Head Accelerations in the vertical and horizontal directions in g (9.81m/s²). Peak vertical in upward (positive) direction (Avmax) in g, Peak Forward (Ahmax) horizontal head accelerations in anterior (positive) direction in g, and Peak Backward (Ahmin) horizontal head accelerations in posterior (negative) direction in g*

	SR	ElS	SpS	ShS
Total seat force, mean (SD)				
Fic, kg	0.58 (0.35)	0.80 (0.34)	0.59 (0.28)	0.77 (0.32)
dF, kg	2.03 (0.35)	1.69 (0.48)	1.87 (0.52)	1.51 (0.64)
dt, msec	59 (14)	68 (14)	90 (7)*	103 (11)*
Head Accelerations, mean (SD)				
Avmax, g	1.69 (0.44)	1.51 (0.41)	1.57 (0.37)	1.33 (0.29)
Ahmax, g	1.95 (0.80)	1.47 (0.60)	1.46 (0.82)	1.08 (0.51)
Ahmin, g	–1.10 (0.41)	–0.96 (0.53)	–0.65 (0.51)	–0.23 (0.43)*

Note: Values are mean (1 SD).
*Significantly different from ElS and SR ($p < 0.05$).
SR: standard rigid; ElS: elastomer-based suspension; SpS: spring-based suspension; ShS: monoshock suspension; SD: standard deviation.

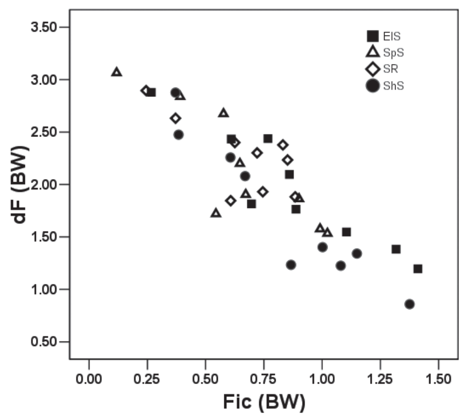

Fig. 4. Seat force at initial contact (Fic) in body weight (BW) vs change in seat force (dF) in BW from initial contact (Fic) to peak force (Fmax) for each subject during curve descents with the non-suspension rigid (SR) (◇), elastomer-based (ElS) (■), spring-based (SpS) (△), and monoshock-based (ShS) (●) wheelchairs. Each point represents an average of 5 trials per subject per wheelchair. This relationship is used to examine the level of "pull-up" users may use prior landing in order to modulate the seat force.

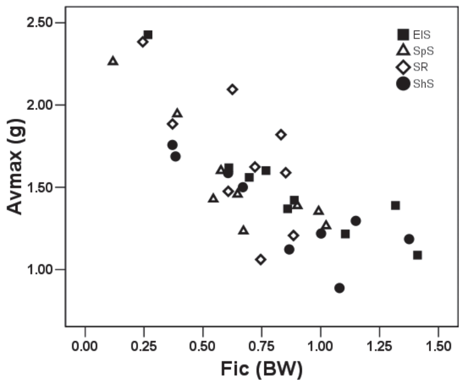

Fig. 5. Seat force at initial contact (Fic) in body weight (BW) vs peak vertical head acceleration (Avmax) in g for each subject during curve descents with the non-suspension rigid (SR) (◇), elastomer-based (ElS) (■), spring-based (SpS) (△), and monoshock-based (ShS) (●) wheelchairs. Each point represents an average of 5 trials per subject per wheelchair. This relationship is used to examine the level of "pull-up" that users may employ prior to landing in order to modulate the vertical head acceleration.

than ElS, and 13% longer than SpS, while the dF of ShS was 26% less than SR, 21% less than SpS, and 9% less than ElS. Lower impact landings are an indicator of softer and more comfortable ride quality.

Wheelchair users showed less seat force and head accelerations when they increased the seat force at initial contact (Fic) (e.g. through "pull-up"), but this was not consistent across wheelchairs and across subjects. For all suspension wheelchairs, Fic was negatively associated with dF (Fig. 4) and Avmax (Fig. 5), and positively associated with Ahmin (Fig. 6) (Table II). These linear associations (i.e. less force and vertical acceleration with greater pull-up) were maintained mostly by ShS; with the highest and most numerous levels of correlations between Fic and dF (r=–0.95), Avmax (r=–0.79), and Ahmin (r=0.85). In addition, ShS showed significant negative correlation between Fic and Ahmax (r=–0.72). In contrast, SR had the lowest correlation levels between Fic and dF (r=–0.71), Avmax (r=–0.67), and Ahmin (r=–0.40) and SpS showed a non-significant linear association between Fic and Ahmin (r=0.58) and no association between Fic and Ahmax (r=0). Fic also tended to be lower in the SR (0.58 ± 0.35BW) and SpS (0.59 ± 0.28BW) (i.e. less pull-up) than ElS (0.80 ± 0.34BW) and ShS (0.77 ± 0.32BW) (i.e. greater pull-up), but did not reach statistical significance (Table I).

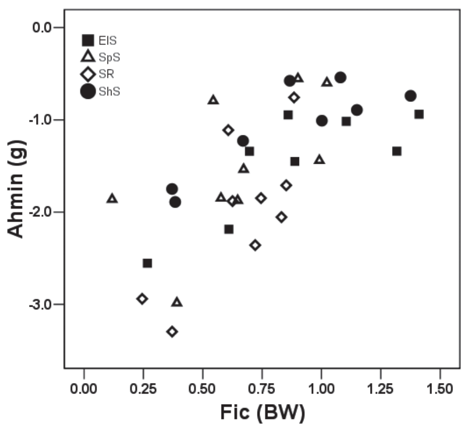

Fig. 6. Seat force at initial contact (Fic) in body weight (BW) vs peak backward (negative) head acceleration (Avmin) in g for each subject during curve descents with the non-suspension rigid (SR) (◇), elastomer-based (ElS) (■), spring-based (SpS) (△), and monoshock-based (ShS) (●) wheelchairs. Each point represents an average of 5 trials per subject per wheelchair. This relationship is used to examine the level of "pull-up" that users may employ prior to landing in order to modulate the horizontal head acceleration.

Table II. *Pearson product correlations (r) between the seat force at impact (Fic) and magnitude of seat force (dF), peak upward head accelerations (Avmax), and peak backward head accelerations (Ahmin) during landings in a rigid (SR) and suspension type (ElS, SpS, ShS) wheelchairs*

	SR	ElS	SpS	ShS
Pearson Correlation (r)				
Fic vs. dF	–0.71*	–0.95**	–0.89**	0.95**
Fic vs. Avmax	–0.67	–0.86**	–0.86**	–0.79*
Fic vs. Ahmax	–0.40	–0.69	0.00	–0.72*
Fic vs. Ahmin	0.75*	0.78*	0.58	0.85**

*Correlation is significant at the 0.05 level (2-tailed).
**Correlation is significant at the 0.01 level (2-tailed).
ElS: elastomer-based suspension; SpS: spring-based suspension; ShS: monoshock suspension.

DISCUSSION

This study determined the seat force and head accelerations experienced by manual wheelchair users performing curb descent landings with rigid (non-suspension) and rear- suspension-type wheelchairs. Users were found to experience lower seat force and head accelerations when using a suspension-type compared with non-suspension rigid hand-rim wheelchair. The use of a rear-suspension system with a hand-rim wheelchair and the "pull-up" technique that users employ in preparation for landing were found to influence the magnitudes of the seat forces, rise time, and accelerations experienced by the user. Overall, lowering the seat force and head accelerations are indicators of a more comfortable ride and can be associated with a reduced likelihood of the user sustaining musculoskeletal injuries due to shocks and vibration. These quantifiable measures can provide an objective means of assessing characteristics of good ride quality in hand-rim wheelchairs and a means to support the epidemiological studies conducted to establish a link between exposure to vibration and health risks.

Currently, methods of measuring, evaluating and assessing whole body vibration and repeated shock are offered. The most commonly cited method is the ISO 2631 Standard (4). ISO 2631 sets guidelines for how to take measurements and calculate exposure statistics and also recommends acceptable dosage levels. These guidelines, however, do not provide quantitative relationships between the measured quantities and specific health risk. Although many would agree that the available data indicate that shock and vibration probably do lead to health problems (32, 33), epidemiological research has not given us any understanding of the processes by which shock and vibration affect health. It is unclear what type of damage will occur and what mechanisms are involved in the damage process. It is therefore not possible to state with any precision how the damage depends on the physical characteristics of the vibration and shocks or the characteristics of the person or other environmental factors. However, our findings provide quantitative information for developing and assessing the effectiveness of selected interventions and the ability of hand-rim designs to improve ride quality based on the premise that reducing the seat force during curb descent landings will minimize the excessive spinal loads that can increase the risk for injuries. Since spinal loads cannot be measured directly *in vivo*, biomechanical models (25, 34–36) are recognized to play an indispensable role in our future work.

The suspension systems examined in this study reduced the forces transmitted from the wheelchair to the user by extending the peak force rise time (longer loading interval), resulting in lower head accelerations for the wheelchair user. However, the suspension element and configuration of the suspension system can influence the level of attenuation of the seat force and head accelerations. Among the suspension frame wheelchairs, the ShS (monoshock-based) demonstrated the least seat force and head accelerations. This particular suspension design utilizes a monoshock (RockShox®) system that is mounted orthogonally beneath the seat to couple the axle and lower frame section with the seat (Fig. 1D). During the landing phase of the curb descent, the orientation of the travel of the monoshock suspension element appeared to be aligned with the ground reaction force vector. Consequently, the orientation of this type of suspension element contributed to the shock absorbing effectiveness by distributing the loading duration from the ground to the seat over a greater time interval. Suspension elements oriented relative to force results in more deformation (and more time) in the same direction, resulting in longer time to bottom out at a lower velocity. Similarly, the SpS (spring-based suspension) demonstrated lower seat force and head accelerations. This particular design utilizes an independent rear suspension system composed of an A-arm assembly and metal spring (Fig. 1B). During landings, the A-arm vertical travel is controlled by the compression of the metal spring. Since the compression travel of the SpS is less than that of the ShS, a shorter time to bottom out and greater velocity was observed for the SpS wheelchair. In contrast, the ElS (elastomer-based suspension) utilizes an elastic polymer disks placed between the axle and the lower frame section (Fig. 1C). During landing, the compression travel of disk was very short, resulting in earlier time to bottom out; resulting in larger seat force and head accelerations.

The head accelerations during curb descent landings were also reduced when an increase in seat force was observed at impact. An increase in seat force at impact (i.e. closer to 1 BW) indicated that the relative velocity between the seat and the user was low prior to the wheel–ground contact. As a result, there was minimal relative motion between the user and the chair, thereby minimizing the head acceleration. This increased seat force was the result of the user "pulling-up" against the hand rim in preparation for landing. This pull-up is similar to the strategies seen in non-disabled individuals in preparation for feet first landings from a height (37). The ability to "pull up" as a way of increasing load dissipation may vary with the skill level of the wheelchair user, as observed during landings performed by elite gymnasts (38). In our study, not all subjects use pull-up and they were not consistent across wheelchairs; indicating that pull-up is a self-selected strategy that might be facilitated via the suspension. In particular, the high correlation seen between the seat force at initial contact and force and accelerations during landing in the monoshock suspension (ShS) is the combined result of the pull-up and suspension system. This information may prove invaluable in identifying optimal wheelchair frame configuration and suspension design, and will be explored further in future studies. Furthermore, future studies will need to explore the consequences of using the pull-up strategy among wheelchair users with varying upper extremities and trunk strengths, and injury-causing potential.

Our current study was limited to a group of subjects with the same level of SCI (T12) and with full upper extremity strengths, which allowed them to perform repeated self-initiated curb descent landings. Individuals with diminished muscular strength (e.g. those with tetraplegia), who are yet able to independently descend curbs may experience greater overall exposure to shock and vibration than those tested here. The orientation of the user's body segment, particularly the trunk and head orientation relative to the wheelchair, will also influence the biodynamic responses during curb descent landings. For instance, those who landed with a more forward rotated head will most likely experience greater bending moment at the neck than those who landed with a less forward rotated head. Those who are aging will likely develop spine curvatures (39, 40) that may predispose the back to greater load-bearing situations (41). For individuals with tetraplegia and high paraplegia, paralysis of critical trunk stabilizing musculature greatly impairs sitting balance (42). By "slouching" in a "C"-shaped posture, patients lacking normal extensor function of the trunk and hips achieve some passive stability by shifting their trunk center of mass more posteriorly, thereby decreasing the need for trunk extension. However, this adaptive posture leads to increased thoracic kyphosis, a more forward head position, and cervical hyperextension and a protracted scapulae (43). With exposure to shock, this posture can lead to a negative biomechanical chain of events that contribute to increased development of neck and back discomfort and pain (13). A more complete understanding of shock and vibration exposure from wheelchair use in individuals with varying levels of SCI has the potential to improve comfort and ride quality through optimal seating interventions and equipment design; and play a role in delaying the onset of well-documented accelerated functional changes found among persons who are aging with a disability (44). Future studies will focus on the response among wheelchair users with varying physical characteristics.

We determined that the monoshock suspension (ShS) system provided the most effective attenuation of the force and accelerations during curb descent landings using a time domain peak analysis. In comparison, Kwarciak et al. (31) used the ISO-2631 standard for analysis to also determine that the monoshock suspension element resulted in significantly lower peak seat accelerations than the folding and rigid-frame wheelchairs during curb descents from 3 heights (0.05, 0.10, and 0.15 m). They also attributed this superior performance to the small suspension angle at impact (i.e. suspension element most aligned with the ground reaction force). Furthermore, the results presented in this study were limited to a single (0.10 m) drop height; a curb drop height typically encountered in urban sidewalk areas. In a preliminary test, we measured the seat force and head accelerations from 3 heights (0.05, 0.10, and 0.15 m) in all wheelchairs by one participant. We determined that the magnitude of the seat force and head accelerations was linearly related to the curb height. Similarly, a study by Kwarciak et al. (31) determined that the peak and frequency-weighted peak seat accelerations (using the ISO-2631 analysis guidelines) also increased with height (0.05, 0.10, and 0.15 m) across frame types: rigid, suspension, and folding wheelchairs. However, the vibration-suppression performance decreased as the height increased, regardless of wheelchair frame type. Likewise, we expect that the rigid and suspension-type frames will reach a load-dissipating limit at higher drop heights. Based on our findings however, we also expect that users will use the pull-up as a means to minimize the head accelerations experienced at curb descents from greater heights.

In conclusion, hand-rim wheelchairs with suspension systems can provide softer landings by attenuating the magnitude and duration of the force and by reducing head accelerations. The level of attenuation, however, varies with the specific suspension element and design of the suspension system. Hand-rim wheelchair users can also soften the landings by utilizing a pull-up strategy that reduces the force and head accelerations. Softer landings can contribute to better ride quality. Further research is needed to gain a full understanding of relationships between shock and vibration, comfort level, and onset of injury. Ultimately, this approach will lead to better designs and the development of interventions aimed at increasing comfort, decreasing fatigue and injuries, and enhancing wheeled mobility.

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