Seat and Footrest Shocks and Vibrations in Manual Wheelchairs With and Without Suspension

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ABSTRACT. Cooper RA, Wolf E, Fitzgerald SG, Boninger ML, Ulerich R, Ammer WA. Seat and footrest shocks and vibrations in manual wheelchairs with and without suspension. Arch Phys Med Rehabil 2003;84:96-102.

Objective: To examine differences in the shock and vibration transmitted to an occupant of a manual wheelchair with and without suspension caster forks and with and without rear-suspension systems.

Design: Repeated-measures engineering testing.

Setting: Rehabilitation engineering center with a wheelchair standards test laboratory.

Specimens: Six manual wheelchairs.

Interventions: An American National Standards Institute/Rehabilitation Engineering and Assistive Technology Society of North America wheelchair test dummy and a Hybrid III test dummy were used to test shock and vibration transmission in wheelchairs equipped with original equipment manufacturer (OEM) caster forks and suspension caster forks. Ultralight wheelchairs, half of which had factory-equipped rear-suspension systems, were tested. Testing was conducted on a double-drum wheelchair test machine.

Main Outcome Measures: Shocks were examined by using peak acceleration and the frequency at which peak acceleration occurs for the seat and footrest. Vibration was characterized by the acceleration power per octave for the seat and footrest.

Results: Significant differences were found in the peak accelerations at the seat (P=.0004) and footrest (P=.0007) between the wheelchairs with the OEM caster forks and those with the suspension casters. The wheelchairs with suspension had significantly different frequencies at which the peak accelerations occurred for both the seat (P=.01) and footrest (P=.0001). The wheelchairs with suspension caster forks had a lower total power per octave than the wheelchairs with the OEM caster forks. For the footrest vibrations, significant differences were found between the types of caster forks for all octaves except those associated with frequencies more than 78.75Hz. There were significant differences for wheelchairs with and without rear suspension for total power per octave of seat vibrations in the octaves between 7.81 and 9.84Hz (P=.01) and 12.40 and 15.63Hz (P=.008).

Conclusions: Suspension caster forks reduce the shock and vibration exposure to the user of a manual wheelchair. Rear-suspension systems reduce some of the factors related to shock and vibration exposure, but they are not clearly superior to traditional designs.

Key Words: Cumulative trauma disorders; Rehabilitation; Vibration; Wheelchairs.

IMPORTANT ADVANCES in wheelchairs have occurred over the past 20 years.[1] Recently, attention has been paid to improving rider comfort and reducing the risk of developing secondary disabilities.[2-4] The secondary injuries due to whole-body vibration in workers in the trucking, aircraft, helicopter, maritime, and construction industries are well defined.[5] These studies have correlated vibration exposure with the risk of diskogenic back injuries. As a result of these injuries, industry seating systems (suspension, cushions, back supports) and sitting styles (ie, posture, muscle tone) have been compared to determine the conditions that minimize the transmission of vibration.[6-12] Human models have been developed and used to design optimal seating systems for industrial and vehicle applications.[13-18]

Understanding biomechanics can provide insight into the cause of low back injury in individuals who use manual wheelchairs for mobility.[19] McGill[19] found that the injury processes were associated with high loads and with repeated or sustained low loads. The vibrations and shocks individuals encounter on a daily basis while propelling their wheelchair may be sufficient to cause injury. In addition, wheelchair users may be at risk of injury due to the secondary effects of vibration exposure when they perform transfers or lift their chairs. Dupuis et al[20] recommended the avoidance of heavy lifting after vibration exposure (ie, unloading a truck), because individuals have a decreased ability to respond to suddenly applied loads after whole-body vibration. This was further supported by Magnusson et al,[21] who found that the spine is in a poorer condition to sustain large loads after long-term vibration exposure. Lings and Leboeuf-Yde,[22] on the basis of a review of literature from 1992 to 1999, concluded that there is a reasonable probability of an association between whole-body vibrations and low back pain.

The effects of whole-body vibration on wheelchair users are not well defined. Boninger et al[23] reported that people with tetraplegia due to spinal cord injury show changes in their spinal curvature after only a short period of wheelchair use. In a study by DiGiovine et al,[4] wheelchair users reported differences in the ride comfort of lightweight and ultralight manual wheelchairs, with ultralight wheelchairs being more comfortable. VanSickle et al[2] recorded the forces and moments exerted on a manual wheelchair and reported that during common activities, the forces exceeded several times the body weight. In

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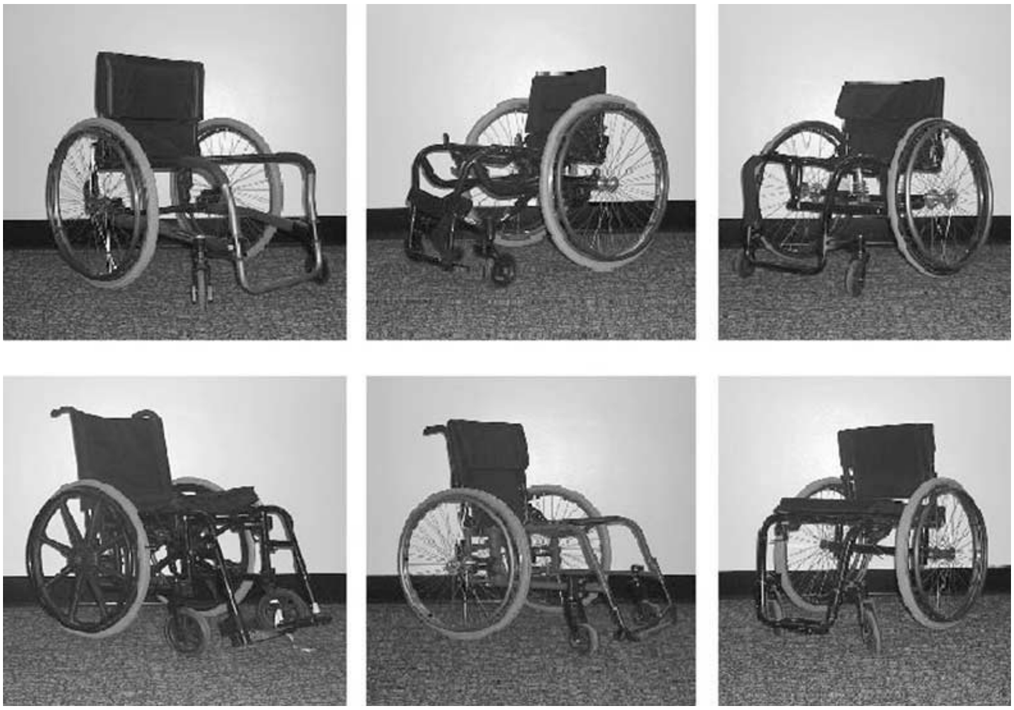

Fig 1. Wheelchairs used in this study.

another study, VanSickle et al[24] showed that the vibrations experienced during manual wheelchair use exceed international standards for vibration exposure for vehicles. In a recent study, DiGiovine et al[25] recorded the shocks and vibrations at the wheelchair seat and at the rider's head with a selection of cushions. This study found differences in the shocks and vibrations absorbed by cushions.

The American National Standards Institute (ANSI)/Rehabilitation Engineering and Assistive Technology Society of North America (RESNA) wheelchair test dummy (WTD) is available to most wheelchair manufacturers and is approximately one tenth the cost of a Hybrid III test dummy (HTD).[a] Therefore, we wished to determine whether the ANSI/RESNA WTD was suitable for loading a wheelchair when recording shock and vibration.

The purpose of this study was to determine whether suspension systems attenuated the shocks and vibrations induced in the footrests and seats of manual wheelchairs. We also studied the effect of 2 types of test dummies: an ANSI/RESNA WTD and an HTD.[1] We hypothesized that there would be differences in the shocks and vibrations imparted through the wheelchair when suspension was used and that there would be no differences on the basis of the dummy type.

METHODS

Test Wheelchairs

Six manual wheelchairs were tested; 3 were rear-suspension wheelchairs (Quickie XTR,[b] Invacare A6S,[c] Everest & Jennings Barracuda[d]), and 3 contained no suspension system. These wheelchairs included a rigid box-frame ultralight (Quickie GP[b]), a rigid cantilever-frame ultralight (Kuschall 3000[d]), and a folding lightweight (Kuschall Champion 1000[d]) (fig 1). The Barracuda and A6S use polymer-based shock absorbers, whereas the XTR uses a spring-dampener design (fig 2). All of the wheelchair frames were made of aircraft aluminum. The wheelchairs were selected because they represent a cross-section of common frame styles used on ultralight wheelchairs.

ANSI/RESNA committees have been developing wheelchair standards for more than 20 years.[1] The ANSI/RESNA standards are quite comprehensive and serve as a guide in the design and quality assurance of wheelchairs.[26] The ANSI/RESNA wheelchair standards provide a method of measuring and evaluating both manual and power wheelchairs. All of the wheelchairs tested in this study were of similar dimensions (table 1). The measurements were based on section 7 of the ANSI/RESNA standards—Measurement of Seating and Wheel Dimensions.[27] The dimensions were matched to control for the position of the center of mass of the wheelchair.

Four different conditions were tested with each of the 6 manual wheelchairs, for a total of 24 trials. Each wheelchair was evaluated with 2 testing dummies (ANSI/RESNA WTD, HTD). The mass of the ANSI/RESNA WTD is 100kg, whereas the HTD has a mass of 75kg. These 2 types of dummies were selected because they are the most commonly used in wheelchair testing. The ANSI/RESNA WTD is primarily used for stability and strength testing, whereas the HTD is often used for ergonomic analyses and crash testing. Each wheelchair was tested with 2 types of caster forks: standard original equipment manufacturer (OEM) caster forks and polymer-based suspension caster forks (SCFs) (Frog-Legs[e]) (fig 3). The OEM caster wheels were used in all tests.

Data Collection

To collect the desired data, 2 triaxial accelerometers[f] were used. One was attached to a specially constructed ¼-in-thick aluminum plate, which was then laid on the seat of the wheelchair being tested (fig 4). The standard foam used during wheelchair fatigue testing was used to simulate a cushion and a user's soft tissue. The second accelerometer was attached to a specially constructed plate bolted to the footrest of the wheelchair (fig 4). A custom data-collection program was written with LabVIEW[g] software to interface with a data-acquisition

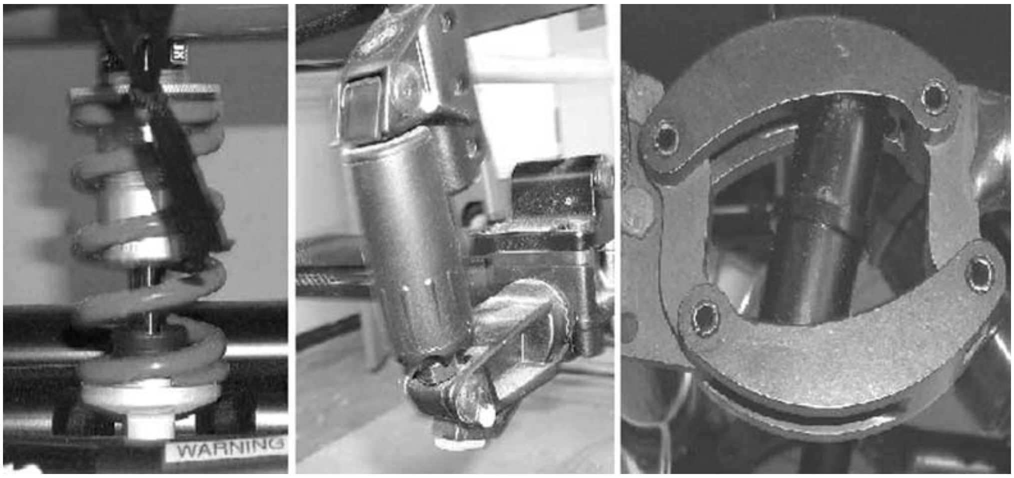

Fig 2. The suspension methods for each of the suspension wheelchairs.

card.[g] The acceleration data from all 6 channels were collected at 1000Hz.[2] Data were collected on an ANSI/RESNA double-drum test machine[h] for 250 cycles.[2] Each wheelchair was run for 250 cycles in each of the configurations.

Data Reduction

The acceleration data were calibrated and converted for analysis in custom software written with MATLAB®.[i] The first part of the data analysis consisted of converting each of the 3 axes of the accelerometers into a resultant acceleration vector (a) for both the seat and the footrest:

$$a(n)=\|a(n)_x^2+a(n)_y^2+a(n)_z^2\|, \qquad (1)$$

where the subscripts x, y, and z represent the fore-aft, mediolateral, and superior-inferior directions, respectively, and the variable n represents an individual sample. To reduce the noise in the signal, it was processed by using an autocorrelation sequence, where m is the lag index[28]:

$$\hat{r}_{xx}(m)=\frac{1}{N-|m|}\sum_{n=0}^{N-|m|-1} x(n)\times x(n+|m|) \qquad (2)$$

The resultant autocorrelated vectors for both the seat and the footrest were conditioned by using a Hamming window $[W(n)]$[29]:

$$W(k)=.54\pm.46\cos\left(\frac{2\pi k}{N}\right)\; k=-\frac{N}{2}\cdots,-1,0,1,\ldots\frac{N}{2}, \qquad (3)$$

where N represents the number of samples and k is an index. The conditioned acceleration data were then entered into a fast Fourier transform algorithm to determine their respective frequency spectra[28]:

$$A(K+1)=\sum_{n=0}^{N-1} a(n+1)\times e^{-j(2\pi/N)}, \qquad (4)$$

where variable A represents the resultant acceleration in the frequency domain, K is the frequency index of each sample, and $j=\sqrt{-1}$.

The power spectral density (PSD) is a means of showing how the power of the acceleration is distributed over frequency. Studies have shown that acceleration power in specific frequencies may cause injury or induce premature fatigue. Hence, the PSD was divided into the frequency octaves for human vibration exposure[5]:

$$f_2=(2^{1/3})\times f_1 \qquad (5)$$

The PSD for each octave was determined by integrating the signal over the length of the octave being measured. In equation 5, one starts with f_2=500Hz and then calculates f_1=397Hz; then the resulting frequency, f_1=397Hz, becomes the new f_2=397Hz, and a new f_1=315Hz is calculated. This process is repeated until the resulting frequency is $\leq$1.25, which is the lowest boundary as specified in International Organization for Standardization 2631.[5] The area under the curve at each octave was used as a measure of the total vibration power per octave. The total power per octave and peak accelerations were used in the statistical analysis for comparing the wheelchairs and test dummies.

Table 1: Dimensional Characteristics of Wheelchairs Tested

Characteristic	Quickie XTR	Barracuda	Invacare A6S	Kuschall 1000	Quickie GP	Kuschall 3000
Propelling wheel diameter (cm)	61	61	61	61	61	61
Caster wheel diameter (cm)	12.7	12.5	12.5	17	11.5	14.5
Wheelbase (cm)	37	43.5	43.8	34	38	34.5
Width between rear wheels (cm)	50.3	57.5	55.2	48.2	52.6	54.2
Width between caster wheels (cm)	43.7	50	45.4	45.5	45	39.3
Horizontal distance of wheel axle forward of backrest (cm)	5	7.5	2.5	3.5	3.5	5.5

Fig 3. Suspension caster fork from Frog-Legs.

Statistical Analysis

Comparisons of power per octave were made within each octave between the caster forks (with and without suspension), wheelchair types (with and without rear suspension), and dummy types (ANSI/RESNA WTD vs HTD). This analysis was completed for both seat and footrest accelerations by using paired Student *t* tests. Four separate mixed models were developed for the peak seat accelerations, peak foot accelerations, and seat and foot frequencies. Mixed-model analysis was used because both fixed and random effects were incorporated into the model. The random effects were the different types of wheelchairs, whereas the fixed effects were the dummy type, the type of caster, and whether the chair was considered to have suspension. Group comparisons were used to determine whether suspension systems reduced shock and vibration over traditional wheelchair designs. Analysis was completed with SAS.[j] The significance level was set at .05.

RESULTS

The peak accelerations recorded at the footrests and seats and the frequencies at which they occurred are reported in table 2. Significant differences were found in the peak accelerations at the seat ($P=.0004$) and footrest ($P=.0007$) between the wheelchairs when the OEM caster forks were used versus the SCFs. In both cases, the peak accelerations were lower with the SCFs. The wheelchairs with suspension had significantly different frequencies at which the peak accelerations occurred for both the seat ($P=.01$) and footrest ($P=.0001$). The frequencies at which the peak vibrations occurred were higher with the suspension wheelchairs. There were no significant differences in the peak accelerations when the suspension and nonsuspension wheelchairs or the 2 test dummies were compared. There were no significant differences in the frequencies at which the peak accelerations occurred for the caster forks or the test dummies.

Table 3 shows that for the seat vibrations, significant differences were found between the wheelchairs with and without SCFs for the majority of octaves. No significant difference was found for the octaves from 6.20 to 7.81Hz and 125.00 to 157.49Hz. The wheelchairs that had the SCFs had lower total power per octave than the wheelchairs with the OEM caster forks for each of the octaves. For the footrest vibrations, significant differences were found between the types of caster forks for all octaves except those associated with frequencies greater than 78.75Hz. As with the seat vibrations, the wheelchairs with the SCFs had lower total power per octave than those with the OEM caster forks.

The results of comparing the vibration power in each octave between the wheelchairs with and without rear wheel suspension are presented in table 4. There were significant differences between the styles (with and without rear suspension) of wheelchair for the total power per octave of the seat vibrations in the octaves from 7.81 to 9.84Hz ($P=.01$) and 12.40 to 15.63Hz ($P=.008$). The total power per octave was lower for the wheelchair with rear suspension between 7.81 and 9.84Hz and was higher for these same chairs between 12.40 and 15.63Hz. There were no significant differences in footrest vibration power per octave between wheelchairs with and without rear wheel suspension.

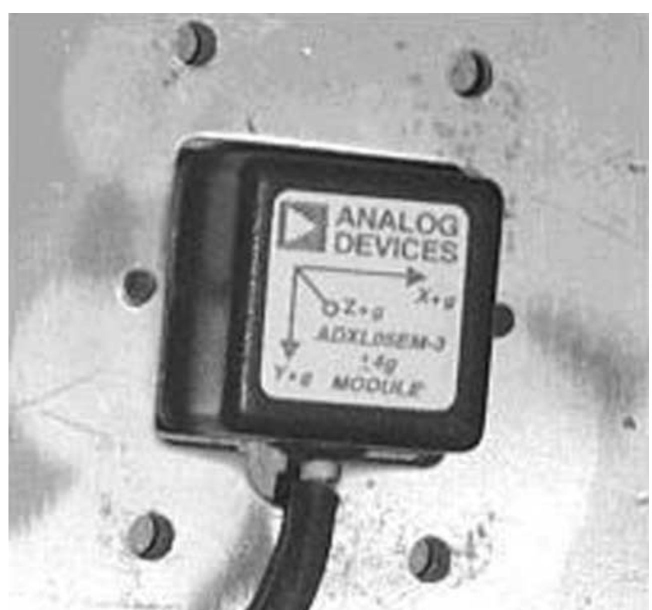

Fig 4. Instrumentation used to collect acceleration data at the seat and footrest.

Table 2: Peak Accelerations (m/s^2) for the Seat and Footrest and the Frequencies (Hz) at Which They Occur

Variable	Seat		Footrest	
	Peak Acceleration	Frequency	Peak Acceleration	Frequency
Dummy: WTD	10.5±6.0	7.7±4.9	10.1±6.6	10.8±14.8
Dummy: HTD	13.9±2.7	9.1±5.3	14.7±9.1	8.2±10.1
Casters: OEM	18.2±1.8*	8.4±4.9	18.6±11.4*	9.2±13.8
Casters: SCF	6.3±4.3*	8.5±5.1	6.1±3.6*	9.9±11.2
With suspension	13.1±0.6	9.8±6.8†	13.0±6.4	15.2±16.6†
Without suspension	11.3±2.8	7.0±2.5†	11.8±8.6	3.9±2.1†

NOTE. Values are mean ± standard deviation (SD).
*Significant differences found in the peak accelerations, seat (P=.0004) and foot (P=.007), between caster type in the mixed model.
†Significant differences found in seat frequency (P=.01) and foot frequency (P=.0001) with and without suspension in the mixed model.

Table 5 shows that for all of the octaves from 157 to 500Hz, significant differences were found between the dummy types for both footrest and seat vibrations. The ANSI/RESNA WTD had power per octave values approximately twice those of the HTD in the octaves in which differences were significant.

DISCUSSION

Long-term wheelchair use may lead to secondary injuries as a result of exposure to shock and vibration. The interest in reducing the shocks and vibrations transmitted to the users of manual wheelchairs has resulted in both suspension forks and rear-suspension systems. Front- and rear-suspension systems may also facilitate obstacle negotiation, much as suspension does in other vehicles. Although the variety and popularity of suspension wheelchairs are growing, there is little information to determine whether they are indeed effective at reducing the shock and vibration transmitted to the user.

The greatest risk for injury due to shock and vibration exposure is when the frequency is near the natural frequency of seated humans.[5] The natural frequency of seated humans is between 4 and 12Hz.[10] At the natural frequency, the shock and vibration induced in the body are amplified, thus increasing the risk of injury. It is desirable either to reduce the amplitude or power of the shock and vibration or to shift it in frequency so that it is outside the range of natural frequencies of humans. In our study, the peak frequencies typically occurred in the 4- to 15-Hz range. There was a significant difference in the frequency at which the peak acceleration occurred for the 2 types of wheelchairs, with the suspension wheelchairs increasing the frequency at which the peak occurred. However, both styles of wheelchairs tended to transmit peak accelerations in the natural frequency range of humans. This indicates that both the rear-suspension systems and SCFs may not be optimized for shock and vibration reduction among manual wheelchair users. The SCFs did significantly reduce, by nearly one third, the peak acceleration amplitudes at both the seat and footrest.

The power per octave is a good measure of vibration or repeated shock exposure, whereas the peak acceleration is closely related to infrequent shock exposure. Our analysis of the power per octave showed that the SCFs were successful at reducing both the seat and footrest vibrations across a broad range of frequencies. The footrest vibrations were significantly

Table 3: Seat and Footrest Vibrations: Comparisons of Caster Forks at Different Octaves

Octave (Hz)	Seat Vibration Power (m·Hz^{-1}·s^{-2})			Footrest Vibration Power (m·Hz^{-1}·s^{-2})		
	SCF	OEM Forks	P	SCF	OEM Forks	P
<1.25	.04±.02	.19±.24	.05	.05±.03	.27±.30	.03
1.25–1.55	.02±.01	.06±.04	.006	.03±.01	.10±.07	.006
1.55–1.95	.02±.01	.08±.05	.002	.03±.01	.12±.08	.002
1.95–2.46	.43±0.5	1.7±1.9	.04	.88±.68	3.37±3.11	.01
2.46–4.92	.05±0.1	0.2±0.1	.003	.09±.04	.31±.23	.007
4.92–6.20	.10±0.1	0.3±0.2	.02	.15±.05	.42±.31	.01
6.20–7.81	.62±1.1	1.2±1.8	.17	.79±.61	1.99±1.73	.04
7.81–9.84	.20±0.1	0.7±0.6	.008	.41±.25	.84±.37	.003
9.84–12.40	1.0±0.4	3.8±2.3	.001	1.0±1.1	2.2±1.4	.03
12.40–15.63	1.0±0.9	2.4±0.9	.001	1.5±2.1	2.8±1.7	.08
15.63–19.69	1.2±1.3	2.6±1.2	.01	1.8±2.3	3.7±2.04	.04
19.69–24.80	1.4±1.7	3.5±2.1	.01	2.4±3.7	4.1±2.7	.21
24.80–31.25	1.3±1.4	3.1±1.5	.005	2.3±2.8	4.5±3.3	.10
31.25–39.37	1.2±1.1	3.1±3.3	.08	2.3±2.6	6.5±6.8	.06
39.37–49.61	0.9±0.6	2.5±1.3	.001	1.6±1.5	5.3±3.0	.002
49.61–62.50	1.1±0.6	3.6±2.5	.005	2.1±1.9	6.9±5.1	.008
62.50–78.75	1.4±0.8	3.0±1.9	.02	2.5±2.1	6.9±5.7	.03
78.75–99.21	1.3±0.7	2.5±1.2	.009	—	—	—
99.21–125.00	1.5±1.1	2.7±1.4	.04	—	—	—
125.00–157.49	1.6±1.0	2.0±0.8	.24	—	—	—
157.49–198.43	0.9±0.1	0.2±1.9	.03	—	—	—

NOTE. Values are mean ± SD. Significant at P<.05.

Table 4: Seat and Footrest Vibrations: Comparisons of Wheelchair Types at Different Octaves

Octave (Hz)	Seat Vibration Power ($m \cdot Hz^{-1} \cdot s^{-2}$)			Footrest Vibration Power ($m \cdot Hz^{-1} \cdot s^{-2}$)		
	Suspension Wheelchairs	Standard Wheelchairs	*P*	Suspension Wheelchairs	Standard Wheelchairs	*P*
<1.25	.11	.12	.93	.14	.18	.73
1.25–1.55	.03	.05	.51	.05	.08	.36
1.55–1.951	.05	.05	.99	.07	.08	.79
1.95–2.46	.99	1.16	.88	1.80	2.46	.69
2.46–4.92	.09	.14	.47	.15	.26	.29
4.92–6.20	.15	.22	.53	.21	.35	.16
6.20–7.81	.64	1.15	.47	.93	1.88	.36
7.81–9.84	.32	.63	.01	.63	.62	.97
9.84–12.40	2.39	2.49	.89	2.14	1.06	.25
12.40–15.63	2.10	1.29	.01	3.11	1.18	.26
15.63–19.69	2.32	1.39	.20	3.80	1.67	.23
19.69–24.80	2.93	2.02	.43	4.90	1.64	.21
24.80–31.25	2.70	1.68	.31	4.86	1.93	.22
31.25–39.37	3.08	1.14	.28	6.31	2.50	.27
39.37–49.61	2.14	1.34	.29	4.32	2.59	.31
49.61–62.50	2.70	2.03	.62	5.86	3.23	.36
62.50–78.75	2.47	1.90	.58	6.18	3.15	.37
78.75–99.21	2.11	1.71	.64	4.24	2.83	.45
99.21–125	2.36	1.81	.56	5.30	2.98	.33
125.00–157.49	2.08	1.52	.24	4.85	3.03	.34

NOTE. Values are mean ± SD. Significant at $P<.05$.

reduced for the entire range (0–50Hz) of frequencies of interest for human vibration exposure. Because the SCFs reduced the peak amplitude and power per octave at the footrests, they may be useful for reducing stretch reflex response spasms among manual wheelchair users. The suspension casters yielded mixed results for the power per octave at the seat. For most frequency octaves of interest, the SCFs significantly reduced the power per octave, with the exception of the octave between 6.20 and 7.81Hz, which is in the middle of the seated human natural frequency range. Given its effects on vibration and shock transmission suspension, caster forks, such as Frog-Legs, should be considered for active clients or individuals who have chronic pain. Frog-Legs may be retrofitted to many manual wheelchairs.

When comparing the wheelchairs with rear-suspension systems with those without, there were 2 important differences in the vibration power per octave. The wheelchairs with rear suspensions performed better, reducing the vibration power by nearly one half, in the octave between 7.81 and 9.84Hz. This octave is within the range of the natural frequency of seated humans. It is interesting to note that there was also a significant difference at the octave between 12.40 and 15.63Hz, with the traditional wheelchairs showing a lower vibration power. These results indicate that the vibration transmitted to the wheelchair at the lower octave (7.81–9.84Hz) may be shifted to the higher octave (12.40–15.63Hz) by the rear-suspension system. Shifting vibrations away from the natural frequency range of humans and reducing the amplitude or power are all desirable for reducing the risk of injury due to cumulative trauma. Ideally, the frequency shift would be higher to ensure no overlap with the natural frequency range of humans. Once suspension systems are shown to reduce vibration within the entire natural frequency range of humans, they could be recommended as a further means of reducing the risk of injury.

Data were collected with 2 types of test dummies: a WTD constructed in accordance with the ANSI/RESNA standards[27,30] and an HTD.[31] Our results indicate that no significant differences existed between the 2 dummies for peak accelerations and the frequencies at which they occurred. There were significant differences in the vibration power for octaves from 157 to 500Hz; however, frequencies greater than 150Hz are far beyond those of interest for human vibration exposure. The results at the higher-frequency octaves were as we expected. The ANSI/RESNA WTD is designed for fatigue testing and is built to be more durable than the Hybrid III. The Hybrid III is designed to show high biofidelity during crash testing. The difference in the design emphasis results in the ANSI/RESNA WTD's having a stiffer skeletal structure than the Hybrid III; this makes the WTD more responsive to high-frequency vibrations. Our results indicate that the WTD is suitable for studying wheelchair shocks and vibrations. This should assist manufac-

Table 5: Seat and Footrest Vibrations: Comparisons of Test Dummies at Different Octaves

Octave (Hz)	WTD	HTD	*P*
Seat Vibration Power ($m \cdot Hz^{-1} \cdot s^{-2}$)			
157–198	1.1±0.7	.56±0.6	.03
198–250	.68±0.4	.23±0.2	.01
250–315	.44±0.3	.14±0.1	.01
315–397	.26±0.2	.09±0.1	.007
397–500	.25±0.1	.08±0.1	.02
Footrest Vibration Power ($m \cdot Hz^{-1} \cdot s^{-2}$)			
157–198	2.2±1.4	1.1±0.8	.04
198–250	1.7±1.3	0.6±0.4	.004
250–315	1.2±0.9	0.4±0.2	.007
315–397	0.8±0.7	.14±0.1	.02
397–500	0.7±0.8	.09±0.1	.005

NOTE. Values are mean ± SD. Only octaves in which significant differences were observed are presented.

turers in the development of improved suspension wheelchairs and suspension forks.

Future studies need to examine whether SCFs, rear-suspension systems, or both increase the fatigue life of manual wheelchairs. Life-cycle cost analysis should also be used to determine the effect of the additional complexity (eg, more moving parts) on the long-term costs. Studies also need to be conducted with human subjects. However, there are several issues that need to be addressed before effective human studies are designed. Currently, the seat instrumentation is an aluminum plate with an accelerometer located under the ischial tuberosities. A wheelchair user would need to sit on a cushion to avoid injury. The cushion would confound the vibration and shock results. Any study involving human subjects would also need either to control for body mass or to determine a method to control for the need to tune the suspension for each individual. In addition, muscle function and spasticity may influence results.

CONCLUSION

SCFs can reduce the shock and vibration exposure to the user of a manual wheelchair. Consumers should give SCFs serious consideration, especially if they are active or experiencing chronic pain. Rear-suspension systems reduce some of the factors related to shock and vibration exposure, but they are not clearly superior to traditional wheelchair designs. Further study is needed to improve rear-suspension system designs. Our results indicate that shock and vibration can be studied with an ANSI/RESNA WTD, which should help manufacturers to accelerate the development of new wheelchair suspension systems.

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Suppliers

a. First Technology Safety Systems Inc, 47460 Galleon Dr, Plymouth, MI 48170.
b. Sunrise Medical Inc, 2382 Faraday Ave, Ste 200, Carlsbad, CA 92008.
c. Invacare Corp, 1 Invacare Way, Elyria, OH 44036-2125.
d. Everest & Jennings, 1100 Corporate Square Dr, St Louis, MO 63132.
e. Frog Legs Inc, PO Box 465, Vinton, IA 52349.
f. Crossbow Technology Inc, 41 E Daggett Dr, San Jose, CA 95134.
g. National Instruments Corp, 11500 N Mopac Expwy, Austin, TX 78759-3504.
h. Human Engineering Research Laboratories, 7180 Highland Dr, Pittsburgh, PA 15206.
i. The MathWorks Inc, 3 Apple Hill Dr, Natick, MA 01760-2098.
j. SAS Institute Inc, SAS Campus Dr, Cary, NC 27513.