ASSESSING THE RISK OF VIBRATION EXPOSURE DURING WHEELCHAIR PROPULSION

by

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Although the exposure to whole-body vibrations (WBV) has been shown to be detrimental to seated humans, the vibration levels to which wheelchair (WC) users are exposed to in their communities have not been thoroughly examined. Furthermore, some evidence suggests that the cushions used in WCs, the first line of protection, may amplify WBV, although conclusive evidence has not been presented in the literature. The purpose of this work was twofold. First, to evaluate and compare the transmissibility of commercially WC cushions with two laboratory test methods: (1) direct measurement of transmissibility while human subjects propelled a WC over a road course with different cushions and (2) characterization of cushions with a material testing system (MTS) combined with mathematical models of the apparent mass of the human body. Second, to evaluate WBV exposure to WC users in their communities using ISO 2631-1 methods, and determine whether exposure levels are correlated with WC type and/or back pain, which is a physiological symptom of WBV exposures. Results showed that although dynamic characterization of WC cushions is possible with an MTS, the results did not correlate well with the transmissibility obtained in the WC road course. Significant differences were found for transmissibility among the cushions tested, with the air-based cushions having lower transmissibility than the foam- or gel-based cushions. All WC users who participated in this community-based trial were continuously exposed to WBV levels that were within and above the health caution zone specified by ISO 2631-1 during their day-to-day activities. Our evidence

suggested that WCs with suspension did not significantly impact the WBV transmitted to WC users. Finally, we found that WC users are exposed to other risk factors to LBP such as prolonged sitting and transfers. WBV exposure to WC users may be an important contributor to LBP as it has been shown to exceed international standards. Suspension systems need to be improved to reduce vibrations transmitted to the users. More research is needed to understand the interplay between posture, WC configuration, and WBV.

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PREFACE

I would like to thank everyone that has supported me through this process including the faculty and staff of the Human Engineering Research Laboratories for their continuing support, especially Dr. Jon Pearlman for all of his teaching, guidance, and support in this project. Thank you to the members of my committee: Dr. Rory Cooper and Dr. Michael Boninger. I especially appreciate everyone who helped me with my subject testing. Thank you to all the study participants. I want to thank my friends in Pittsburgh that have made this town my home away from home. I want to thank my family for their constant support and inspiration, and Alberto for all his support and love.

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Key words: dynamic stiffness, low back pain, manual wheelchair propulsion, manual wheelchair users, mathematical model, neck pain, suspension wheelchair, transmissibility, vibration, vibration dose value, wheelchair, wheelchair cushions, whole-body vibration.

Abbreviations: 3-D = three-dimensional, ANOVA = analysis of variance, a_{rms} = frequency-weighted r.m.s. acceleration, a_v = point vibration total value, a_w = frequency-weighted acceleration, cf = crest factor, CI = confidence interval, FSR = Force sensitive resistor, ISO = International Organization for Standardization, LBP = Low back pain, MANOVA = Multiple analysis of variance, Max = Maximum, MDL = Manual wheelchair datalogger, MTS = material testing system, NDI = Neck disability index, NP = Neck pain, OR = odds ratio, ODOF = one-degree-of-freedom, PSD = power spectral density, SAE = Society of Automotive Engineers, SCI = Spinal cord injury, SD = Standard deviation, SE = Standard error, r.m.s = root-mean-square, TDOF = two-degree-of-freedom, VA = Department of Veterans Affairs, VAS = Visual analog scale, VAPHS = VA Pittsburgh Healthcare System, VDL = Vibration datalogger, VDV = vibration dose value, VDV_v = Poin vibration dose value, WBV = whole-body vibration, WC = wheelchair, WRC = WC road course.

1.0 INTRODUCTION

1.1 MORBIDITY, COST AND THE BURDEN OF LOW BACK PAIN

Although many efforts have been made in the past years to reduce incidence rate and claims accounting for occupational low back pain (LBP), it is still one of the most disabling and major socioeconomic problem in the United States working population [1-3]. It affects hundreds of millions of people around the world; about 80 to 85% of people at some stage in their life; and about 30% of employees in the US each year [3-6]. In 1995, 1.8 per 100 workers in the US complained about LBP and 8.8 billion was spent on just in medical payments and indemnities related with LBP [2]. LBP is a major cause of job absence, a major cause for low job satisfaction, a common reason for people claiming disability pension, and the second most common reason for visiting a doctor; it constitutes up to 10-20% of primary care consultations [5, 7].

LBP is usually localized below the line of the twelfth rib and above the inferior gluteal folds [5] and can include pain in the legs. Only about 10% of LBP cases can be related to a specific cause whereas the rest are non-specific in cause [5]. Chronic LBP can last more than 3 months with persistent pain and symptoms which could restrict mobility, affect sleep, activities of daily living and leisure [5]. Until recently, the burden of LBP had been described in terms of

morbidity and cost; however, more recently studies had been paying more attention to the effect of LBP on disability and the individual's quality of life [6].

LBP is a major cause of years lived with disability around the world [5]. Moreover, it affects social functioning and mental health thereby, reducing the individual's quality of life. A recent study developing a conceptual and measurement model of the burden of LBP from the individual's perspective identified four major clusters that group the individual's most affected areas of life when living with LBP [6]. These areas are related to the reaction of others and the individual's psychological state, the effort of living with LBP that involves changes in lifestyle, the people's interactions with societal institutions such as workplaces and treatment services, and the effect of treatment and health states. For instance, an individual suffering from LBP will experience loss of enjoyment of life, low self-stem, loss of roles, feeling of helpless, irritation, anger and frustration, worry about the future and fear of pain [6]. They will also have difficulty taking care of other health issues, gain weight, feel tired, be left out of family activities, wrongly considered lazy by others, and lose friends [6]. Regarding the physical area, individuals with LBP may have difficulty functioning outside home, performing activities of daily living, self-care, and leisure [6]. They may experience frustration with treatment services and medical professionals, in addition to the economic burden of medicine, treatment, and care givers [6]. Furthermore, the challenges imposed on employment can make it difficult to get a paid job, reduced employment options, challenges to perform job functions, reduced income and thereby resulting in poverty [6].

1.2 RISK FACTORS TO LOW BACK PAIN

Because many studies have demonstrating that LBP is a growing public health concern; and it does not only impact on the individuals physical, psychological, social and economic functioning but also impacts their families, communities, industries and governments [6, 8], many studies have been carried out to help understand the epidemiology of LBP and have identified many physical, psychological and occupational risk factors have been identified. LBP has been associated with age, smoking, excessive body weight and physical fitness [5]. Among psychological factors that influence the occurrence of LBP are anxiety, depression, emotional stability, and exaggerated outward display of pain [5]. Prolonged seating, awkward postures, vibration exposure, heavy-object lifting, working with hands above shoulder level, bending, twisting, pulling, pushing, as well as job dissatisfaction have been pointed out as the most important occupational risks factors contributing to LBP [1, 3-5, 9]. Prevalence of LBP has been shown to be higher in occupational groups that spent more than four hours per day sitting and engaging in awkward postures (i.e. non-neutral trunk posture such as bending forward, slouching and/or twisting on the trunk) [3]. Furthermore, many studies have reported a strong significant association between LBP and awkward postures while sitting. A person who seats with a non-neutral spine position is 2 to 10 times at higher risk to LBP than a person who does not [3]. Non-natural body positions may increase intradiscal pressure thereby increasing the risk to damage the spine [3]. Combining these exposure factors with vibration exposure significantly increases the risk for LBP [3, 10].

Vibration transmitted by supporting surfaces to the entire human body is referred as whole-body vibration (WBV) [11]. In a seated position, vibration is transmitted from supporting surfaces at the feet, the buttocks and back of the person significantly affecting the magnitude of

the loads transmitted. WBV exposure has been found to have an effect on humans' health, activities, and comfort [12]. In addition, there is compelling evidence that WBV exposure is a risk factor to spinal disorders, excessive muscle fatigue, and damage to the connected nerves [1, 9, 13-15]. Losses of the spinal height and overcompensated spinal muscular responses also have been observed during WBV exposure thereby increasing the risk to LBP [13, 14]. Many factors can affect the amount of transmitted WBV such as amplitude and frequency of the vibration exposure, as well as the body position [3, 13]. For instance, amplitude increments above 4 m/s^2 and frequency of shock-type vibration exposure lead to an exponential-like amplification of the vibration of the body at the lumbar, cervical and forehead, with the first being the most affected region during WBV [3, 15-17]. WBV with frequency content near the human body's resonance frequency transmit motion in excess to the input [12]. Duration of the exposure also plays an important role in WBV association with LBP due to the vibration's cumulative effect [3, 15].

Motor vehicle drivers and heavy equipment workers are at considerable higher risk of LBP and spinal disorders due to WBV exposition [1, 3, 14] combined with their seated posture. To address these risks, WBV on this population has been extensively studied. Others have reported pathological changes in the spine of motor vehicle drivers after WBV [13, 14]. For instance, a study carried out by Pope et al (1998) [13] in tractor drivers found that increasing driving hours also increased prevalence of spinal disorders among the participants. In this study, 61% of drivers who drove an average of 700 hours per year had pathologic changes in the spine, between 700 and 1200 driving hours per year led to 68%, and more than 1200 hours resulted in 94% prevalence of spinal disorders [13]. As a result, motor vehicle driving has been identified as the most common way to transmit WBV [18]. Motor vehicle drivers report higher prevalence of LBP than the general population [19]. Although many studies have indicated positive association

between LBP and WBV, a dose-response relationship has not been yet established probably due to its multifactorial etiology [14, 19, 20]. To protect motor vehicle drivers against injuries resulting from WBV exposure the International Standards Organization (ISO) has developed guidance to evaluate occupational exposure levels of vibration. These standards prescribe how to measure and assess vibration exposure and provide a "health guidance zone," (see Figure 1) which conveys the potential for health consequences based on the duration and amplitude of the WBV exposure [11].

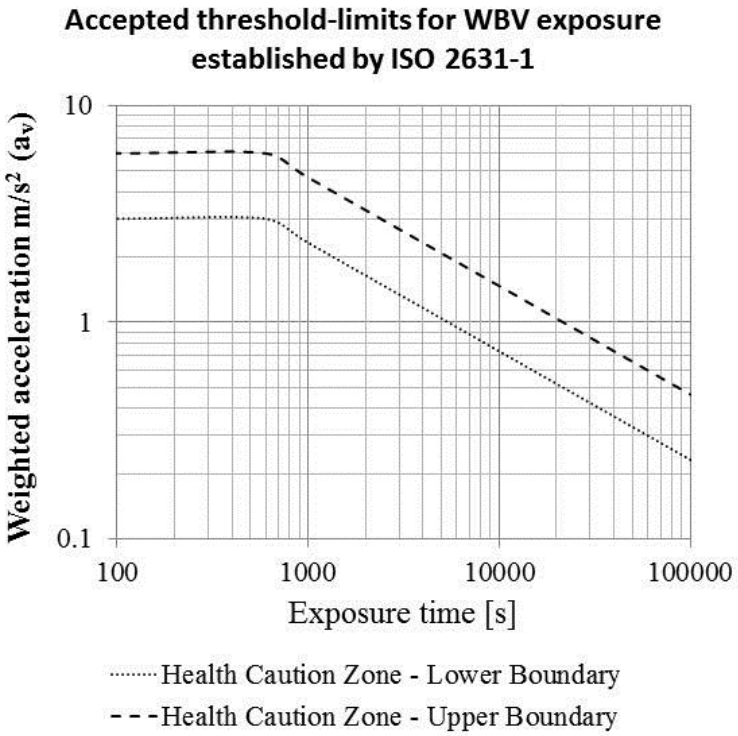

Figure 1. Accepted threshold-limits for WBV exposure established by ISO 2631-1.

1.3 WHEELCHAIR USERS: ANOTHER POPULATION AFFECTED BY LOW BACK PAIN

Although most of the research regarding LBP and seated WBV have been conducted on motor vehicle operators, it is logical that other population groups with high LBP prevalence rates are exposed to similar risks. For instance, LBP prevalence rate in WC users is significantly higher than in the general population [21]. Studies carried out with WC users report that approximately 66% of the participants had back or neck pain, although this prevalence may be higher WC users with severe pain may had limited their participation in the study [21, 22]. These studies also reported that 60 percent of their subjects visited the doctor because of their pain and 40 percent of them had to limit their daily activities.

Risk factors such as prolonged seating, poor posture, and exposure to WBV are highly present among WC users as indicated in the literature [23-28]. WC users rely on their WC to perform the majority of their activities, and may sit more than 20 hours a day [23]. In addition, individuals with spinal cord injury (SCI) have been found to have a higher incidence and greater degree of spinal deformities (i.e. kyphosis and scoliosis), developed during the first years after the injury, than people without disabilities [24]. Lack of trunk control as well as lack of an accessible home and work environment may lead to unnatural body positions, such as slouching, twisting, stretching, and bending forward of the lower back [25, 28]. Furthermore, WC users are exposed to levels of vibration that enter the health guidance zone [26, 27, 29-31]. VanSickle et al. studied the dynamic reaction forces and moments applied to WC frames during laboratory and field tests [26]. Three important conclusions were drawn from this study: (1) both the WC and the user are exposed to infrequent but high-magnitude shocks; (2) the rider seems to absorb

most of this energy; and (3) WC users are exposed to some high-impact vibrations of 50 m/s^2 or greater each day, which exceed the safety threshold indicated by the ISO standards.

1.4 VIBRATION-DAMPING STRATEGIES USED BY WC USERS AND THEIR LIMITATIONS

Although there is reasonable evidence that seated WBV exposure likely contributes to back and neck pain, most engineering and research efforts have been focused on the improvement of WC frame design to reduce weight and meet with fatigue, stability and cost-effectiveness standards [32, 33]. Even WC cushions have been designed to improve postural support and pressure relief [29].

1.4.1 Commercially available wheelchair cushions

In terms of vibration transmission by WC cushions, a study carried out to evaluate different seating systems showed that individuals are not being provided with the most effective WC cushion and that all seating-human systems included in the study tended to amplify vibrations in the frequency range most harmful to humans defined by ISO 2631-1 [29]. This study also suggests that cushion characteristics such as wear, age, material properties, and configuration significantly influence the transmission of vibrations. Other studies that compared the ability of seat cushions to minimize vibration exposure during manual WC propulsion support these results. For instance, Wolf et al. and DiGiovine et al. suggest that cushions made with a combination of foam and air transmit fewer impact and cyclic vibrations [34, 35]. These

studies provide important findings about how vibrations are transmitted. They found that high-impact shocks were reduced whereas cyclic vibrations were amplified when accelerations at the head were compared with those measured under the seat cushion. These results demonstrated that the human body and seating system absorb the energy of high-impact vibrations, while cyclic vibrations in the same frequency range as the natural frequencies of the human body appear to be amplified [35].

Up to now, studies have been performed by considering the seating system and the human body as one mechanical unit. This method may underestimate the energy absorbed by the human body because vibration measurements are collected below the seat and in some cases at the head (see Figure 2) [29, 34, 35]. Other methodologies have tried to characterize the dynamic response of the seating systems independently of the human body [12, 36]. However, they require a human subject to be seated on an indenter of top of the cushion while exposing it to vibration, which could be dangerous for subjects without sensation.

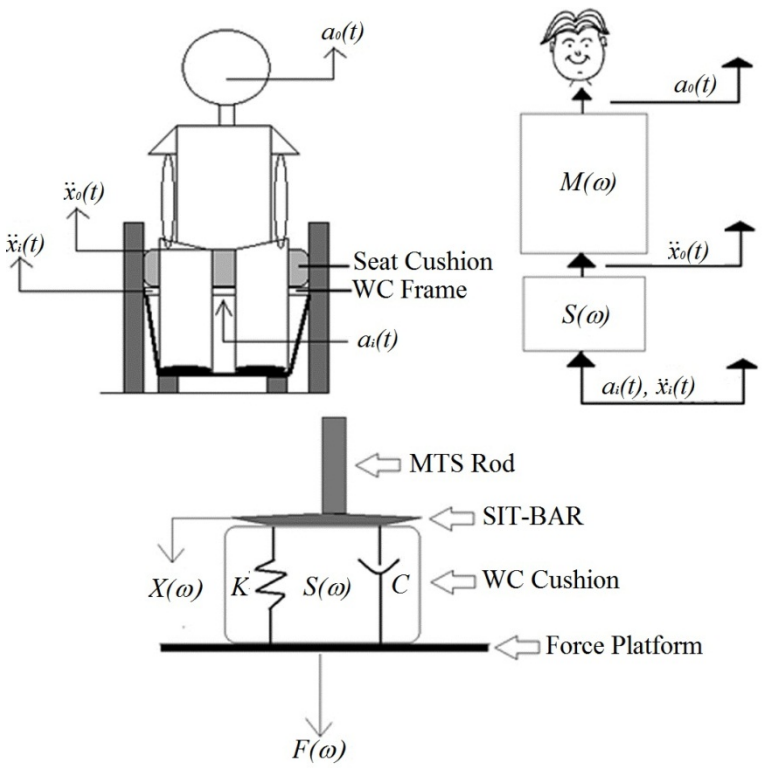

Figure 2. Left, comparison of (1) independent dynamic characterization of the wheelchair (WC) cushion represented by input acceleration below cushion ($x_i(t)$) under output acceleration at cushion surface ($x_o(t)$); and (2) current methods to measure vibration transmissibility represented by input acceleration below cushion ($a_i(t)$) under output acceleration at head ($a_o(t)$). Right, free-body diagram of comparison on left. Bottom, dynamic stiffness model $S(\omega)$ of cushion when measured with SIT-BAR attached to material testing system (MTS) and force platform.

Methods have also been developed to characterize seated transmissibility with a laboratory technique that includes mathematical models to represent the dynamic response of the human body [41–42]. With this approach, the dynamic behavior of the WC cushions can be characterized with laboratory equipment without risk for individuals. If transmissibility measured with human subjects was shown to be similar to that measured according to a mathematical model, future work characterizing cushions could be performed without human subjects, thereby reducing the complexity and cost of these studies.

1.4.2 Suspension wheelchairs

It has been documented during simulated laboratory and 4-8 hour field tests that WC users are exposed to WBV that exceed exposure limits set by ISO-2631 and that riders seem to be absorbing most of this energy [26, 27], thereby increasing the risk of spine injuries in WC users. The potential discomfort caused by prolonged WC riding and vibration exposure has motivated the development of WC suspension systems to reduce external reaction forces transmitted to WC users during daily WC use [28]. Suspension systems can be composed by coil springs attached to the WC frame, single spring-damper units supporting the WC seat, or polymer-based units supporting each wheel [28]. A few studies have been carried out to evaluate the vibration-reduction effectiveness of rear-wheel suspension and shock absorbing caster forks [28, 30, 37]. The results of these studies have shown that suspension casters can significantly reduce peak accelerations transmitted to users (at the seat and footrest) and that rear-wheel suspension systems do reduce some of these vibrations, although they do not outperform traditional frame designs and still transmit vibration in the frequency range most harmful for humans [30]. Although the vibration-dampening characteristics of WC suspension components might be satisfactory during simulated laboratory-tests [28], their performance in real-world conditions is currently unknown.

Given the fact that pain relates to lower quality of life and reduced function [24], more studies are needed to investigate whether WBV exposure during WC riding is linked to pain. To our knowledge, only controlled laboratory tests or short (4-8 hour) community-based trials [26, 27, 31, 38], with small exposure duration data and sometimes static WC riders, have been performed to evaluate levels of vibration and the effectiveness of WC suspension systems. These laboratory studies and 4-8 hours of exposure data in the community are not likely to provide a

full picture of vibration exposure of a person because of day-to-day variations and the lack of real environmental factors. There is a need to collect more information about WBV exposure levels during daily mobility-related activities in the community (i.e. real world settings) for representative periods of time, the relationship between LBP and neck pain (NP), and whether WC frame design has any impact on this relationship.

1.5 PROPOSED WORK

The first part of this work presents and evaluates an alternative methodology to characterize the vibration-dampening characteristics of WC cushions independently from the human body. This work characterized and compared transmissibility of seat cushions in two ways: by directly measuring transmissibility while nondisabled individuals propelled a manual WC through a road course and by using a material testing system (MTS) to characterize cushion stiffness, which was then input into a mathematical model of the human-cushion system. We hypothesized that transmissibility measured with an MTS and in human subjects would be correlated, which would suggest that future characterization could be performed with the MTS alone. We also hypothesized that significant differences would exist in the transmissibility of commercially available WC cushions in the range of harmful vibrations (4–12 Hz). This would be of great importance to clinicians when making decisions for cushion selection and prescription for individuals looking for greater comfort and reduced vibration effects (but minding pressure sore prevention).

The second part of this study uses current ISO techniques to evaluate the health risk associated with WBV exposure to WC users in real-world environments and different types of

WC frames. It also attempts to determine if a correlation between WBV exposure level and low back/neck pain exists. We hypothesized that 1) suspension systems would have an effect on vibration transmitted to WC users during propulsion at real-world environments; 2) WBV exposure to WC users in real-world environments would exceed safe vibration level thresholds set by IS0 2631-1 (as shown in laboratory and short field trials [27]); and 3) there would be a significant correlation between the presence of pain and the exposure level of WBV. The results of this work could potentially motivate the development of more effective WC suspension systems that protect WC users from the risk associated with WBV exposure in the community.

2.0 THEORY AND METHODS

2.1 DYNAMIC STIFFNESS AND TRANSMISSIBILITY OF COMMERCIALLY AVAILABLE WHEELCHAIR CUSHIONS USING A LABORATORY TEST METHOD

2.1.1 Cushions

Seven commercially available WC cushions were selected based on advice from clinicians from the Department of Veterans Affairs (VA) Pittsburgh Healthcare System (VAPHS) Wheelchair and Seating Clinic and the Center for Assistive Technology at the University of Pittsburgh. New WC cushions were borrowed from the VAPHS Wheelchair and Seating Clinic and are listed in Table 1 and shown in Figure 3.

Laboratory measurements of seated vibration transmissibility were undertaken with data from both human subjects and WC cushion dynamic characterization from the MTS (858 Bionix II, actuator model 244.12lf, 6 in. vertical stroke, pump model 505.11, MTS Systems Corporation; Eden Prairie, Minnesota).

Table 1. Description of seat cushions

Model	Manufacturer	Description
Meridian Wave	Varilite; Seattle, Washington	Solid foam base and dual chamber of air-foam floatation.
ROHO HIGH PROFILE	The ROHO Group; Belleville, Illinois	Single compartment of high interconnected air cells.
ROHO LOW PROFILE	The ROHO Group; Belleville, Illinois	Single compartment of low interconnected air cells.
Jay J2 Deep Contour	Sunrise Medical; Stourbridge, United Kingdom	Gel compartment over contoured foam base.
Vector with Vicair Technology	The Comfort Company; Bozeman, Montana	Individual air cells inserted into cushion shell that can be removed or added to adjust body positioning and pressure distribution.
Comfort Mate Foam	Invacare; Elyria, Ohio (Pin-dot)	Molded polyurethane foam cushion.
Zoombang Protective Gear with Foam	Invacare; Elyria, Ohio (Zoombang)	Mat of double layer of capsules filled with viscoelastic polymer over Invacare Comfort Mate Foam cushion.

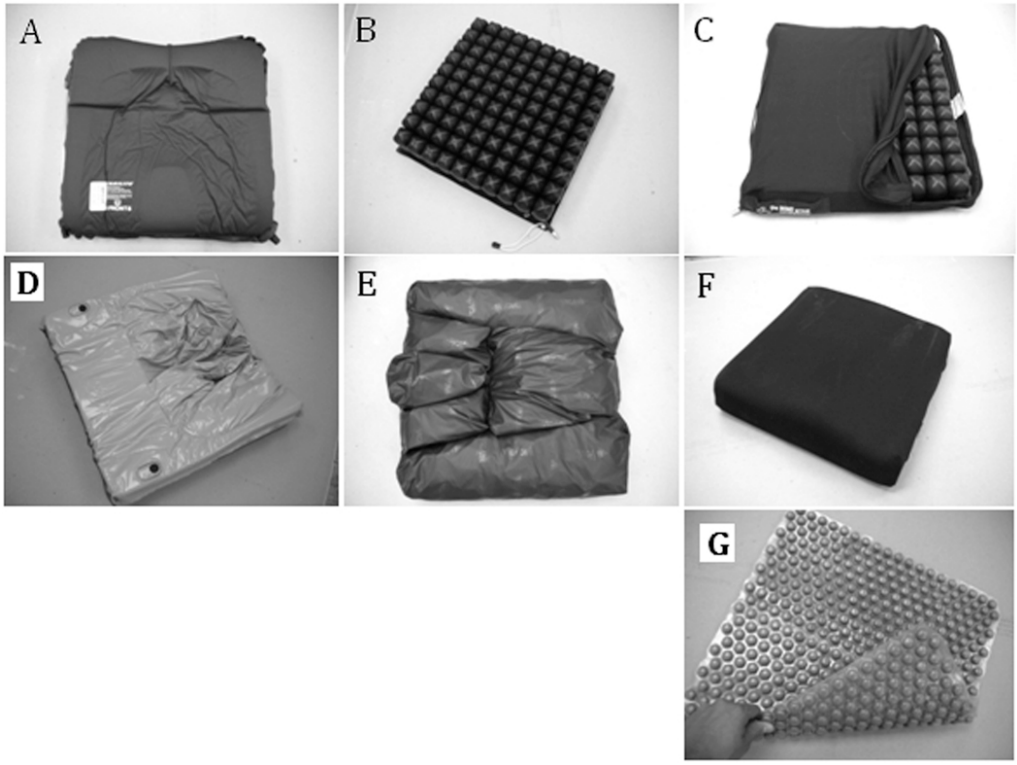

Figure 3. Wheelchair cushions: (a) Meridian Wave, (b) ROHO HIGH PROFILE, (c) ROHO LOW PROFILE, (d) Jay J2 Deep Contour, (e) Vector with Vicair Technology, (f) Comfort Mate Foam, and (g) Zoombang Protective Gear with Foam.

2.1.2 Dynamic characterization of wheelchair cushions

It is possible to estimate seated transmissibility by determining the seat's dynamic stiffness [12]. The dynamic stiffness, *S(ω)*, of the seat is the complex ratio of the force to displacement in the frequency domain (expressed in hertz) and is given by Equation 1:

$$S(\omega) = \frac{F(\omega)}{x(\omega)}, \quad (1)$$

where $F(\omega)$ is the force transmitted by the seat and $x(\omega)$ is the relative displacement of the seat cushion. $x(\omega)$ can also be expressed in terms of its second differentiation $\ddot{x}(\omega)$ (i.e., acceleration):

$$x(\omega) = \omega^{-2}\,\ddot{x}(\omega). \tag{2}$$

Therefore, the dynamic stiffness expressed in terms of the force transmitted by the seat and the acceleration recorded below ($\ddot{x}_b(\omega)$) and above ($\ddot{x}_s(\omega)$) the seat cushion is

$$S(\omega) = \frac{F(\omega)}{\omega^{-2}[\ddot{x}_s(\omega) - \ddot{x}_b(\omega)]}. \tag{3}$$

The dynamic stiffness can also be determined by making the indenter vibrate while the cushion is static [12]. The seat dynamic characterization with the moving indenter may be represented by the function

$$S(\omega) = \frac{F(\omega)}{\omega^{-2}\ddot{x}_s(\omega)} = K + C\omega i, \tag{4}$$

where K is the static stiffness and C is the viscous damping of the seat. Using Equation 4, we can calculate the K and C for each cushion with curve-fitting methods. Note that K and C may vary based on the preload of the indenter.

The dynamic stiffness of each cushion in this study was determined by exposing each cushion to 100 s of random vibration (±4 mm maximum displacement, peak-to-peak acceleration with a flat power spectral density [PSD] over the range 0.5–20.0 Hz) under six different preload

conditions (300–800 N in 100 N increments) with the MTS. The loads and vibrations were applied to the seat surface with a SIT-BAR indenter [12] (Figure 4) attached to the MTS actuator.

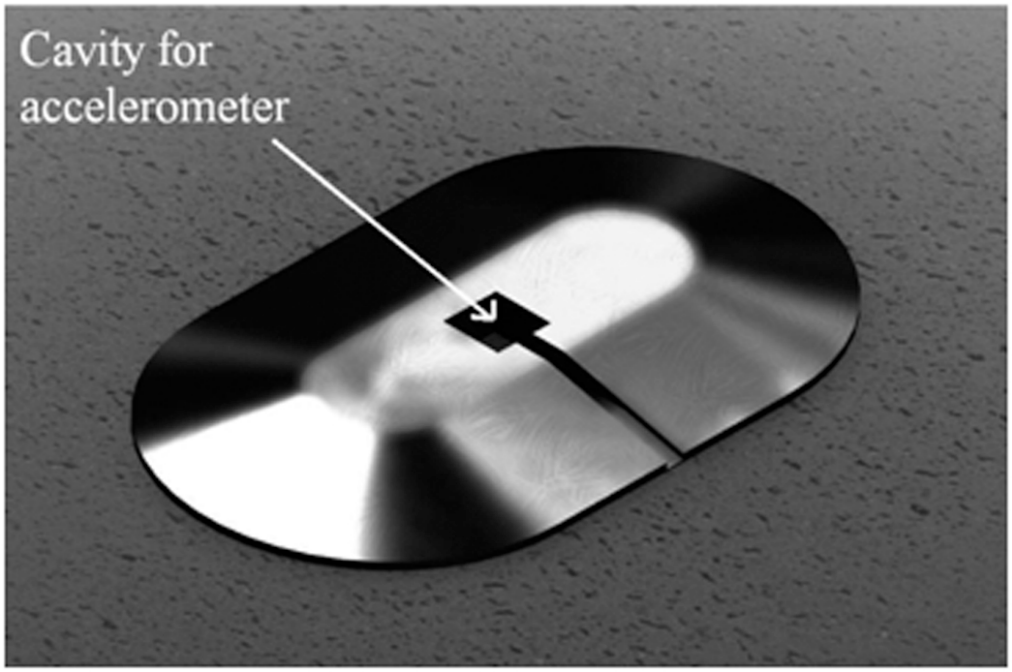

Figure 4. SIT-BAR indenter used to load and vibrate wheelchair cushions.

The reaction force below the cushion was measured with a 4 550 Bertec force plate (Bertec Corporation; Columbus, Ohio) that was rigidly attached to the MTS base. Acceleration of the SIT-BAR was measured with a three-dimensional (3-D) accelerometer (CXL10LP3, ±10 g, 0–100 Hz, Moog Crossbow; Milpitas, California). Force and acceleration measurements were acquired at 200 Hz with an analog-to-digital acquisition card (DAQCardTM-6024E, National Instruments Corporation; Austin, Texas) and LabView Signal Express software (National Instruments Corporation). An algorithm was developed in MATLAB (The MathWorks, Inc; Natick, Massachusetts) to estimate the dynamic stiffness transfer function (Equation 4) from the

acquired signals in the frequency domain and to compute the K and C dynamic parameters from the modulus of the dynamic stiffness: $\vec{S} = \sqrt{K^2 + (C\omega)^2}$.

Nonlinear least-squares curve-fitting methods were used with an optimization algorithm (parameters were obtained with the MATLAB Curve Fit Toolbox).

Figure 5 shows the experimental setup, and the diagram on the bottom of Figure 2 conveys the mathematical representation of the system. Measurement of dynamic stiffness was performed three times per cushion at each preload.

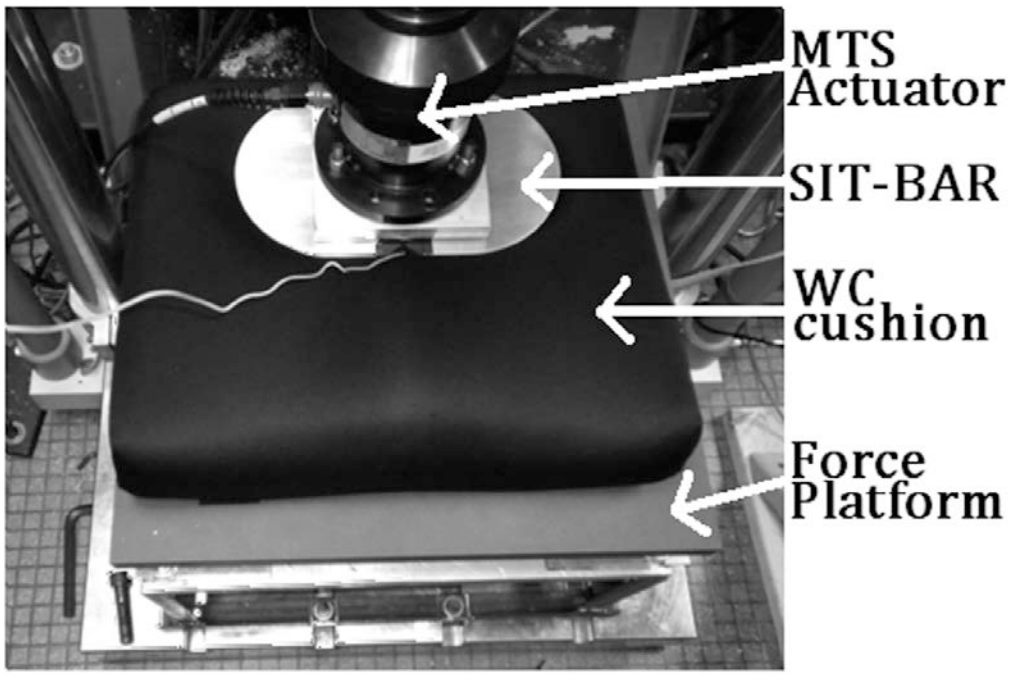

Figure 5. Experimental configuration to measure dynamic stiffness of wheelchair (WC) cushions. MTS = material testing system.

2.1.3 Measurement of seated transmissibility with human subjects

Measurements of seated transmissibility were obtained for each seat cushion during field tests with 14 nondisabled human subjects. Informed consent and demographic information (sex, age, height, and weight) were gathered before each participant performed the field test (mean ± standard deviation weight: 73.15 ± 12.73 kg and height: 1.71 ± 0.11 m). Each person was asked to propel the same rigid frame WC (Quickie GP, Sunrise Medical; Stourbridge, United Kingdom) over a WC road course (WRC) to simulate activities of daily living while seated on each of the seven cushions. Each seat cushion was adjusted for each participant as indicated by the manufacturer. Measurements of accelerations at the WC-cushion interface and at the cushion-human interface were recorded using two 3-D accelerometers (Moog Crossbow) both located at the midline of the body under the ischial tuberosities. One accelerometer was placed under the cushion and attached to the WC frame using a 5/16 in.-thick aluminum seat pan (Figure 6). The second accelerometer was mounted in a SIT-BAR indenter and placed above the cushion for the subject to sit on (Figure 6). The SIT-BAR and seat-pan instrumentation were positioned and aligned as defined in ISO 2631-1 and ISO 10326-1 [11, 39]. Data acquisition was performed using the same equipment and at the same frequency as described in the previous MTS section.

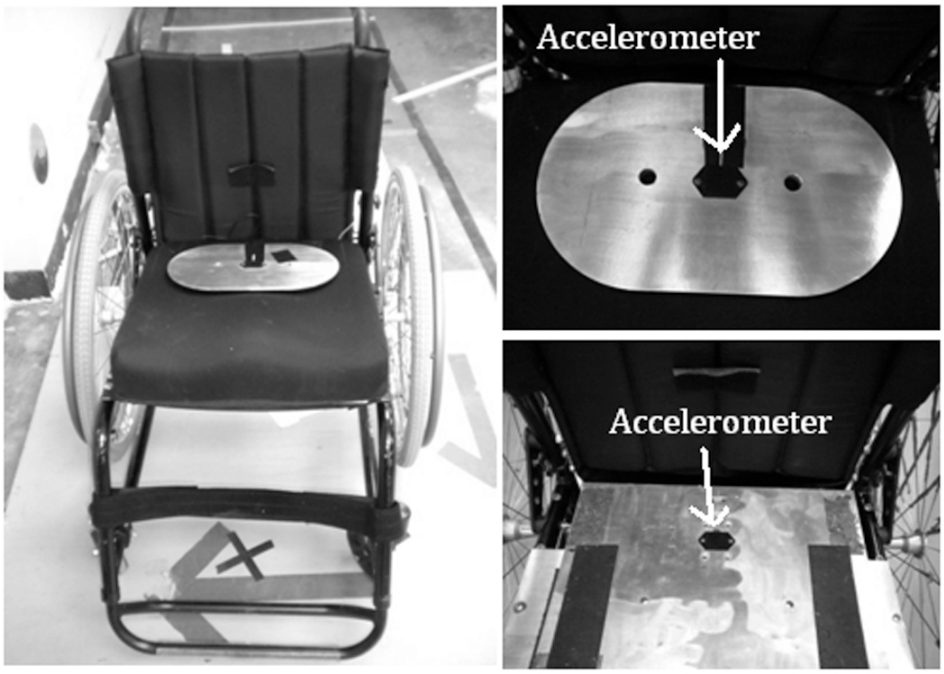

Figure 6. (a) Manual wheelchair (WC) with Comfort Mate Foam cushion and SIT-BAR. (b) SIT-BAR with accelerometer attached. (c) Aluminum pan attached to WC frame with fixed accelerometer.

The WRC was created to replicate obstacles that WC users encounter in activities of daily living [26, 27]. Obstacles included in the WRC were two 5 cm curb descents, a dimple strip mat, three sine-wave bumps (2.5, 5.0, and 7.5 cm, respectively), a simulated door threshold, industrial carpet, and a rumble bump mat. The WRC and the obstacles are shown in Figure 7.

Figure 7. Wheelchair road course.

Individuals propelled the WC in their most comfortable erect posture and at their self-selected velocity. The trial was repeated three times by every subject for each of the seven randomly presented cushions. The participants were allowed to rest or get out of the WC between trials.

Prior to data collection, each subject was trained in the WRC at least once to familiarize them with the obstacles.

Figure 2 shows the WC-seat-human system for this experiment. Seated transmissibility was calculated by the given equation:

$$T(\omega) = \frac{\ddot{x}_o(\omega)}{\ddot{x}_i(\omega)}, \quad (5)$$

where $T(\omega)$ is the seated transmissibility, $\ddot{x}_o(\omega)$ is the output acceleration measured under the cushion, and $\ddot{x}_i(\omega)$ is the input acceleration measured above the cushion.

An algorithm written in MATLAB was developed to estimate the averaged transfer function (i.e., seated transmissibility) from the input and output acceleration signals for all the subjects and for each type of cushion at each sensor site (WC frame and seat surface).

2.1.4 Mathematical models of seat transmissibility

Due to the dynamic interplay between the human body and the cushion, cushion vibration transmissibility from the MTS data must be calculated taking into account the response of the human body seated on it [12]. Wei and Griffin have developed different mathematical models for the human body that can be used to represent a person seated on a dynamic seating system (when vibrations occur) [40]. To predict seat vibration transmissibility without exposing human subjects to vibration, we used two mathematical models of the apparent mass of the human body (i.e., one-degree-of-freedom [ODOF] and two-degree-of-freedom [TDOF] models) in conjunction with the mechanical model of the seating system characterized previously and illustrated in the bottom of Figure 2. This methodology has already been shown to be appropriate for investigating seat dynamic performance from separate measurements of seat dynamic stiffness and the apparent mass of the human body [41]. The ODOF and TDOF models used to

represent the cushion-human body response to vertical vibrations are shown in Figure 8. Both models are represented by a support structure, m, and one or two mass-spring-damper systems (i.e., K_1, C_1, m_1 and/or K_2, C_2, m_2) that represent different parts of the human body supported by tissues. The model parameters K_1, C_1, K^2, C_2, m, m_1, and m_2 (obtained by Wei and Griffin from the analysis of measurements of the apparent masses of 60 persons [40, 41]) are shown in Table 2. The K and C represent the seat dynamic parameters obtained from the characterization of each seat cushion in the first section of this study.

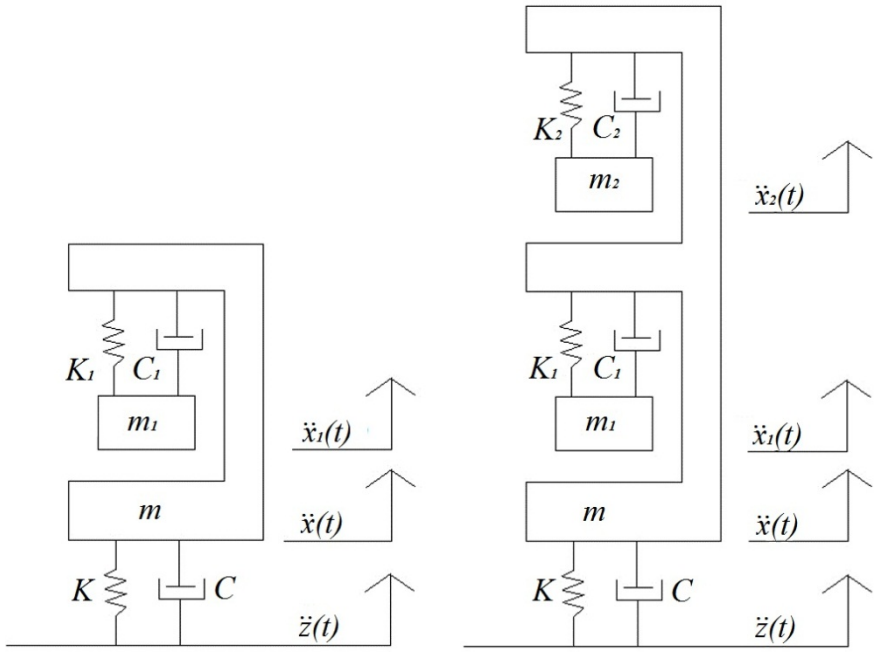

Figure 8. (a) One-degree-of-freedom model of apparent mass of human body (m_1). (b) Two-degree-of-freedom model of apparent mass of human body (m_1 and m_2). $\ddot{x}(t)$, $\ddot{x}_1(t)$, and $\ddot{x}_2(t)$ represent acceleration of each mass, respectively; and $\ddot{z}(t)$ acceleration and base of seat cushion.

Table 2. Parameters of one-degree-of-freedom (ODOF) model and two-degree-of-freedom (TDOF) model of apparent mass of human body.

Mathematical model	K_1(N/m)	C_1(Ns/m)	K_2(N/m)	C_1(Ns/m)	m(kg)	m_1(kg)	m_2(kg)
ODOF	44,130	1,485	-	-	7.8	43.4	-
TDOF	35,776	761	38,374	458	6.7	33.4	10.7

The response of the ODOF model shown in Figure 8 is given by

$$m_1\ddot{x}_1 + K_1(x_1 - x) + C_1(\dot{x}_1 - \dot{x}) = 0 \ , \quad (6) \text{ and}$$

$$m_1\ddot{x}_1 + m\ddot{x} = K(z - x) + C(\dot{z} - \dot{x}); \quad (7)$$

and the response of the TDOF model also shown in Figure 8 is given by

$$m_1\ddot{x}_1 + K_1(x_1 - x) + C_1(\dot{x}_1 - \dot{x}) = 0, \quad (8)$$

$$m_2\ddot{x}_2 + K_2(x_2 - x) + C_2(\dot{x}_2 - \dot{x}) = 0, \quad (9) \text{ and}$$

$$m\ddot{x} + m_1\ddot{x}_1 + m_2\ddot{x}_2 = K(z - x) + C(\dot{z} - \dot{x}). \quad (10)$$

The transmissibility of the seat cushion is given by the magnitude of the transfer function, $|H(\omega)|$, which can be derived from Equations 6 and 7 for the ODOF model [41] by

$$|H_{\mathrm{ODOF}}(\omega)| = \left|\frac{\ddot{x}(\omega)}{\ddot{z}(\omega)}\right| = \left|\frac{(A+Bi)}{(D+Ei)}\right| = \sqrt{\frac{(A^2+B^2)}{(D^2+E^2)}}; \quad (11)$$

and from Equations 8 to 10 for the TDOF model by

$$|H_{\mathrm{TDOF}}(\omega)| = \left|\frac{(F+Gi)}{\{(H+L)+(M+N)i\}}\right| = \sqrt{\frac{(F^2+G^2)}{(H+L)^2+(M+N)^2}}, \quad (12)$$

where *A*, *B*, *D*, *E*, *F*, *G*, *H*, *L*, *M*, and *N* are the polynomials of Equations 11 and 12 as defined in Figure 9 and simplified in the previous equations to improve readability.

For the ODOF model only, the mass was changed according to the averaged assumed sitting weight for all the subjects in the second section of this study. In other words, the term ($m + m_1$) of the polynomial *D* was made equal to 75 percent of the average total weight for all the subjects, as indicated in a previously cited study [40].

$A = KK_1 - (m_1K + CC_1)\omega^2$	$B = (C_1K + CK_1)\omega - m_1C\omega^3$	$D = (K - (m+m_1)\omega^2)K_1 + (mm_1\omega^2 - Km_1 - CC_1)\omega^2$	$E = (KC_1 + K_1C - (m_1C + mC_1 + m_1C_1)\omega^2)\omega$	$F = KP_1 - CP_2\omega$
$G = KP_2 - CP_1\omega$	$H = P_1P_5 - P_2C\omega - m_1K_1P_3\omega^2$	$L = m_1C_1C_2\omega^4 - (m_2K_2P_4\omega^2 - m_2C_1C_2\omega^4)$	$N = (m_2C_2P_4\omega^3 + m_2K_2C_1\omega^3)$	$P_1 = m_1m_2\omega^4 + K_1K_2 - (m_1K_2 + m_2K_1 + C_1C_2)\omega^2$
$P_2 = (C_1K_2 + C_2K_1)\omega - (m_1C_2 + m_2C_1)\omega^3$	$P_3 = K_2 - m_2\omega^2$	$P_4 = K_1 - m_1\omega^2$	$P_5 = K - m\omega^2$	

Figure 9. Definitions of polynomials found in Equations 11 and 12.

2.1.5 Vibration dose value of wheelchair cushions

Vibration dose value (VDV) is a vibration evaluation method defined by ISO 2631-1 [11] that is sensitive and useful for evaluating transient vibrations with occasional shocks. The VDV unit is meters per second to the power of 1.75 and is defined as

$$\mathrm{VDV} = \left\{\int_0^T [a_w(t)]^4 \; dt\right\}^{\frac{1}{4}}, \tag{13}$$

where $a_w(t)$ is the instantaneous frequency-weighted acceleration and T is the duration of the measurement.

Vibration frequency content and axis have different effects on health than on comfort, perception, and motion sickness. To appropriately evaluate the effects of vertical vibrations on the health of seated humans, we applied a frequency-weighting filter as indicated by ISO 2631-1 [11] to the measured input and output accelerations before VDVs were estimated for both vibration measurements. The input and output accelerations were collected at the WC frame below the cushion and between the person and the cushion with the SIT-BAR during the WRC test. The transmissibility of each cushion was calculated with the VDV as defined in Equation 14:

$$T = \frac{\mathrm{VDV}_o}{\mathrm{VDV}_i}, \tag{14}$$

where VDV_i is the VDV estimated with the input acceleration data and VDV_o is the VDV estimated with the output acceleration data, both estimated over the entire WRC.

2.1.6 Statistical analysis

After testing for normality, we performed two statistical analyses with repeated measures one-way analysis of variance (ANOVA) to evaluate whether differences existed (1) in the measured and estimated ODOF and TDOF transmissibility for all the seating systems and (2) in the estimated VDV transmissibility for all the seating systems. If significant differences were present, a Sidak post hoc analysis was performed for each ANOVA. Additionally, a correlation analysis was performed between predicted transmissibility obtained with the ODOF and TDOF models and the measured values of transmissibility with the WRC test.

2.2 HEALTH RISKS OF VIBRATION EXPOSURE TO WHEELCHAIR USERS IN THE COMMUNITY

2.2.1 Protocol

Individuals were asked to participate in an IRB-approved community-based study to record vibration exposure for at least two weeks. During the first week, subjects participated in a national veterans sporting event followed by an additional week in their home environment. At the beginning of the study; demographics, participation in physical activities, contact information, and manual WC information (make, model and frame style) were recorded. Subjects also answered a questionnaire previously used in another study [21] that captures data about the presence and characteristics of neck and back pain (long questionnaire). A vibration datalogger (VDL) and a manual wheelchair datalogger (MDL) were then mounted on the

subject's WC frame and wheel spokes, respectively. Participants were provided with a self-addressed, stamped package to return the VDL and MDL after at least two weeks had passed. The subjects were contacted by mail or phone to be given a shortened version of the NP and LBP questionnaire [21] two weeks after recruitment and reminded to remove and return the VDL and MDL.

2.2.2 Long Questionnaire

This questionnaire, which was previously used in another study [21], captures whether the WC users have experienced NP or LBP since one year after the onset of the condition that caused them to use a WC, within the past month, and within the past 24 hours. To determine the severity of NP experienced within the past 24 hours, individuals were asked to mark along a 10-cm line the intensity of their pain. This pain assessment tool ranges from no pain (0) to the worst pain imaginable (10) on a visual analog scale (VAS). The pain questionnaire also included the Neck Disability Index (NDI) to determine the effect of the intensity of pain on activities of daily living and concentration. The NDI contains ten items scaled from 0 to 5. A total score (expressed in percentage) greater than 40% represents severe disability due to pain. The questionnaire also collects information about aggravating factors to LBP and NP, participation in WC sports, and WC characteristics.

2.2.3 Instrumentation

Each participant agreed to have a custom built VDL [42] and a MDL [43] attached on their WC. The VDL is an instrument designed to record WBV levels that WC users are exposed to during

their day-to-day lives. The VDL can be located at different locations on the WC frame to prevent it from interfering with the user's activities and transfers. The VDL records acceleration at the supporting surfaces of the seated individuals where vibration was considered to enter the human body (seat, backrest and footrest) for two weeks. The VDL is composed of an ultralow-power microcontroller (Texas Instruments M430F2618T), a 2 GB memory card and two AA alkaline batteries. The microcontroller has a build-in Analog-to-Digital Converter with high frequency response (up to 12MHz). The VDL is connected to three tri-axial acceleration sensors (Analog Devices ADXL335, ±3g, 0.5-550Hz in z axis and 0.5-1600Hz in x and y axis) and a seat occupancy sensor (see Figure 10) to collect data only when the person is in the WC. The seat occupancy sensor's design involves the use of two interlinked force sensitive resistors (FSR), one on the right and one on the left sides on top of the WC tubes. The sensors are mounted on a rigid plastic strip and then attached on the WC seat. The FSR configuration was tested by attaching it to a sling seat of a WC and then placing different types of WC cushions on it to test for a valid response. An accelerometer was located at the WC seat below the seat cushion and midline beneath the ischial tuberosities to prevent damage to the skin. Whenever possible, the accelerometer at the backrest was centered at the interface between the subject's lumbar spine and the backrest of the WC. However, if the accelerometer caused discomfort, the subject was allowed to relocate the accelerometer at the same centered position but at the back of the backrest or behind the backrest's cushion. The seat and backrest accelerometers were covered with a soft and flexible material (3mm water proof neoprene) for comfort. To measure accelerations at the feet, an accelerometer was placed closely adjacent to the feet (usually within 10cm off the center of this area) at a non-removable portion of the footrest that was considered to

be a single rigid part. Figure 10 illustrates common location of the accelerometers attached to the participants' WC.

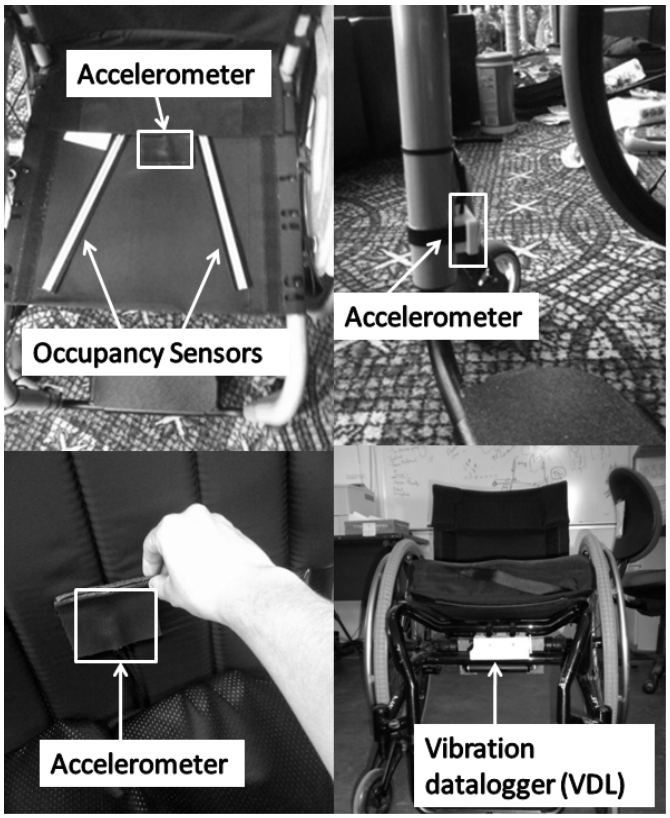

Figure 10. Vibration datalogger, occupancy sensors and accelerometers localization.

The accelerometers' direction of measurements were oriented relevant to the axes of the WC, which were assumed to be the same as the seated body and in accordance with the coordinate system for seated persons as defined in ISO 2631-1 [11]. In this right-handed orthogonal coordinate system, the x axis is positively oriented to the individuals forward, the y axis is positively oriented to the individuals left and the z axis is vertically oriented. A low-pass filter was implemented with a 0.5-22Hz (-3bB) bandwidth and a linear phase to limit

acceleration measurements up to the first two resonance frequencies of humans (0-20 Hz). The VDL collected acceleration data at 60Hz sampling rate and logged into the memory card. Only accelerations of the z and x axes at each surface were recorded.

The MDL used in this study was developed by researchers to objectively measure long-term WC-related activity (distance, speed, and continuous movement time) in real world environments [43]. It can collect and store data from the rotation of the WC rear-wheel, at a sample frequency of 10 Hz for more than 3 months. The MDL is 5cm in diameter by 3.8 cm in depth and can be mounted in the spokes of the wheel with no modifications and without interfering with the WC rider's activities. The MDL has been validated and used in previous studies [43].

2.2.4 Participants

The participants included individuals with a physical impairment who use a manual WC as their primary source of mobility. The inclusion criteria for the study were: no active pressure sore, 18 years old or older, and able to perform independent transfers. Subjects were recruited at the National Veterans (NV) Summer Sports Clinic 2010 in San Diego, CA; at the National Disabled Veterans (NDV) Winter Sports Clinic 2011 in Snowmass, CO; and at the NV Wheelchair Games 2011 in Pittsburgh, PA. All participants gave written informed consent previous to any data collection or subject's screening. Only data collected from individuals who showed activity during the 2-week data collection were included in the analysis.

2.2.5 Data reduction

Data recorded with the VDL was divided into individual files for each axis of measurement at each point of vibration transmission. A Matlab algorithm (The MathWorks, Inc; Natick, Massachusetts) was developed to analyze vibration levels per-day. In this algorithm, vibration data were frequency weighted according to standard vibration evaluation methodologies before performing any data reduction [11].

ISO 2631-1 only provides guidelines for the evaluation of the effect on health of vibration transmitted through the seat (in three orthogonal axis) and to the x-axis of the backrest. Due to the lack of evidence of the effect of vibration transmitted through the footrest and the other two orthogonal axis of the backrest, these directions of measurements were not included in the health assessment.

According to ISO 2631-1, two basic evaluation metrics must be included in any vibration assessment: the weighted root-mean-square (r.m.s) acceleration and the vibration crest factor.

The frequency-weighted r.m.s. acceleration, a_{rms}, is expressed in meter per second squared (m/s^2) and is calculated according to the following equation:

$$a_{rms} = \left[\frac{1}{T}\int_0^T a_w^2\,(t)dt\right]^{\frac{1}{2}} \qquad (1)$$

Where $a_w(t)$ is the frequency-weighted acceleration as a function of the time at each direction of measurement, and T is the duration of the measurement.

The crest factor, *cf*, is a metric used to determine whether a_{rms} alone is appropriate to describe the severity of the effects of the vibration on health, and is defined as the modulus of the maximum peak value of a_w to a_{rms} determined over T [11]. *cf* values greater than 9 indicates the

presence of occasional shocks and the need for an additional evaluation method such as the fourth power vibration dose value (VDV).

VDV is a shock sensitive vibration evaluation method defined by ISO 2631-1. The VDV unit is meters per second to the power of 1.75 and is defined as:

$$\mathrm{VDV} = \left\{\int_0^T [a_w(t)]^4 \, dt\right\}^{\frac{1}{4}} \qquad (2)$$

Where $a_w(t)$ is the frequency-weighted acceleration as a function of the time at each direction of measurement, and T is the duration of the measurement.

The use of VDV in this study was included because studies have shown that WC users are exposed to infrequent but high magnitude shocks and that the use of a_{rms} alone could underestimate its effects on the human body [16, 26]. VDVs for the seat surface and x-axis of the backrest were included in the analysis since computation of crest factor for the first nine participants of the study revealed crest factor values greater than 9 (mean = 19.86, SD = 9.38, n = 9).

The ISO 2631-1 indicates that vibration shall be evaluated independently along each axis of exposure. For the assessment of the health effects of a vibration at the seat surface, the vibration evaluated shall be the highest a_w determined in any seat axis. However, when vibration is comparable in two or more axes, it is permitted to combine the vibrations in more than one direction to perform the assessment.

To combine vibrations measured in two directions (x and z axis), the point vibration total value, a_v, was calculated for the seat surface by the equation:

$$a_v = (k_x^2 a_{rmsx}^2 + k_z^2 a_{rmsz}^2)^{\frac{1}{2}} \qquad (3)$$

Where a_{rmsx} and a_{rmsz} are the a_{rms} each with respect to the orthogonal axes x and z, respectively; and k_x and k_z are the multiplying factors specified in ISO 2631-1 [11].

Point vibration total dose value, VDV_v, at the seat surface was calculated by substituting a_{rms} for the respective VDV of each direction of measurement.

Vibration exposure levels measured in two directions of the seat surface (z and x axis) were combined and reported as a_v and VDV_v at the seat surface as they were seen as comparable (i.e. the lowest vibration at any axis was at least 30% and sometimes was the same as the vibration in another axis of measurement).

Vibration exposure levels in the x-axis of the backrest were evaluated independently along this direction of measurement. Therefore, only a_{rms} and VDV were calculated for this axis of the backrest.

Data recorded with the MDL were decompressed and analyzed with a previously used Matlab algorithm to estimate mobility characteristics variables such as daily distance the WC user traveled, average daily speed, daily accumulated movement time, maximum distance traveled during a continuous movement and maximum time period of continuous movement [43]. Average daily speed provides an indication of the level of activity of the WC user in the real world environment [43]. Daily accumulated movement time refers to the total amount of time the WC user moved in their WC a given day. Maximum distance and maximum time period refers to those maximum values per day between consecutive stops.

2.2.6 Statistical analysis

Data collected from the VDL, MDL and the pain questionnaires were analyzed using descriptive statistics, mean and standard deviation (SD) or median and standard error (SE) for data at the interval level, and frequencies for categorical data.

After checking for assumptions, Mann-Whitney tests were performed to test whether significant differences in age and length of time of manual WC use existed among LBP and no LBP in the past month groups. Fisher's exact test and the Likelihood ratio were used to determine whether gender and type of WC frame (folding, rigid, and suspension), respectively, were significantly different among the same groups. Independent means t-tests, Fisher's exact test, and the Likelihood ratio were performed to test whether significant differences in age and length of time of manual WC use, gender and type of WC frame, respectively, existed among NP and no NP in the past month groups.

Independent t-tests were used to determine whether significant differences in age and length of time of manual WC use existed between LBP and no LBP in the 2-week period of data collection groups. Fisher's exact test and Likelihood ratio were used to determine whether significant differences exist in gender and type of WC frame between the same groups.

McNemar's test was used to compare self-reported prevalence of LBP before and during the data collection. This test was used because samples were related; each individual was questioned twice about presence of LBP, first in the pain questionnaire and a second time after 2 weeks.

Two simple logistic regression models were used to determine whether some of the aggravating factors are related to whether the subject reported LBP or NP. Only three independent variables were used in both regression models due to the amount of observations available. The independent variables used were *self-reported amount of time spent seated in WC, number of transfers per day, and amount of time working at a desk*. Self-reported amount of time spent seated in WC was categorized in 3 levels with 0 representing the lowest amount of time option available to select (3-5 hours per day). Number of transfers per day and amount of time

working at a desk were at the interval level. In the first logistic regression model, the LBP variable was categorized into 2 levels with 0 representing the absence of LBP and 1 the presence of LBP. Likewise, NP variable, in the second logistic regression model, was categorized into 2 levels with 0 representing the absence of NP and 1 the presence of NP. LBP, NP and time spent in WC categorical variables were referred to 0. Post-hoc power analyses of the predictors to LBP were conducted using Sample Power (SPSS, Inc; Chicago, IL). The sample size of 37, and an alpha level set at 0.05 (2-tailed) were used for the statistical analyses. The effect sizes used for this assessment were those corresponding to each predictor and were selected as the smallest effect that would be important to detect.

Mobility characteristics were analyzed using three-way Mixed MANOVA that included two between-subjects variables: type of WC frame, and presence of LBP in the past month; and one within-subjects variable: type of environment (home or national event). This test was used because 1) participants only had one type of WC frame and either had LBP or not in the past month, and 2) because mobility characteristics were measured for all participants in two environments (repeated measure design): home and national event. Dependent variables analyzed included average daily distance traveled, speed, and accumulative driving time.

Vibration levels were analyzed using the same approach as activity levels above: three-way Mixed MANOVA. This also included the same two between-subjects variables (type of WC frame and LBP presence) and the same within-subjects variable (environment). Dependent variables analyzed included VDV_v and a_v at the seat, and a_{rms} and VDV at the x-axis of the backrest.

The relationship between the presence of LBP in the past month and risk factors to LBP was examined using binary logistic regression. Only three independent variables were used to

predict LBP due to the number of observations available. The independent variables -average daily distance, reported amount of transfers per day, and a_v at the seat- were selected. LBP variable was categorized into 2 levels with 0 representing the absence of LBP and 1 the presence of LBP. In the logistic regression the LBP categorical variable was referred to 0 (no pain). Before regression analysis was performed, linearity, multicollinearity and residuals diagnosis were tested.

All analyses were performed using SPSS software version 19.0 (SPSS, Inc; Chicago, IL) and significance level of .05.

3.0 RESULTS

3.1 DYNAMIC STIFFNESS AND TRANSMISSIBILITY OF COMMERCIALLY AVAILABLE WHEELCHAIR CUSHIONS USING A LABORATORY TEST METHOD

3.1.1 Dynamic characterization of wheelchair cushions

Figure 11 shows the measured modulus of the dynamic stiffness, $\vec{S} = \sqrt{K^2 + (C\omega)^2}$, of the seven seating systems for all the preloads (300–800 N). All curves of the modulus of the dynamic stiffness over the range of loads (300–800 N) were fitted within the 95 pe rcent goodness of fit to estimate stiffness and damping values. The estimated stiffness and the damping parameters for the seven different seating systems are shown in Table 3. All the cushions show increases in the stiffness and damping characteristics with increments in the preload. The Comfort Mate Foam (Invacare; Elyria, Ohio) and the Zoombang Protective Gear with Foam (Invacare) had the largest stiffness and damping values compared with the other cushions. The Jay J2 Deep Contour (Sunrise Medical) cushion had the lowest stiffness coefficients over the preload range, whereas the Meridian Wave (Varilite; Seattle, Washington) cushion showed the lowest damping coefficients. The former pair of seating systems also had the highest rate of increase in both the stiffness and damping parameters with increasing preloads. The Jay J2 Deep Contour and the Meridian Wave cushions had the lowest rates of increase of

damping, and the Jay J2 Deep Contour and the ROHO LOW PROFILE (The ROHO Group; Belleville, Illinois) had the lowest rates of increase of stiffness.

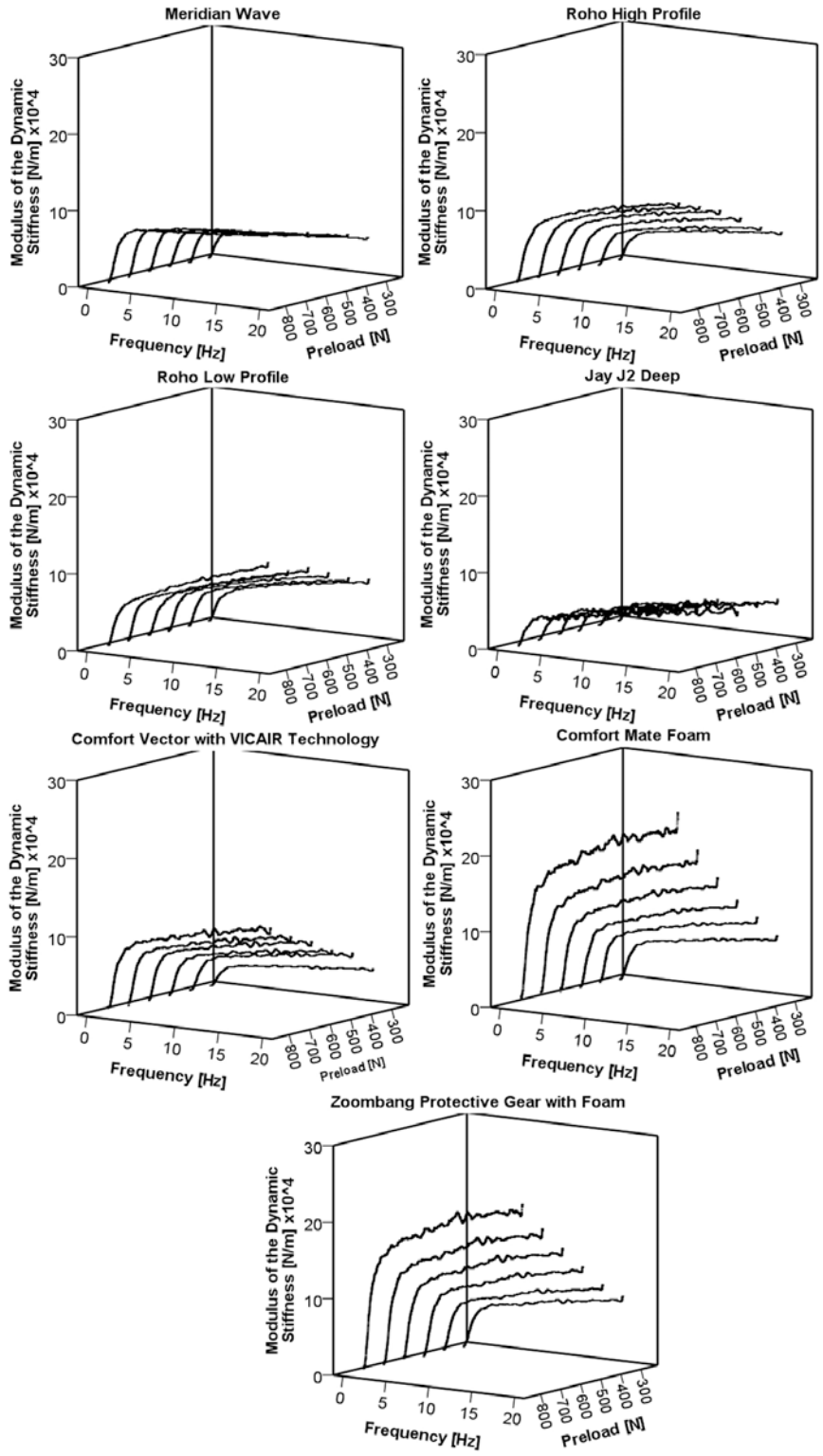

Figure 11. Material testing system-measured modulus of dynamic stiffness for all preload forces of seven seating systems.

Table 3. Stiffness and damping parameters, *K* (in N/m) and *C* (in Ns/m), of seven wheelchair cushions over range of preloads (300–800 N).

Load (N)	Vector with Vicair		Meridian Wave		ROHO HIGH PROFILE		Jay J2 Deep Contour		ROHO LOW PROFILE		Zoombang Protective Gear with Foam		Comfort Mate Foam	
	K	***C***	***K***	***C***	***K***	***C***	***K***	***C***	***K***	***C***	***K***	***C***	***K***	***C***
300	32,080	318	40,230	301	43,180	389	26,730	365	50,520	544	60,350	573	56,780	487
400	50,340	478	50,500	322	51,740	475	27,940	406	56,730	602	74,160	727	76,140	696
500	59,360	542	57,550	337	64,840	596	28,710	377	62,140	708	94,900	938	93,070	908
600	71,580	665	64,760	372	77,390	688	36,890	508	64,460	835	116,900	1,132	115,500	1,149
700	80,870	748	69,730	392	86,470	760	36,090	522	65,900	855	141,800	1,311	143,200	1,408
800	95,040	840	76,010	397	94,220	834	39,970	571	68,600	1,015	174,900	1,507	183,200	1,689
Rate of Increase/100N	11,960	101	6,966	21	10,627	91	26,730	365	3,435	93	22,791	189	24,449	240

3.1.2 Measurement and prediction of cushion transmissibility with nondisabled subjects and mathematical models

Fourteen subjects participated in the WRC test to measure and estimate vibration transmissibility of seven seating systems. The participants ranged in age from 18 to 57 years, with a mean ± SD of 30.4 ± 9.4 years; weight from 98.8 to 48 Kg (73.1 ±12.7); and height from 1.44 to 1.83 meters (1.7 ± 0.1). Figure 12 shows the averaged measured seat transmissibility obtained in the WRC test of the seven seating systems compared with the ODOF and TDOF models of the seat transmissibility calculated with Equations 11 and 12. As can be seen from these figures, all seating systems amplified vibrations in a frequency range harmful to humans when measured or predicted with mathematical models. However, neither ODOF nor TDOF models accurately predicted measured seat transmissibility during the WRC test. Both models overestimated seat transmissibility during WC propulsion at low frequency ranges (below 8 Hz) and underestimated transmissibilities at frequencies between 8 and 12 Hz compared to the measured WRC results.

An average PSD of the input vibration exposure of the subjects in the WRC test is shown in Figure 12 (bottom right corner). This figure shows that the spectral content is not the same throughout the frequency range from 0–20 Hz and differs from the flat spectrum the cushions are subjected to with the MTS. The PSD of the vibration exposure during the WRC test is higher at frequencies from 8–20 Hz than frequencies below 8 Hz.

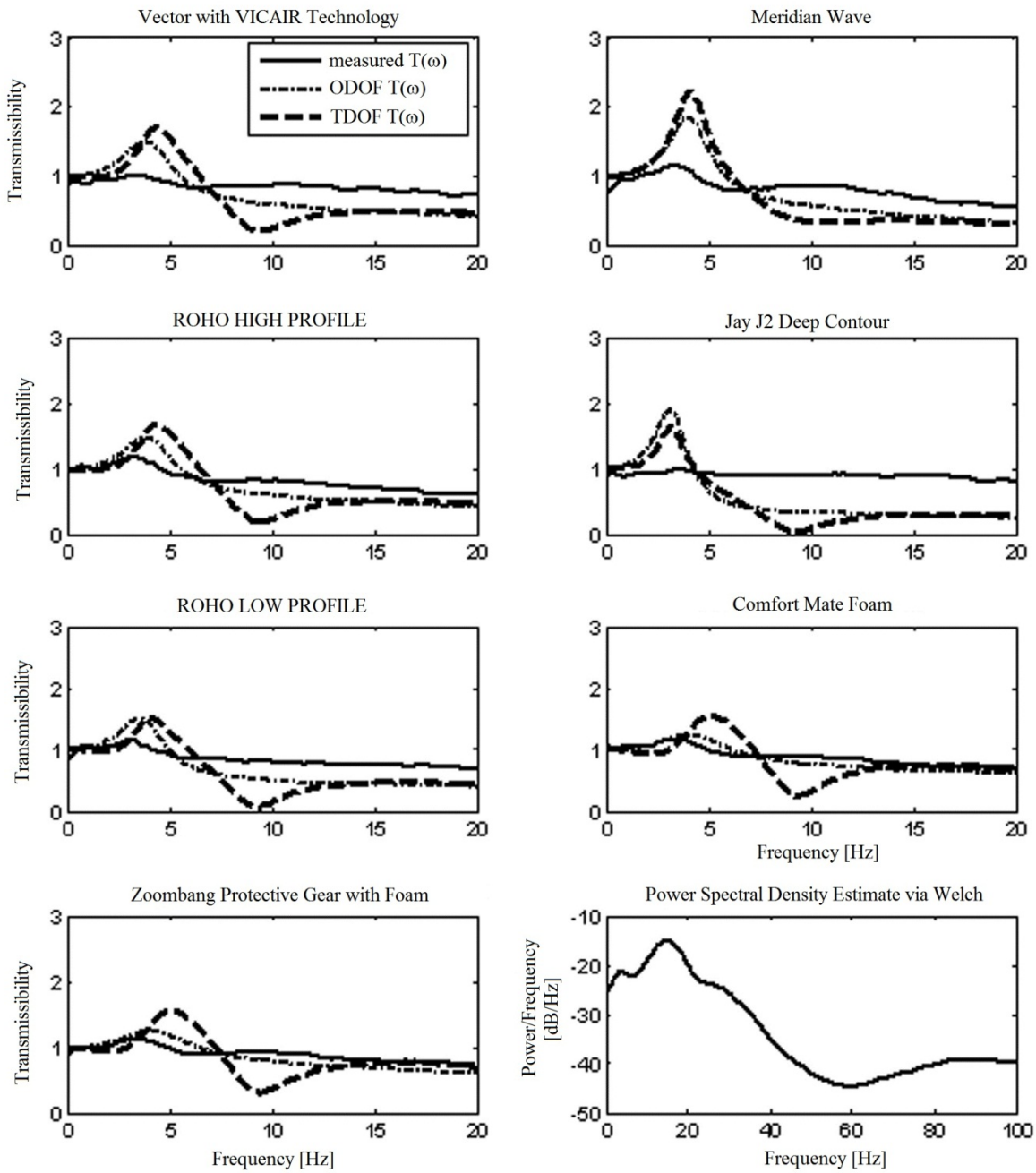

Figure 12. Measured and predicted seat transmissibility for seven seating systems. Bottom-right corner: power spectral density of averaged input acceleration of wheelchair road course.

A summary of the maximum measured and predicted transmissibilities and corresponding frequencies for the seven seating systems are listed in Table 4. Neither mathematical model accurately predicted the maximum measured seat transmissibility. The ODOF model predicted that the Jay J2 Deep Contour and Meridian Wave have the highest maximum transmissibility and that the Comfort Mate Foam and Zoombang Protective Gear with Foam have the lowest maximum transmissibility. The TDOF model predicted different results. With this model, the Meridian Wave was identified as the cushion with the highest maximum transmissibility, whereas the ROHO LOW PROFILE, the Comfort Mate Foam, and the Zoombang Protective Gear with Foam were identified as the cushions with the lowest maximum transmissibility. These results differ from measured results, which identified the Comfort Mate Foam and the ROHO HIGH PROFILE as the cushions with the highest maximum transmissibility and the Jay J2 Deep Contour and the Vector with Vicair Technology as the cushions with the lowest maximum transmissibility.

Maximum seat transmissibility and frequency estimations from the ODOF and TDOF models were significantly higher than those measured in the WRC test. After log-transformation of data were performed for normality, we performed repeated-measures ANOVA with Sidak adjustment for multiple comparisons. We found significant effects of the methods used on both the transmissibility and the corresponding frequency (F = 24.127, df = 2, $\rho < 0.001$; and F = 13.352, df = 1.084, $\rho = 0.009$, respectively). Sidak post hoc analysis results showed a significant effect on transmissibility between measured data and ODOF model predictions ($\rho = 0.02$, $\alpha = 0.05$) and between measured data and TDOF model predictions ($\rho = 0.001$, $\alpha = 0.05$) and a significant effect on frequency between measured data and TDOF model predictions ($\rho = 0.03$, $\alpha = 0.05$) and ODOF and TDOF model predictions ($\rho = 0.02$, $\alpha = 0.05$).

Table 4. Maximum transmissibility and corresponding frequency values obtained in WRC test and ODOF and TDOF models of seating systems.

Cushion	WRC		ODOF		TDOF	
	Frequency (Hz)	T	Frequency (Hz)	T	Frequency (Hz)	T
ROHO LOW PROFILE	3.12	1.16	3.56	1.51	4.10	1.52
Jay J2 Deep Contour	3.48	0.99	3.13	1.90	3.17	1.64
ROHO HIGH PROFILE	3.37	1.18	3.88	1.48	4.40	1.67
Meridian Wave	3.34	1.16	3.99	1.85	4.10	2.22
Vector with Vicair Technology	3.27	1.01	3.87	1.50	4.35	1.69
Zoombang Protective Gear with Foam	3.36	1.13	4.21	1.27	5.15	1.56
Comfort Mate Foam	3.16	1.18	4.13	1.26	5.13	1.55
Mean ± Standard Deviation	3.30 ± 0.13	1.12 ± 0.08	3.82 ± 0.37	1.54 ± 0.25	4.34 ± 0.68	1.69 ± 0.24

ODOF = one-degree-of-freedom, T = transmissibility, TDOF = two-degree-of-freedom, WRC = wheelchair road course.

3.1.3 Vibration dose values of wheelchair cushions

Figure 13 shows the VDV transmissibility measured in the WRC test. After log-transformations of data were performed for normality, we performed repeated-measures ANOVA with Sidak adjustment for multiple comparisons. We found significant effects of cushion on the VDV transmissibility (F = 7.219, df = 2.985, ρ = 0.001). The Meridian Wave, Vector with Vicair Technology, and ROHO HIGH PROFILE cushions had the lowest transmissibility, whereas the

Jay J2 Deep Contour, Comfort Mate Foam, and Zoombang Protective Gear with Foam had the highest transmissibility.

Figure 13 shows a box-plot of the VDV transmissibility and statistical groupings. Sidak post hoc analysis results revealed significant differences between (1) Jay J2 Deep Contour and Vector with Vicair technology ($\rho = 0.003$, $\alpha = 0.05$), (2) Meridian Wave and Jay J2 Deep Contour ($\rho < 0.001$, $\alpha = 0.05$), (3) Meridian Wave and Zoombang Protective Gear with Foam ($\rho = 0.014$, $\alpha = 0.05$), (4) Meridian Wave and Comfort Mate Foam ($\rho = 0.012$, $\alpha = 0.05$), and (5) ROHO HIGH PROFILE and Comfort Mate Foam ($\rho = 0.03$, $\alpha = 0.05$).

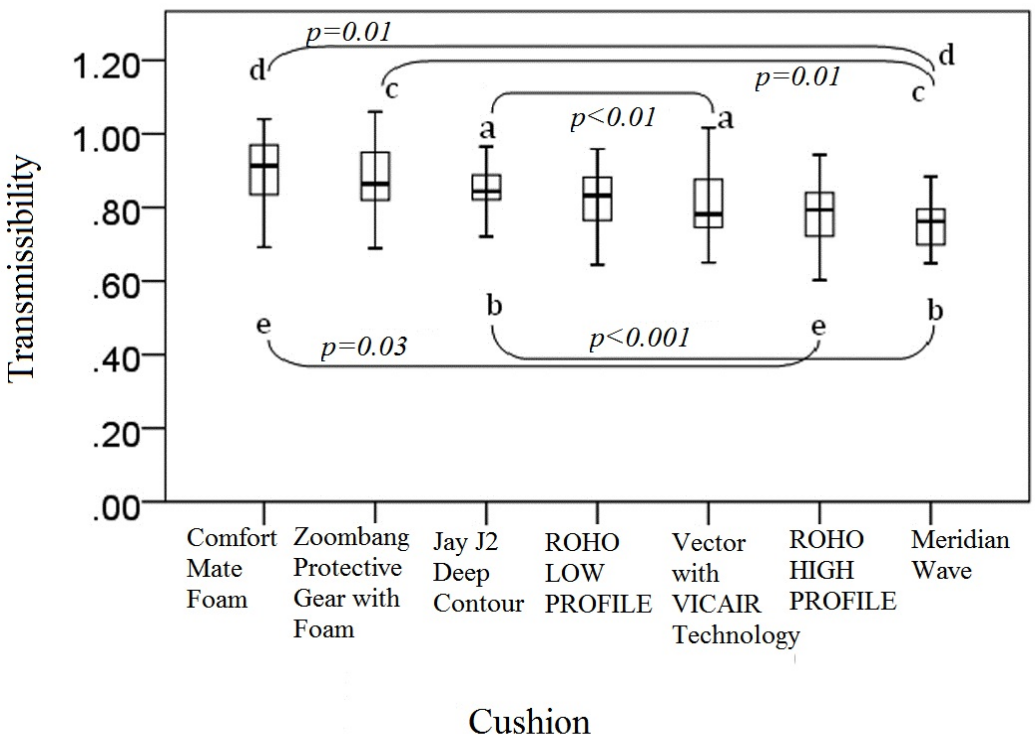

Figure 13. Distribution of vibration dose value transmissibilities measured in wheelchair road course for seven seating systems. a, b, c, d, and e identify cushions with significant differences.

3.2 HEALTH RISKS OF VIBRATION EXPOSURE TO WHEELCHAIR USERS IN THE COMMUNITY

3.2.1 Subjects

A total of forty-eight subjects consented to participate in the study. One subject did not meet the inclusion criteria. Two subjects did not return the VDL. Eight subjects did not finish the protocol because either they did not use their WC for the second week of the study or the seat sensor of the VDL did not turn off thereby collecting data even when the participant was not seated in their WC. Follow-up contact with these participants revealed that they were not using their WC the week after the national event for different reasons, such as a WC repair or a long trip that required them to be out of their WC, and which do not represent daily use.

The remaining thirty-seven individuals were included in the data analysis, of whom 5 were female and 32 were male. The participants ranged in age from 26 to 64 years, with a mean ± SD of 47.6 ± 11.6 years. The amount of time participants have used a WC ranged from 1 to 43 years, 15.0 ± 11.5 years. Of the 37 subjects, 25 (67.6%) used a WC because of a SCI. Of these 25 individuals, 20 had paraplegia and 5 had quadriplegia. The rest of the participants reported lower extremity amputation (n = 6), multiple sclerosis (n = 2), arthritis, post-polio and traumatic brain injury (n = 3). Nineteen percent (n = 7) of the participants had been diagnosed with curvature of the spine, 16.2% (n = 6) with vertebral fracture, 13.5% (n = 5) with arthritis of the spine, and 8.1% (n = 3) with pinched nerve in neck. Sixty-two percent (n = 23) indicated other secondary conditions. There were no demographic differences between individuals completing the study and those who did not complete the study.

3.2.2 Wheelchair characteristics

All individuals independently propelled their manual WC and indicated that they use it as their primary means of mobility. Thirteen subjects (35.1%) used a folding frame WC and twenty-four (64.9%) used a rigid frame WC. Of the folding frame WCs, 9 (69.2%) had no suspension, 3 (23.1%) had suspension in the casters and 1 (7.7%) had suspension in both the casters and frame. Of the rigid frame WC, 20 (83.3%) had no suspension, 2 (8.3%) had suspension in the casters, and 2 (8.3%) had suspension in both the casters and the frame. Rear-wheel suspension WCs included in this study were the following: Quickie Q7, TiLite TR, and Colours Shockblade.

3.2.3 Presence of pain

Table 5 summarizes the prevalence of LBP and NP among participants. Of those respondents who reported LBP at any time, 52.2% (n = 12) visited a doctor regarding the LBP, and 34.8% (n = 8) limited their daily activities due to the LBP. In relation of NP, 36.8% (n = 7) visited a doctor, and 31.6% (n = 6) limited their daily activities. 58% (n = 11) of the participants who reported NP at any time said they experienced NP while propelling their WC, while 31.6% (n = 6) of them had the pain before they started using a WC. Of the subjects who reported NP before WC use, 50% (n = 3) said the pain had worsened since WC use.

Table 5. Frequency of pain among participants (n = 37).

	Since 1 year after the onset of the condition that caused to use a wheelchair	within the past month	within the past 24 hours
Number of participants with Lower back pain, n (%)	20 (54.1%)	18 (48.6%)	12 (32.4%)
Number of participants with Neck/upper back pain, n (%)	14 (37.8%)	16 (43.2%)	11 (29.7%)

When analyzing groups according to the presence of LBP or NP in the past month, there was no significant difference in age, length of time of manual WC use, gender, or type of wheelchair (see Table 6). McNemar's chi-square statistic suggested that there was not a statistically significant difference in self-reported prevalence of LBP before (in the past month) and during the 2-week period of data collection $\rho > .05$, suggesting that the LBP level did not change during the study.

Table 6. Participants' demographics when divided into LBP and no LBP groups.

	No LBP group (n = 19)	LBP group (n = 18)	No NP group (n = 21)	NP group (n = 16)
Age, year (range)	47.7 ± 9.6 (29 - 64)	47.4 ± 13.7 (26 - 64)*	50.3 ± 10.7 (30 – 64)	43.0 ± 12.0 (26 – 64)**
Years of WC use (range)	16.8 ± 11.5 (2 - 43)	12.0 ± 10.6 (1 - 41)	16.0 ± 12.3 (2 - 43)	12.5 ± 9.5 (1 – 30)
Gender, female/male	4/15	1/17	3/18	2/14
WC type, folding/rigid/suspension	6/9/4	3/11/4	6/10/5	3/10/3

Subjects were divided according to presence or absence of self-reported LBP and NP in the past month. There were no significant differences in demographics between LBP and no LBP groups. *n = 17, ** n = 15, since one participant did not specified age.

The VAS, mean ± SD in cm (range), for participants reporting NP over the 24 hours previous to the study (n = 11) was as follows: 1) VAS, current pain: 4.90 ± 3.42 (0 - 9.5); 2) VAS, worst pain in past 24 hours: 5.63 ± 3.76 (0 - 10); and 3) VAS, best pain in past 24 hours: 2.82 ± 2.67 (0 - 8.5). The NDI for the same group was 22.26 ± 13.18 (2 - 51).

3.2.4 Aggravating factors related to LBP and NP and wheelchair use characteristics

Table 7 summarizes WC use characteristics of the individuals who participated in the study as well as other aggravating factors linked to LBP and NP. These risk factors include amount of time in seated position, working with hands above shoulder level, heavy object lifting, and weight bearing. As can be seen in Table 7, almost half of the participants (48.6%) reported that they spend more than 12 hours per day seated in their WC and 56.8% said they propel their WC for more than 2 hours per day. Data measured with the indicated similar results regarding amount of time participants spent seated in their WC since 46% of them spent between 6-12 hours and the rest, 54% more than 12 hours. In the other hand, MDL measurements showed that only 2.7% of participants spent less than 30 minutes propelling their WC whereas the rest spent between 30 minutes and 2 hours of their time.

In addition, subjects support weight on their upper extremities and lift objects heavier than 20 lbs. approximately 20 times per day.

Table 7. Summary of subjects' exposure to aggravating factors to LBP and NP

Aggravating factor to LBP and NP

Aggravating factor to LBP and NP				
Amount of time spend seated in a WC per day [frequency]	3-5 hours 16.2% (n = 6)	6-12 hours 35.1% (n = 13)	12-24 hours 48.6% (n = 18)	
Amount of time spend propelling their WC per day [frequency]	10-30 min 2.7% (n = 1)	30-60 min 16.2% (n = 6)	1-2 hours 24.3% (n = 9)	More than 2 hours 56.8% (n = 21)
Average amount of time spend on (in hours)				
Working at a desk [mean ± SD]	(1.85 ± 2.75)			
Working at a computer [mean ± SD]	(2.74 ± 2.54)			
Working with arms overhead [mean ± SD]	(0.36 ± 0.65)			
Working with hands [mean ± SD]	(4.53 ± 4.24)			
Driving [mean ± SD]	(1.81 ± 1.27)			
Reading [mean ± SD]	(1.96 ± 1.84)			
Average number of transfers per day [mean ± SD]	(9.51 ± 4.17)			
Average number of times per day lifting objects 20lbs or heavier [mean ± SD]	(10.03 ± 18.63)			

Results of the regression analysis were non-significant to predict LBP or NP based on aggravating factors (Table 7).

3.2.5 Mobility characteristics and vibration exposure levels

Average levels of vibration exposure at the seat and in the x-axis of the backrest, as well as mobility characteristics of participants during the 2-week period of data collection are shown in Table 8 in the "Combined" column. These data are based on the duration of exposure recorded by the VDL. Duration of vibration exposure was calculated based on the length of vibration data collected every time a person was seated in the WC. Exposure time was then the ratio of the amount of acceleration data samples to 60 (i.e. the sampling frequency). On average, participants

spent an average of 13.07 ± 3.85 hours per day seated on their WC during the 2-weeks of data collection.

In the second week of data collection, in the community environment, there were cases where no activity was recorded for entire days. It was assumed that participants used a backup WC those days. Although information about ownership of a backup WC was not recorded in this study, Tolerico et al (2007) found that 83% of the subjects in their study, who were also veterans participating in national events, owned a backup WC; and that 38% of them used their backup WC at least once a week. They included inactive days in their analysis, as did we; because it was assumed that these patterns of activities are representative of day to day life [43].

After checking for assumptions of multivariate normality and homogeneity of covariance matrices for MANOVA analysis, the Pillai's statistic indicated that no significant differences existed on mobility characteristics (distance, speed, and accumulative driving time), vibration level exposures, or duration of exposure based on the effect of LBP presence or type of WC frame. No significant interaction effects were found either. A summary of these data is shown in Table 8 and Table 9.

Table 8. Comparison of mobility characteristics and vibration exposure levels (mean ± SD) to participants according to presence of self-reported LBP in the past month.

	No LBP group (n = 19)	**LBP group (n = 18)**	**Combined (n = 37)**
Mobility characteristics			
Max distance of continued movement (m)	251.0 ± 151.1	200.4 ± 87.9	226.4 ± 125.4
Max period of continued movement (min)	3.8 ± 1.9	3.0 ± 0.9	3.4 ± 1.5
Distance (m)	2,931.2 ± 1041.1	2,324.2 ± 690.1	2,635.9 ± 928.1
Speed (m/s)	0.74 ± 0.15	0.71 ± 0.16	0.73 ± 0.16
Accumulated movement time (min)	64.2 ± 18.2	50.1 ± 15.3	57.3 ± 18.1
Seat vibration measurements			
a_v, **in m/s^2**	0.81 ± 0.13	0.85 ± 0.21	0.83 ± 0.17
VDV_v, **in m/s$^{1.75}$**	17.27 ± 3.39	17.26 ± 3.15	17.26 ± 3.23
x-axis backrest vibration measurements			
a_{rms}, **in m/s^2**	0.54 ± 0.11	0.57 ± 0.14	0.55 ± 0.13
VDV **in m/s$^{1.75}$**	12.44 ± 2.77	11.66 ± 1.85	12.06 ± 2.37
Duration of exposure (hours)	16.69 ± 3.88	12.41 ± 3.82	13.07 ± 3.85

Max = maximum. a_v = Point vibration total value. VDV_v = Point vibration dose total value. a_{rms} = Weighted r.m.s. acceleration. VDV = Vibration dose value.

Table 9. Comparison of mobility characteristics and vibration exposure levels (mean ± SD) to participants according to type of WC frame (folding, rigid, and suspension).

	Folding (n = 9)	**Rigid (n = 20)**	**Suspension (n = 8)**
Mobility characteristics			
Max distance of continued movement (m)	315.4 ± 197.2	196.3 ± 69.4	201.3 ± 99.2
Max period of continued movement (min)	4.7 ± 2.3	3.1 ± 0.8	2.8 ± 1.0
Distance (m)	2,863.0 ± 649.5	2,445.1 ± 810.3	2,857.5 ± 1392.8
Speed (m/s)	0.7 ± 0.2	0.7 ± 0.1	0.7 ± 0.2
Accumulated movement time (min)	64.7 ± 15.1	54.0 ± 18.1	57.3 ± 20.7
Seat vibration measurements			
a_v**, in m/s^2**	0.87 ± 0.14	0.82 ± 0.18	0.82 ± 0.20
VDV_v**, in m/s$^{1.75}$**	16.99 ± 2.60	17.27 ± 3.43	17.57 ± 3.70
x-axis backrest vibration measurements			
a_{rms}**, in m/s^2**	0.60 ± 0.13	0.53 ± 0.11	0.58 ± 0.16
VDV**, in m/s$^{1.75}$**	12.20 ± 1.59	11.97 ± 2.55	12.11 ± 2.87

Max = maximum. a_v= Point vibration total value. VDV_v = Point vibration dose total value. a_{rms} = Weighted r.m.s. acceleration. VDV = Vibration dose value

Significant differences were found for mobility characteristics (distance, speed, and accumulative driving time) and vibration levels based on the effect of the environment: $V = 0.48$ $F(3, 27) = 8.31$, $\rho < .001$; and $V = 0.704$, $F(4, 27) = 16.09$, $\rho < .001$; respectively, based on the Pillai's statistic. Subsequent univariate pairwise comparisons on dependent variables, with a Sidak correction, revealed that participants traveled greater distances (mean = 3324.32, SE = 241.33), and accumulated longer periods of movement (mean = 68.47, SE = 4.34) at national event settings than they did in their home environments (mean = 1883.73, SE = 172.72, for distance, and mean = 43.53, SE = 3.78, for accumulated continued movement time), t (34) = -4.75, $\rho < 0.001$, r = .63; and t (34) = -4.46, $\rho = 0.001$, $r = .61$, respectively. Similarly, in their home environment setting, a_v (median = 0.72) and VDV_v (median = 14.91), measured at the seat,

were significantly lower than at the national event environment (median = 0.85, for a_v and median = 16.79, for VDV_v), z = -4.346, $\rho < 0.001$, r = -.73, and z = -3.88, $\rho < 0.05$, r = -.65, respectively. Likewise, a_{rms} (mean = 0.52, SE = 0.02) and VDV (mean = 11.07, SE = 0.51), measured at the x-axis of the backrest, were significantly lower in their home environment than at the national event environment (mean = 0.58, SE = 0.02, for a_{rms}, and mean = 12.77, SE = 0.38, for VDV), t (35) = -5.40, $\rho < 0.001$, r = .68, and t (35) = -4.88, $\rho < 0.001$, r = .64, respectively. When duration of exposures were compared based on environment settings, no significant differences were found $T = 0$, $\rho > 0.05$, r = -.03.

3.2.6 Evaluation of vibration exposure levels based on their risk to health

ISO 2631-1 has established a health guidance caution zone to evaluate the effects of vibration on health. According to these guidelines, for a 1 3-hour duration of vibration exposure (i.e. the average duration of vibration exposure to participants in this study) the maximum weighted acceleration exposures (a_{rms} or a_v) for a potential effect on health (lower bound of the zone) is 0.34 m/s^2, and to be likely (upper bound of the zone) is 0.68 m/s^2. The estimated VDV corresponding to the lower and upper bounds are 8.5 $m/s^{1.75}$ and 17 $m/s^{1.75}$, respectively. An analysis of vibration exposure at the home and national event environment, revealed that ***all the participants were exposed to vibration at the seat surface (VDV_v and av) that was within or above the health caution zone specified in ISO 2631-1*** (see Table 10). However, vibration exposure at national event environment tended to be higher. **Ninety-seven percent of a_v measurements were above the health caution zone**. Participants' exposure to vibration measured at the x-axis of the backrest was lower and tended to be localized within the health caution zone in comparison to exposure measured at the seat. Table 10 shows how vibration

exposures at the seat and at the x-axis of the backrest were distributed in the health caution zone specified in ISO 2631-1.

Table 10. Frequency of vibration exposure levels for participants on the Health Caution Zone.

	Health Caution Zone		
	Below (home / national event) (%)	**Within (home / national event) (%)**	**Above (home / national event) (%)**
Seat vibration measurements			
a_v**, in m/s²**	0.0 / 0.0	30.6 / 2.8	69.4 / 97.2
VDV_v**, in m/s$^{1.75}$**	0.0 / 0.0	66.7 / 54.0	33.3 / 46.0
x-axis backrest vibration measurements (home / national event)			
a_{rms}**, in m/s²**	2.8 / 0.0	80.6 / 78.4	16.7 / 21.6
VDV**, in m/s$^{1.75}$**	16.7 / 0.0	77.8 / 94.6	5.6 / 5.4

a_v = Point vibration total value. VDV_v= Point vibration dose total value. a_{rms} = Weighted r.m.s. acceleration.VDV = Vibration dose value. Vibration exposure levels are based on acceleration measurements at the home environment (second week of data collection).

Figure 14 shows vibration exposure levels, in a_v and VDV_v, recorded at the seat surface of the WC during the two weeks of data collection. It can be seen in this plot that the all the participants were exposure to vibration levels within and above the health caution zone established by the ISO 2631-1.

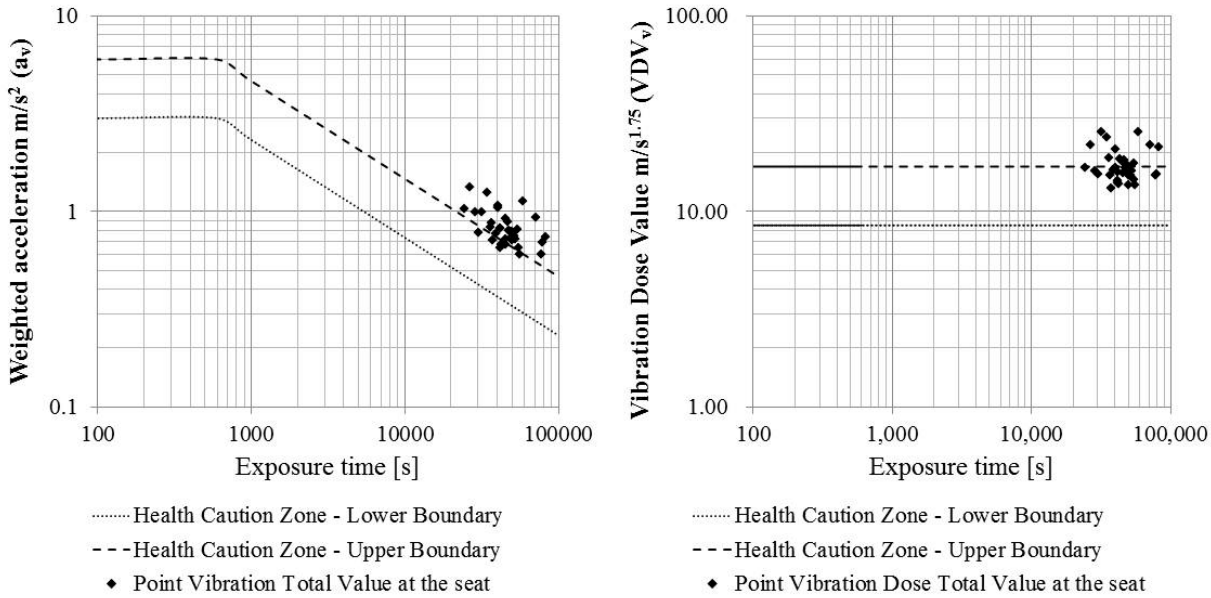

Figure 14. Average daily point vibration total value (a_v) and point vibration dose total value (VDV_v) at the seat of two weeks of data collection for all the participants compared to the acceptable threshold-limits for WBV exposure established by ISO 2631-1.

3.2.7 Prediction of LBP based on risk factors

The binary logistic regression coefficients for LBP based on daily distance traveled, self-reported amount of transfers, and vibration level at the seat are reported in Table 11. The results of the regression analysis with LBP as the outcome variable showed daily distance as the only significant predictor ($\rho < .05$) although, with a 1-meter difference in distance, the odds of having LBP or not are essentially even. a_v and the amount of transfers per day were not found to be useful predictors of LBP ($\rho > .05$). The binary regression model with these variables included explained 30.2% of the variation in the probability of having LBP based on the Nagelkerke R^2 statistic. When diagnosing the residuals of the model, 3 influential cases were found by

examining the standardized Beta values. However, no significant reasons for excluding such cases were found in the data. Furthermore, a logistic regression analysis was performed excluding influential cases with only a 3% improvement in the predicted accuracy of the model, which did not support excluding the influential cases. In general, the logistic regression model accuracy rate was 70.3% (compared to the 51.4% of accuracy rate of the model including only the constant).

Overall, the logistic regression model is useful to predict LBP presence only for the participants of this study. A cross validation analysis showed discrepancies with the full data set analysis.

Table 11. Binary logistic regression analysis of aggravating factors with presence of LBP.

Predictor	**B (SE)**	**Significance**	**OR**	**95% CI for OR**	
				Lower	Upper
a_v, in m/s^2	3.30 (2.50)	.19	27.07	0.20	3671.68
Distance (m)	-0.002* (.001)	.03	1.00	1.00	1.00
Average number of transfers per day	0.24 (0.13)	.07	1.27	0.98	1.65
Constant	-0.95 (2.25)	.68	0.39		

a_v= Point vibration total value; CI = confidence interval; OR = odds ratio; SE = standard error. Note: R^2 = .23 (Cox &Snell), .30 (Nagelkerke). Model χ^2 (3) = 9.50, p <.05. * p < .05.

4.0 DISCUSSION

4.1 DYNAMIC STIFFNESS AND TRANSMISSIBILITY OF COMMERCIALLY AVAILABLE WHEELCHAIR CUSHIONS USING A LABORATORY TEST METHOD

Dynamic properties of seven different WC cushions were obtained with two laboratory test methods. Different stiffness and damping parameters (Table 3) among the WC cushions were found and are attributable to their construction materials. The Comfort Mate Foam and the Zoombang Protective Gear with Foam cushions had the highest values of dynamic stiffness in contrast with the other seating systems, which have more complex constructions of air-foam, air-gel, or air. For instance, the air suspension properties of the ROHO HIGH PROFILE and ROHO LOW PROFILE cushions resulted in low stiffness values in contrast with the Comfort Mate Foam cushion. In the same way, the Meridian Wave cushion showed similar stiffness but lower damping to the air group cushions. This may be due to the low stiffness of its foam base component (lower than the foam cushion), which contrasted with the higher stiffness of the compartments of the ROHO cushions during compression. On the other hand, the Jay J2 Deep Contour cushion showed low values of stiffness and damping. The viscous gel contained in this cushion behaves like a solid during high-impact vibrations and is not good for suppressing vibrations [35]. It was observed that the gel slowly moved out of the area under the indenter, which may have decreased its dynamic stiffness. As preload increased, this effect was more

noticeable and may have been a consequence of using the SIT-BAR as the indenter, although it is the industry standard for measuring vibration exposure under the seated individual. This effect occurs with WC users as well.

Figure 12 and Table 4 show that the ODOF and TDOF models failed to predict the seat transmissibility obtained in the WRC test. A possible reason for this could be that the ODOF and TDOF mathematical models were developed to fit seat transmissibility data obtained by directly vibrating human subjects with a vibrator table excited by random vibration with a flat acceleration PSD over the range 1–30 Hz [41], whereas the frequency content of the vibration produced by the WRC was variable. In addition, the WRC may not excite all modes of the cushion, and those that are excited may not be excited with equal power. A PSD of the input vibration exposure of the subjects in the WRC test is shown in Figure 12 (bottom right corner). This figure shows that the spectral content is neither flat nor the same throughout the range from 0–20 Hz. In addition to the PSD of the input vibration, other factors could affect seat transmissibility measures. For instance, subjects' body position (i.e., leg and back position, contact with backrest, contact with footrest, and arm support) has been show to affect seat transmissibility measurements [12]. Asking individuals to propel the WC may have had an important effect on their body position and load distribution on the seat, thereby affecting seat transmissibility measurements. For example, leaning forward off of the backrest has been shown to decrease the resonance frequency [12]. With vibrations at a frequency just above resonance, subjects' leaning forward off of the backrest also caused significant decreases in seat transmissibility [12]. In addition, during the WRC test, the individuals had support for their feet, had to lean forward to ride the WC over obstacles (especially for the ramp and the sine-wave bumps), and had constant hand contact with the hand rim of the rear wheels. Furthermore,

changes in trunk position during WC propulsion may also have a significant effect on the center of mass of the body (i.e., the total load applied on the seat). Because seat transmissibility is a function of both the apparent mass and the dynamic stiffness of the cushion, changes in the load applied on the seat cushion (as either a p reload or the subject's apparent mass) may have significantly affected seat transmissibility measurements. An example of the effect of changes in preload on the dynamic stiffness of the cushion is clearly illustrated in Figure 11. Some studies have also found very little effect on vertical seat-to-head transmissibility measurements when vibrating the same subject without change of posture, whereas varying the body posture to a more erect position significantly increased the transmissibility at all frequencies above 3 Hz. Table 4 shows significant decreases in maximum seat transmissibility and corresponding frequency during the WRC, which were probably caused by changes in body position. Furthermore, increasing user contact with the seat surface while the user "pulls-up" against the hand rim has been shown to decrease accelerations measured at the head [44].

Although pull-ups were not allowed in this study, subjects' use of the hand rim while traversing obstacles in the WRC may have had an effect on the amount of user contact with the seat surface, thereby changing the acceleration experienced at the user-seat interface. Additionally, individuals in the WRC propelled the WC at self-selected speeds that were not constant among all the participants. Variations in speed cause variations in vibration magnitude and frequency, which can affect the cushion-human system, thereby affecting seat transmissibility and resonance frequency measurements [12]. Griffin and Erdreich point out that increases in vibration magnitude produce decreases in seat transmissibility and resonance frequency [12]. The WRC includes obstacles that produce high-magnitude vibration impacts (e.g., the double curb descent shown in Figure 7), which may also contribute to reductions in

maximum transmissibility and corresponding frequency. To measure seat vibration, we collected acceleration measurements at the human-cushion interface using the SIT-BAR to attach the accelerometers instead of the interface device developed by the Society of Automotive Engineers (SAE), the SAE pad [12], which was used for the development of the mathematical models. This may have had an effect on the cushion's surface deformation (especially for the gel cushion as it shifts the body down), the participant's position on the cushion, and the body contact area, thereby affecting transmissibility measurements. Unlike the SIT-BAR, the SAE pad is a thin "semi-rigid" ring that adjusts to the surface of the seat when loaded [12]. We preferred the SIT-BAR because its contour is similar to the human buttocks [12] and its rigid material and thickness allow it to be easily attached to the MTS. Finally, although the WC was not fit to participants as it would be to a client, the use of a single WC helped eliminate the effects of different WC types on the vibration transmissibility.

Although a drawback of this study was that the transmissibility measured with the MTS and on the WRC did not correspond well, we demonstrated the challenge of trying to accurately model WC propulsion with simple mathematical models. Future work will explore how to improve these models to better represent vibration transmission during WC propulsion and avoid use of disabled subjects for testing dynamic stiffness and measuring seat transmissibility. As a first step, we plan to develop a dynamic road simulator so subjects can be exposed to prescribed vibrations while acceleration measurements are collected above the cushion and at the head.

Significant differences were found between the VDV transmissibility measured on the WRC and the transmissibility from the models. These results may have important implications for WC cushion recommendations and selections for active WC users who are exposed to WBV on a daily basis. Figure 13 shows different transmissibility values among cushions estimated via

the VDV method. Cushions with an air component had lower VDV than foam or gel cushions. For instance, the ROHO HIGH PROFILE and the Meridian Wave cushions had the lowest transmissibility values, while the Comfort Mate Foam and Jay J2 Deep Contour had the highest. No significant differences were found between the Comfort Mate Foam and the Zoombang Protective Gear with Foam ($\rho = 0.439$, $\alpha = 0.05$), suggesting that the Zoombang Protective Gear with Foam was not successful in reducing vibration exposure. The WRC results are consistent with prior results reported by DiGiovine et al. [35] and suggest that air-based cushions, or those with an air component, may be better at reducing vibration transmissibility than foam- or gel-based cushions. Future work should focus on the design of WC cushions, and suspension systems, with better vibration-dampening characteristics, or that shift frequency content of vibration out of the range most detrimental for the human.

It is important to note that the WRC test and VDV transmissibility methods produced different results. While VDV is an evaluation method whose output is a single value over the entire range of frequencies that does not provide information at specific frequencies, the WRC test is a means to simulate and measure vibrations over a range of frequencies and does not provide an assessment score. Moreover, the VDV method applies a weighted frequency filter to the acceleration data for VDV transmissibility estimation. The filter weights those accelerations at frequencies more dangerous for health.

Future research to characterize WC cushions should investigate the use of other indenters. For example, the rigid cushion loading indenter described by ISO 16840-2 may be a better representation of human anatomy than the SIT-BAR and may provide more accurate measurements of seat dynamic stiffness. Dynamic stiffness values are input parameters for mathematical models of human-cushion systems. In addition to measurements of the dynamic

response of WC cushions to vibration (such as those provided in this study), future studies should include measurements of the cushion's ability to absorb impacts related to curb descents and other obstacles as defined by ISO 16840-2 [45]. For instance, Sprigle et al. found different results than those found in this study regarding the ability of foam cushions to absorb vibrations and impact [46]. Yet Sprigle et al. used initial impact accelerations and rebounds as metrics of vibration and impact absorption rather than seat transmissibility. Their results showed that a 3 in.-flat foam cushion decreased initial impact acceleration during impact loading more than the other cushions in the test, suggesting greater dampening characteristics. Still the authors point out that more research is needed to relate results to clinical outcomes.

A final limitation of this study was the assumption that WC users can be represented by nondisabled subjects. It has been shown that disabled subjects have different biomechanics and body characteristics than nondisabled subjects, thereby affecting measurements of seat transmissibility [24, 25, 47, 48]. Future investigations should consider these factors when examining dynamic seat response to vibrations.

Future research should attempt to collect vibration exposure in the community for long durations. Combined with accurate measures of cushion transmissibility and comfort levels, this exposure data would be helpful in determining the health risks posed to the WC user and which cushions would reduce that risk. As a final recommendation, future work should also focus on objective measurements of ride comfort and back pain levels to explore their relationship with vibration exposure.

4.2 HEALTH RISKS OF VIBRATION EXPOSURE TO WHEELCHAIR USERS IN THE COMMUNITY

The severity of LBP that a person can experience can have many implications for their daily life, affecting participation not only in society but also in basic activities of daily life [49]. Researchers in the field of rehabilitation and occupational injuries have tried to elucidate risk factors for LBP that are present on common real-world situations, e.g. motor vehicle driving. Only few have focused their attention on other groups that may be exposed to similar risks and for which implications on daily life can be even more severe and disabling, such as WC users. This study showed that LBP can affect not only a person's participation in daily life activities, but also may require additional resources and energy to alleviate its consequences. For instance, 35% of the participants with LBP reported limiting their participation in daily activities, and more than half had to visit a doctor because of their pain.

LBP prevalence in this study also supports the fact that LBP prevalence is higher in WC users than in the general working population [3, 4]. LBP prevalence among participants was nearly 48.6%. Other studies investigating LBP prevalence in WC users have reported higher rates of LBP: between 61 and 63% [49, 50]. This difference may be explained by the difference between the studies' participants. Subjects in this study were WC users participating in the VA WC events who were in an appropriate physical and mental condition to travel and participate, thereby under representing those who stayed home because of severe pain [21].

In general, the pain questionnaire responses revealed that WC users are highly exposed to contributing factors to LBP such as prolonged sitting and lifting heavy objects. The results of this study shows both situations: 1) 84% of the participants in this study reported spending more than 6 hours seated in their WC, and almost 50% spend more than 12 hours in it, which was

confirmed by our VDL results. This result is not surprising since WC riders rely on their WC to perform most part of their activities throughout the day. Periods of rest between seated times may be recommended since excessive time seated cannot be avoided. 2) Participants lift heavy objects an average of 10 times per day, and this number increases to almost 20 when lifting themselves every time they perform a transfer out or into their WC. Studies have shown that back muscle response latency and magnitude to a weight shift stimulus increases after being exposed to WBV [1]. This overcompensated reaction of the back muscles can result in muscle injury and LBP. WC users may experience injury of back muscles and LBP due to this overcompensated reaction when transferring or lifting heavy objects right after riding in their WC. Researchers suggest 5 minutes of light motion of the spine to reverse WBV effects before performing any weight lifting [1]. The finding that aggravating factors are equally present between participants with pain and without pain suggest two things: 1) that some may not be able to sense the pain, and 2) that these activities may need to be performed regardless of pain and that WC users are adopting different movement patterns to avoid pain while performing their activities [51]. For instance, a study carried out by Kim et al (2010) found that individuals with LBP have reduced torso movements and rotation and increased shoulder and arms movements while performing tasks that require manual loading and reach. The design of accessible workplaces and home environments should accommodate mobility limitations and pain-avoidance strategies adopted by WC users to reduce and prevent LBP.

By attaching custom VDLs and MDLs, we were able to objectively measure some of the risk factors associated with LBP of WC users in real-world environments for extended periods of time and without interfering with the person's activities. Measuring mobility characteristics and vibration levels at the WC frame during day-to-day living instead of during laboratory trials

gives the opportunity to evaluate real conditions to which WC users are exposed to, such as vibrations induced when traveling over surfaces in the home and community. It also provides the opportunity to assess current strategies being adopted to prevent LBP such as WC with suspension systems added. The preliminary scanning of vibration level exposures revealed crest factors greater than 9, which indicates that the measured vibrations contain high-peak accelerations. These data support previous findings of a short field trial carried out to investigate the loads applied on manual wheelchairs by road characteristics [26, 27]. These studies suggest that WC and riders are exposed to infrequent but high-magnitude vertical loads. Because of the presence of this acceleration peaks, VDV_v and VDV measures at the seat surface (z and x axis) and at the backrest (x axis) were included. These results also have implications on suspension design, which should be able to dampen these large accelerations.

Our results indicate that ***100% of the subjects were exposed to vibration loads at the seat surface that were either within or above the health-caution zone*** established by the ISO2631-1 standards. This result demonstrates how critical the need is for developing and implementing vibration-dampening strategies to prevent spine injuries among WC users. Nearly 31% of the participants were exposed to vibration levels (a_v) at the seat that were within the health caution zone (above 0.34 m/s^2 and below 0.68 m/s^2, for this specific exposure time) whereas the rest were exposed to vibration levels that were even higher than the health caution zone upper boundary (above 0.68 m/s^2). Regarding VDV_v, 67% of the time subjects were within the health caution zone (above 8.5 $m/s^{1.75}$ and below 17 $m/s^{1.75}$) and the remainder of the time the subjects were exposed to levels above the health caution zone (i.e. vibration doses greater than 17 $m/s^{1.75}$). ***These results show that WC users are at high risk of spine injuries because of the WBV levels they are exposed to, and that there is need of an intervention with this respect.***

Although most of the investigations performed in the past have suggested that WC users are exposed to WBV that contribute to LBP [21], none have actually quantified vibration levels in real-world environment for significant amounts of time. VanSickle et al (2001) [27] indicated that vibration during WC propulsion exceed fatigue-decrease proficiency boundary established in ISO 2631-1 at the seat of the WC during simulated course roads and a short field test. WBV exposures that exceed ISO 2631-1 standards have been positively correlated with LBP, herniated disc, degeneration of the spine, and other musculoskeletal disorders in motor vehicle drivers more so for prolonged periods of exposure [13, 14]. Vibration levels at the seat found in this study (0.82 ± 0.20 m/s^2, all subjects during 2-week of data collection) are comparable to those induced by an interlocking concrete surface with 8-mm bevels (0.80 m/s^2) and higher that those induced by standard poured concrete (0.47 m/s^2) that were measured in another study [31].

Vibration levels at the anterior-posterior axis (x axis) of the backrest also exceed ISO 2631-1 standards. The number of participants who exceeded the lower safety vibration threshold established by ISO was similar when measured with a_{rms} and with VDV. 80% and 78% of the participants exceeded this boundary when vibration was measured with a_{rms} and VDV respectively. a_v measured differed from VDV_v, both at the seat, since the second is a measurement more sensitive to high-acceleration peaks than a_v. Because vibration values measured at the seat combined x and z directions of acceleration, it is not surprising that for the seat measurements a_v and VDV_v differed whereas they were similar for the backrest which only included the anterior-posterior axis of vibration. It has been suggested that acceleration measured at the anterior-posterior axis is mostly composed by voluntary motion of the user during the propulsion activity, which is repetitive and continuous along the day, whereas the vertical

acceleration component of the vibration measured at the seat surface is better explained by a few high-peak acceleration events [27].

Vibration levels were found to be significantly higher at the national event setting than in the home environments. This finding may be explained by the fact that participants were more active at the national event settings. Results showed that participants traveled significantly farther, faster, and were active for more hours per day in these environments. Other studies have found similar results [23, 43]. Although not reported in this study, significant differences were found in vibration levels when comparing the types of national competition settings. Events that included more outdoor competition activities had higher vibration levels induced. Studies have shown that different paver surfaces induce significantly different levels of vibration during manual WC propulsion [31]. The fact that no significant differences were found on vibration levels among LBP groups may be explained by the observation that all the participants propelled their WC on similar surfaces (regardless of pain) at similar speeds during the first week at the national event.

An investigation of the vibration exposure based upon different types of WC frame revealed that suspension systems added to WC do not significantly reduce the amount of vibration measured at the frame. These results are similar to previous studies carried out on suspension WC that showed that adding suspension to manual and power WC does not necessarily reduce the amount of WBV transmitted to the user [30, 37]. However, the fact that suspension WC did not produce a significant reduction in vibration measured in this study has to be taken with caution, since the number of participants with suspension WC recruited in this study was small (only 3 participants had suspension on the frame, and the other 5 included in this category had suspension in the casters). Although not significantly, vibration measured in rigid

and suspension WC was lower than those measured on a folding frame. Other studies' results have suggested that some WC with suspension have similar vibration-dampening performance to rigid frames WC without suspension [30, 37]. The reduction of vibration observed here may be produced by the caster suspension. Caster suspensions have shown some reduction on vibration levels in other studies [30]. Wheelchair suspensions are not the only way to reduce vibration level exposure to WC users. WC cushions have also been identified as a means to decrease vibration exposure. In this study vibration levels were measured below the seat cushion without correcting for the transmissibility of the cushion material. However, other studies investigating WC cushions' dampening characteristics have shown that cushions are not effective in reducing vibration transmitted to the riders and in some cases they amplify them [1, 29, 34, 52, 53], suggesting that our findings may be a lower-bound on the actual exposure to the body.

Individuals in this study remained in their WC for an average of 13.07 ± 3.85 hours per day, which is similar to the self-reported amount of time spent seated in the WC collected from the questionnaire. Long periods of exposure time is one of most important contributing factors for risk of spine injury, and WC users are seated in their WC for even longer periods of time than other occupational groups at risk, for whom the literature reports an average of 8 hour exposure. Because of the accumulative effect of vibration, the risks associated with vibration for this amount of time is higher and therefore the vibration level threshold is lower than for 8 hour exposures to vibration. It is important to mention that caution should be taken when considering time duration as exposure duration, since it represents seated and not propelling time. Therefore, this time may overestimate real exposure duration when WC users are propelling their WC. Seated time includes times in which the WC users are actually not moving and may underestimate vibration levels.

One limitation of this study was the design of the seat sensors used to detect occupancy in a manual WC. During the national events there were a large variety of WC frames and cushions. These factors affected the performance of the seat sensors because these were designed for evenly distributed pressures along the surface. Future generations of VDL will require a different occupancy sensor design able to accommodate a wide variety of seat and cushion characteristics. A sensor that activates the data logging only when the WC user is actually moving would be desirable, for example the MDL.

WC riders are exposed to a combination of risk factors for musculoskeletal disorders and pain is a common symptom present in many of these conditions [54]. It is very difficult to distinguish and measure the contribution of each risk factor associated with LBP [3, 14]. In addition, 67.6 % of the participants in this study had a SCI which may have affected their ability to feel pain below the level of injury, and therefore may have not reported the presence of LBP. Other psychological and socioeconomic factors also add to this complexity - for example, level of education and employment status [49]. In this study an effort was made to identify the possible predictive ability of vibration level and other aggravating factors to the probability of having LBP. However, based on the information of aggravating factors recorded, the number of observations collected, and the multifactorial contributors to LBP, it was not possible to find a significant model to predict LBP. The sample size had an important implication in the regression models to LBP. A post-hoc power analysis was conducted for the most and least significant predictors to LBP (amount of time spent seated in a WC per day and number of transfers, respectively). The corresponding effect size of 0.302, for the categorical variable; the mean difference of -1.06 and the common within-group standard deviation of 4.20 (i.e. $r = 0.25$), for the continuous variable, were selected. The post-hoc analyses revealed the statistical power for

these tests were 35.6% and 11.6%, respectively, to yield a statistically significant result. Thus, there was less than adequate statistical power (i.e. power* .80) at these effect size levels. Larger sample sizes are suggested to yield statistically significant results. A sample size of 106, and 248 on each group, respectively would be needed to obtain statistical power at .80 level.

Vibration exposure is still a potential contributor to LBP, as it has been documented by many studies [1, 13, 15, 16]; and (as shown in Figure 14 of the results section) it is present at unhealthy levels among people using WC. Because vibration exposure levels were similar among participants with and without pain, it is possible that other factors which were not included in this study also influence LBP. Factors such as body posture, seating ergonomics, and the trunk biodynamic during WC propulsion should also be explored.

5.0 CONCLUSIONS

Selection of a WC cushion is critical to the health and safety of the WC user, especially if he or she has lost sensation. The important variables which currently drive the selection of a cushion are the pressure-relieving properties, weight, thermal properties, and ability to be cleaned. In this study, we demonstrated that transmissibility is another important characteristic to consider, as most cushions amplify vibrations (Table 4; T > 1.0). Air-based cushions outperformed the gel- and foam-based cushions and should be considered when selecting a cushion to help reduce vibration exposure or as a precaution against spinal pain.

Our transmissibility measurements calculated using the MTS and on the WRC did not correlate well, indicating that future work needs to focus on developing mathematical models of the body that better predict the dynamic response that occurs when propelling a WC. The development of such mathematical models will have important repercussions on WC cushion design and the prevention of health consequences when subjects test cushion dynamic stiffness.

WC users are exposed to vibration levels that exceed the ISO 2631-1 health caution zone. This level of vibration has been shown to have an effect on the spine, increasing the risk of deformities, LBP, and other types of musculoskeletal disorders. The use of suspension systems did not show a reduction of vibration and high-peak accelerations transmitted to WC users. Future suspension systems and/or cushions should be designed with vibration-dampening capabilities without affecting propulsion [26]. Accessible environments increased activity levels

of WC riders, but also vibration exposure, especially in outdoor activities. Other aggravating factors such as prolonged sitting and weight lifting were present among this population. More research is needed to collect information about how WC seating ergonomics and configuration, in combination with WBV, affect LBP.

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