THESIS

EFFECT OF REAR WHEEL SUSPENSION ON TILT-IN-SPACE WHEELCHAIR SHOCK AND VIBRATION ATTENUATION

Submitted by

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ABSTRACT

EFFECT OF REAR WHEEL SUSPENSION ON TILT-IN-SPACE WHEELCHAIR SHOCK AND VIBRATION ATTENUATION

Suspension systems are designed to reduce shock and vibration exposure. Prior to the QuadshoX LLC suspension kit (Fort Collins, CO), manual tilt-in-space wheelchairs did not have rear wheel suspension available for use. Furthermore, it was anticipated that rear wheel diameter would have an independent effect on shock and vibration transmitted to the wheelchair. The aim of this study was to investigate the shock and vibration reducing capabilities of the newly available aftermarket rear wheel suspension system and wheel diameter for manual tilt-in-space wheelchairs.

Ten healthy non-wheelchair users volunteered for the study (7 men, 3 women: age 22.1 ± 3.36 yrs, height 1.75 ± 0.067 m, weight 73.9 ± 8.87 kg (mean±SD)). Subjects were pushed by the same trained investigator over four different obstacles while using a Quickie IRIS ® Tilt-in-Space manual wheelchair (Sunrise Medical, Phoenix, AZ) with two different diameter solid wheels, (0.381 m and 0.508 m), Primo Cheng Shin Tires (Cheng Shin Rubber, Yuanlin, Taiwan). Surfaces included a/an 1) exterior door threshold, 2) truncated domes, 3) 2 cm descent, and 4) 2 cm ascent. The subjects traversed the obstacles with the wheelchair as manufactured, and followed ~ 2 weeks later with the QuadshoX suspension kit installed. A tri-axial accelerometer, (Model339A31, PCB Piezotronics, Depew, NY), was mounted to the rear of the wheelchair seat pan with signals sampled at 2000 Hz. Peak resultant accelerations were analyzed from surface 1, 3-4, root mean square (RMS) resultant accelerations were analyzed

from surface 2, and vibration dose value (VDV) and total power were analyzed from all surfaces 1-4. Unweighted and ISO 2631-1 frequency weighted (FW) accelerations were analyzed.

The use of suspension decreased the un-weighted peak acceleration at the rear wheel when it impacted the door threshold, and when the rear wheel traversed the 2 cm descent and ascent ($p=0.043$, $p=0.001$, $p=0.001$, respectively) and FW peak accelerations at the rear wheel when it impacted and left the door threshold, and when the rear wheel descended 2 cm ($p=0.049$, $p= 0.001$, $p= 0.005$, respectively).

With suspension, RMS and total VDV significantly decreased 14% and 10- 22% respectively ($p=0.011$, $p=0.004$). There were no significant differences between the rigid and suspended chair in total vibration power in frequency octaves most harmful in human exposure (4 – 12 Hz). The results of wheel diameter were not evaluated because there were significant differences in time spent over the obstacles between the two diameters (door threshold $p= 0.018$, truncated domes $p= 0.028$, 2 cm descent $p= 0.029$, 2 cm ascent $p = 0.024$). However, there were not differences in time spent over the obstacles between rigid and suspended conditions ($p \geq 0.064$).

The results indicate the aftermarket rear wheel suspension reduces some aspects of shock and vibration exposure, specifically at the rear wheel. While low back pain, neck pain, discomfort, and muscle fatigue correlate with shock and vibration exposure there is no set threshold of reduction in shock and vibration exposure to decrease the health risks with exposure. Considering how much time tilt-in-space users spend in their wheelchairs, we expect the observed reductions in shock and vibration with the use of the aftermarket rear wheel suspension may decrease the health risks, such as pain and muscle fatigue.

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Chapter I: INTRODUCTION

In 2010, the U.S. Census estimated 3.6 million people utilized wheelchairs resulting from a variety of physical disabilities including spinal cord injuries, muscular dystrophy, cerebral palsy, amputations, and multiple sclerosis (Brault, 2010; Physical and Mobility Impairments: Information and News, 2015). With advances in medicine, the number of people with physical disabilities requiring wheelchair use is projected to increase markedly in the years ahead, requiring a need for quality, cost effective wheelchairs (Allen, 2006; Cooper R.A., 2006).

Many wheelchair users may have secondary co-morbid conditions in addition to their primary physical disability. The secondary health conditions range from overuse shoulder joint injuries to depression, muscle spasms, osteoporosis, and chronic pain (Kloosterman, 2016; Boninger, 2003; Westerhoof, 2011; Jensen, 2013; Cooper R.A., 1995). Exacerbating the problem is the prolonged time spent in the seated position, which is the most vulnerable position for vibration transmission (VanSickle, 2000). Research has demonstrated wheelchair users experience high magnitude single shocks and low magnitude continuous vibrations exceeding the International Organization for Standardization (ISO) 2631-1 standards (VanSickle, 2001; Wolf, E., 2004; Wolf, E., 2007; Garcia-Medenz, 2013). ISO 2631-1 presents a technique to analyze shock and vibration exposure in seated humans through computing a frequency weighted (FW) acceleration, which may be analyzed for instantaneous peak accelerations, a FW root-mean square (RMS), and vibration dose value (VDV). When there are many individual periods of vibration exposure, the individual exposures may be combined through calculating the total VDV. Instantaneous peak acceleration, or shock, and whole-body vibration exposure has been correlated with low back pain, neck pain, muscle fatigue, early degeneration of the lumbar spine, herniated lumbar discs, and spinal discomfort and pain (Bonvenzi, 1996; Milosavljevic, 2011; Ebe, 2000; Mansfield, 2014; Bonvenzi, 2015; Zimmerman, 1993; VanSickle, 2001; Requejo, 2008; Maeda, 2003).

In an effort to prevent secondary injuries and improve comfort, wheelchair manufacturers have integrated suspension systems on both motorized and manual self-propelled chairs either at the front, rear, or both wheels. Elastomers, springs, and spring damper units are the most common suspension types for wheelchairs (Kwarciak, 2008). Research has demonstrated that suspension systems decrease peak accelerations and vibration exposure in relation to the material type and design of the system over common obstacles such as curb descents, truncated domes, and rumble strips (Cooper, 2003; Raquejo, 2008; Kwarciak, 2008; Requejo, 2009). Current research on rear wheel suspension typically evaluates peak accelerations over the obstacle (regardless of which wheel was accelerated), or will isolate the shock impact to only at the rear wheel. Only Gregg et al. (1988), when evaluating a front caster wheel suspension, specifically isolated the front and rear wheels. Gregg et al. (1988) concluded the front wheels are most responsible for high peak accelerations, suggesting a need to examine any suspension system's efficacy on both front and rear wheels.

Increased health risk resulting from shock and vibration exposure lies mainly within the natural frequency range of the seated operater (4 – 12 Hz) due to resonating effects (Cooper, R.A., 2003). Examining the power spectral density reveals power per frequency, with the area under the power spectral density providing the total power (Cooper, R.A., 2003). From a frequency perspective, the goal of suspension is to either decrease the total power per octave or shift the total power to higher octaves outside of the frequencies of highest risk (Cooper, R.A., 2003). Despite reductions in shock and vibration magnitudes with previously examined rear wheel suspension systems, there has been minimal change in their frequency content from 4 – 12 Hz (Cooper, 2003; Raquejo, 2008; Kwarciak, 2008; Requejo, 2009).

Not all manual wheelchairs are self-propelled. An attendant propelled type of manual wheelchair is the tilt-in-space design. The tilt feature allows the seat to stay in a vertical position or to tilt back up to 55 degrees. The chairs are designed so the back and thighs are at right angles regardless of the seat angle (Dicianna, 2008). Although research on the benefits of tilt-in-space manual wheelchairs is limited and inconclusive (Jan Y., 2013, Harrand, 2014; Schofield, 2012; Jan, Y., 2013), tilt-in-space wheelchairs claim to address the adverse impacts associated with pressure management, spasticity, respiratory and digestive complications, sitting tolerance, pain, edema, postural realignment, pressure sores, and hypotension (Dicianna, 2008). In an attempt to keep their mass manageable, attendant propelled tilt-in-space wheelchairs are not manufactured with suspension. Only recently has an aftermarket system become available. QuadshoX LLC (Fort Collins, CO) has recently introduced a relatively lightweight spring-damper suspension system for the rear axle. In rear wheel suspension, spring-damper units have shown the greatest decreases at the rear wheel in peak accelerations and forces as well as decreases in vibration dose value (VDV) (Requejo, 2008; Kwarciak, 2008; Requejo, 2009). However, QuadshoX attaches at the rear wheel using a patented method that is different from the previously researched rear wheel suspension systems. Besides manual tilt-in-space wheelchairs being slightly heavier than other manual wheelchairs, they also have an increased distance between the front and rear wheels. Both of these factors may affect the shock and vibration reducing capabilities of the wheelchair in addition to the new design. To our knowledge no research has been specifically performed evaluating manual tilt-in-space wheelchairs.

Wheelchair suspension systems and seat cushions have been explored for their shock and vibration reducing abilities but to our knowledge wheel diameter has not been investigated (Cooper, 2003; Requejo, 2008; Kwarciak, 2008; Requejo, 2009; Wolf, 2004; DiGiovine, 2003). Larger diameter wheels roll over obstacles more easily than smaller wheels, and once up to speed, larger diameter wheels more readily maintain speed (Steiner, 2016). Therefore, wheel diameter may influence the shock and vibration transferred to the user, independent of whether it is a rigid wheelchair or one equipped with suspension.

The purpose of this research was to evaluate the effect of the newly available rear wheel suspension system on shock and vibration attenuation and rear wheel diameter for manual tilt-in-space wheelchairs. We hypothesized the use of suspension and a larger diameter wheel will decrease shock and vibration exposure compared with rigid and smaller diameter wheel, respectively. The results may be utilized by clinicians and wheelchair users for proper selection as well as aid in the future development of wheelchair suspension and wheels.

Specific Hypotheses

1. The newly available rear suspension will decrease weighted and un-weighted peak resultant accelerations at the rear wheel over an exterior door threshold, 2 cm descent, and 2 cm ascent.
2. The newly available rear suspension will decrease weighted and un-weighted peak resultant accelerations at the front caster wheel over an exterior door threshold, 2 cm descent, and 2 cm ascent.
3. The newly available rear suspension will decrease the weighted RMS of the resultant acceleration when traversing over the simulated truncated domes compared to a rigid wheelchair.

4. The newly available rear suspension will decrease the VDV of the resultant acceleration and total VDV when traversing over an exterior door threshold, truncated domes, 2 cm descent, and 2 cm ascent compared with a rigid wheelchair
5. The newly available rear suspension will decrease the total power in the 4 – 12 Hz range over an exterior door threshold, 2 cm descent, and 2 cm ascent compared with a rigid wheelchair.
6. Larger rear wheels will have better capabilities at suppressing shock and vibrations over each of the obstacles (door threshold, simulated truncated domes, 2 cm descent, and 2 cm ascent) compared with smaller rear wheels regardless of suspension.

Chapter II: LITERATURE REVIEW

Wheelchairs and Users

A mobility impairment is a category of disability including people with varying types of physical disabilities such as a spinal cord injury, muscular dystrophy, cerebral palsy, amputation, spina bifida, multiple sclerosis, or pulmonary/heart disease (Physical and Mobility Impairments: Information and News, 2015). Many people with mobility impairments utilize wheelchairs. In 2010, the U.S. Census estimated 3.6 million people utilize wheelchairs (Brault, 2010). Wheelchairs enable people with physical disabilities to become mobile, increase their independence, and participate in their community, which ultimately benefits the user's physical health and quality of life (Armstrong, 2008). Because of advances in medicine, increased survival rates after traumatic accident, and decreased birth mortality rates, the number of people with mobility impairments is projected to increase (Allen, 2006). Therefore, a growing need for quality, cost effective wheelchairs exists (Allen, 2006; Cooper, R.A., 2006).

Depending on the physical disability and personal preference, an individual is fitted with either a manual (self or attendant propelled) or a motorized power wheelchair (Fact sheet on wheelchairs, 2010). Manual and motorized power wheelchairs come in many different designs, and have customizable features tailored to the individual. A suitable wheelchair will meet the user's functional needs and environmental conditions while providing proper fit and postural support (Fact sheet on wheelchairs, 2010). Powered wheelchairs allow greater mobility with the least physical exertion, and are easier to modify over time (Minkel, n.d.). Manual wheelchairs are typically lighter in weight, more reliable, easier to transport, and easier to overcome accessibility problems (Minkel, n.d.). The basic components of manual wheelchairs are: foot plates, heel loop, front caster wheel, rear wheel, frame, cross-brace, seat, headrest, armrest/arm, and handgrips (Cooper, R.A., 1995). Many types of manual wheelchairs are on the market including ultra-light weight, ergonomic, transport, standing, active, and tilt-in-space (Different Types of Wheelchairs, 2017).

Although research on the benefits of tilt-in-space manual wheelchairs is limited and inconclusive (Jan, 2013; Jan, Y., 2013; Harrand, 2014; Schofield, 2012), tilt-in-space wheelchairs are prescribed when an individual cannot independently change positions or shift their weight while seated upright (Figure 1) (Clinical Benefits of Tilt-in-Space). The tilt systems are manual or powered, giving the user or caregiver the ability to adjust the angle of the chair (Dicianna, 2008). The wheelchairs are designed so the back and thighs are at right angles regardless of the seat angle (Dewey, 2004). Tilt-in-space wheelchairs are more expensive and heavier than other types of manual wheelchairs. Because of the tilt feature, the chairs tend to have a longer frame which may cause some maneuverability and accessibility challenges (Dewey, 2004). A variety of patients utilize tilt-in-space wheelchairs, most typically individuals with spinal cord injury, muscular dystrophy, multiple sclerosis, cerebral palsy, and spina bifida. Clinicians do not use diagnosis alone when evaluating if a patient should have a tilt system (Watanabe, 2010). They consider other relevant capacities, employment and/or lifestyle interests. Tilt-in-space wheelchairs help to address concerns related to pressure management, spasticity, respiratory and digestive complications, sitting tolerance, pain, edema, postural realignment, pressure sores, and hypotension (Dicianna, 2008). Users of tilt-in-space wheelchairs self-report increased comfort, postural support, enhanced seating stability, relief of pressure, and the ability to rest sitting out of bed for extended periods (Dewey, 2004).

Figure 1. Quickie IRIS ® tilt-in-space manual wheelchair (IRIS, n.d.).

Health Hazards Associated with Wheelchairs

Secondary health conditions are defined as "physical or psychological health conditions that are influenced directly or indirectly by the presence of a disability or underlying physical impairment" (Adriaansen, 2016). "Premature aging", the earlier occurrence of health conditions usually associated with aging such as diabetes and cardiovascular disease, is common in individuals with physical disabilities (Jensen, 2013). Shoulder joint injuries in hand-rim wheelchair users occur frequently because of overuse from the repetitive nature of many activities of daily living (Kloosterman , 2016; Boninger, 2003; Westerhoof, 2011). Wheelchair users have higher rates of lost bowel and bladder control, blood circulatory problems, depression, muscle spasms, fatigue, and osteoporosis (Jensen, 2013).

Wheelchair users also have higher prevalence of back pain, pelvic pain, and neck pain (Jensen, 2013; Cooper, R.A., 1995; Boninger, 2003). Wheelchair users typically have abnormal body postures because of their physical impairment, years of sitting improperly, or both (Cooper, R.A., 1995). Poor posture and improper support cause spinal deformities such as scoliosis, lordosis, and kyphosis as well as abnormal pelvic tilts or rotations, and loss or reduction of hip range of motion (Cooper, R.A., 1995). Poor posture, working with the arms above shoulder height, and exposure to whole body vibrations are risk factors contributing to neck pain (Boninger, 2003). Adriaansen et al. (2016) reported some secondary health conditions of wheelchair users associated with a lower quality of life are musculoskeletal pain, pressure ulcers, and problematic spasticity. In order to improve community participation and increase the quality of life within the wheelchair population, research has been dedicated to finding risk factors associated with common secondary health conditions. Wheelchair users are mainly in a seated position and exposed to whole body vibrations over long periods and therefore they have an increased risk of developing secondary injuries such as low back pain (Griffin, 1990).

Wheelchair users are exposed to shock and vibrations. Vibrations "are the mechanical oscillations of an object about an equilibrium point", and vibration energy can enter the body through any point of contact to experience whole-body vibrations (Canadian Centre for Occupational Health and Safety, 2008). Mechanical shocks are single high magnitude accelerations also entering the body through a seat, floor, or any point of contact. Assessing the risks of shock and whole-body vibration exposure in regard to health consequences started in occupational groups most susceptible such as drivers of off-road vehicles, industrial vehicles, and buses (Bovenzi, 1996). Occupational drivers reported low back pain, early degeneration of the lumbar spine, and herniated lumbar discs at higher rates than the rest of the population

(Bovenzi, 1996). The role of whole body vibration's impacts are not fully understood in occupational exposure because individuals also have periods of prolonged sitting, bending forward, and frequent twisting of the spine which could also contribute to back pain (Bovenzi, 1996).

The International Organization for Standardization (ISO) established methods to quantify shock and whole-body vibrations exposure and risks associated with exposure documented in the ISO 2631-1 (International Organization for Standardization (ISO), 1997c). There are two main parts of the ISO 2631-1: (1) human health and (2) comfort and perception. Elevated health risks, such as low back pain and neck pain, correlate with whole-body vibration exposure (Milosavljevic, 2011; Bovenzi, 2015). Investigators assessing comfort and perception report higher levels of discomfort at higher vibration magnitudes (Ebe, 2000). Mansfield et al. (2014) reported occupants of automobiles have increased discomfort with time, but more discomfort is experienced when the subjects are exposed to whole-body vibration. Their research described a model of the relationship between seat shape and materials, long-term discomfort, and vibration. They concluded that ignoring parts of the model, seat characteristics and vibration exposure, will not adequately represent the components contributing to perceived discomfort (Mansfield, 2014). Many studies document the correlation of developing back pain with vibration exposure, but no direct causative relationship necessarily exists between the two (Bovenzi, 1996; Milosavljevic, 2011; Mansfield, 2014; Bovenzi, 2015). Vehicle investigations on the relationship between shock and vibrations and a person's health motivated the research of shock and vibration exposure in wheelchair users especially since this population has a higher prevalence of neck pain, back pain, and spinal disorders (Jensen, 2013; Boninger, 2003; Cooper, R.A., 1995).

The ISO 2631-1 states vibrations should be evaluated separately in each direction. The document states x is the anterior/posterior direction, y is the medial/lateral direction, and z is the vertical direction (International Organization for Standardization (ISO), 1997c). In assessing the health effects of vibration at the seat, the vibration evaluated should be the highest acceleration. If the accelerations are comparable in two or more directions, they can be combined in the evaluation (International Organization for Standardization (ISO), 1997c). The vibration total value, a_v, is calculated as *follows:*

$$a_v = (1.4a_{wx}^2 + 1.4a_{wy}^2 + a_{wz}^2)^{\frac{1}{2}} \quad (1)$$

where 1.4, 1.4, and 1 are the multiplying factors k_x, k_y, and k_z, respectively, as defined by ISO 2631-1, and where a_w is the instantaneous FW acceleration in the x, y, and z directions .

The ISO 2631-1 frequency-weightings put more emphasis on the frequencies most harmful to the human body, (4 – 12 Hz) (Figure 2). The weightings are defined as follows. The data is processed through a combination of 4 filters defined by the ISO 2631-1. The first two are Butterworth high pass and low pass filers, combining to form a band-pass filter defined as:

$$|H_h(p)| = \left|\frac{1}{1+\frac{\sqrt{2}\omega_1}{p}+\left(\frac{\omega_1}{p}\right)^2}\right| \quad (2)$$

$$|H_l(p)| = \left|\frac{1}{1+\frac{\sqrt{2}p}{\omega_2}+(p/\omega_2)^2}\right| \quad (3)$$

where

$H_h =$ High pass filter

$H_l =$ Low pass filter

p= Laplace domain variable

$\omega_1 = 2\pi f_1$

$f_1 = 0.4$ Hz

$\omega_2 = 2\pi f_2$

$f_2 = 100$ Hz

Two additional filters (Equation 4 and Equation 5), the acceleration-velocity transition and upward step, respectively) were used to weight the amplitude at frequencies depending on the effect they have on the human body in the vertical direction. One additional filter (Equation 4 acceleration-velocity filter) was used for weighting the vibrations in the horizontal plane. The acceleration-velocity transition filter is proportional to acceleration at lower frequencies, and proportional to velocity at higher frequencies (International Organization for Standardization (ISO), 1997c). The upward step filter takes into account the steepness of the slope, and is proportional to jerk (International Organization for Standardization (ISO), 1997c).

$$|H_t(p)| = \left|\frac{1+p/\omega_3}{1+p/(Q_4\omega_4)+(p/\omega_4)^2}\right| \tag{4}$$

Where,

$H_t =$ Acceleration-velocity transition filter

$\omega_3 = 2\pi f_3$

$\omega_4 = 2\pi f_4$

$Q_4 = 0.63$

$f_3 = f_4 = 12.5$ Hz (for vertical motion)

$f_3 = f_4 = 2$ Hz (for lateral motion)

$$|H_s(p)| = \left|\frac{1+p/Q_5\omega_5+(p/\omega_5)^2}{1+p/(Q_6\omega_6)+(p/\omega_6)^2} \cdot \left(\frac{\omega_5}{\omega_6}\right)^2\right| \tag{5}$$

Where,

$H_s =$ Upward step filter

$Q_5 = Q_6 = 0.91$

$f_5 =$2.37 Hz

$f_6 = 3.35$ Hz

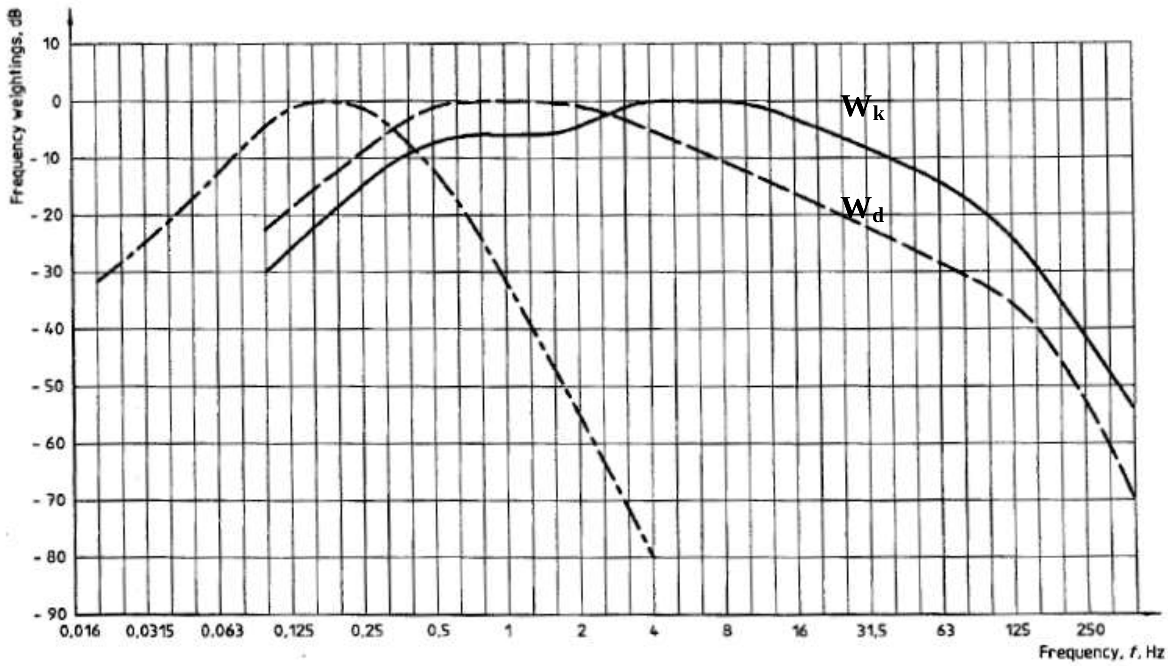

Figure 2. Frequency weighting curves for the principal weights (W_k= z-axis, W_d = x and y-axis). Graph from the International Organization for Standardization (ISO), 1997c.

Vibration assessment should always include measurements of the weighted root-mean-square (RMS) acceleration, a_w, and be expressed in meters per second squares (m/s^2) (International Organization for Standardization (ISO), 1997c). The weighted RMS acceleration is calculated using:

$$a_w = \left[\frac{1}{T}\int_0^T a_w^2(t)dt\right]^{1/2} \tag{6}$$

where $a_w(t)$ is the weighted acceleration at time *t*, and T is the duration of the measurement in seconds. The ratio of the maximum peak value of the frequency-weighted acceleration to its RMS value is referred to as the crest factor. It is used to describe the severity of the vibration. For vibration crest factors below or equal to 9, the basic evaluation method is sufficient (i.e. using frequency-weighted RMS). If a vibration crest factor is above 9, the ISO 2631-1 suggests using the running RMS or fourth power vibration dose value. The fourth power vibration dose value (VDV) is purported to be more sensitive to peaks than weighted RMS, and is in meters per second to the power 1.75. VDV is calculated using:

$$\mathrm{VDV} = \left\{\int_0^{\mathrm{T}} \left[\mathrm{a}_{\mathrm{w(t)}}\right]^4 \mathrm{dt}\right\}^{\frac{1}{4}} \tag{7}$$

where $a_w(t)$is the instantaneous frequency-weighted acceleration at time *t*, and T is the duration of the measurement. When vibration exposure consists of more periods, the sum of the individual exposures to find the total exposure can be calculated by:

$$VDV_{total} = \left(\sum_i VDV_i{}^4\right)^{\frac{1}{4}} \tag{8}$$

A health guidance caution zone has been proposed to assess exposures in the range of 4 hours to 8 hours. When plotting the zone, the y-axis is the weighted RMS acceleration in m/s^2 and the x-axis the exposure in hours. The lower bound for 8 hours of exposure is 0.45 m/s^2 and the upper bound for 8 hours of exposure is 0.90 m/s^2. Between and above the bounds indicate increased health risks associated with that level of vibration exposure as seen in Figure 3. The ISO 2631-1 states two different daily vibration exposures are equivalent when:

$$a_{w1} * T_1^{\frac{1}{2}} = a_{w2} * T_2^{\frac{1}{2}} \tag{9}$$

where a_{w1} and a_{w2} are the weighted RMS acceleration values for the different exposures, and T_1 and T_2 are the corresponding durations for the exposures.

Within the ISO 2631-1, other studies indicate a relationship when:

$$a_{w1} * T_1^{\frac{1}{4}} = a_{w2} * T_2^{\frac{1}{4}} \tag{10}$$

f

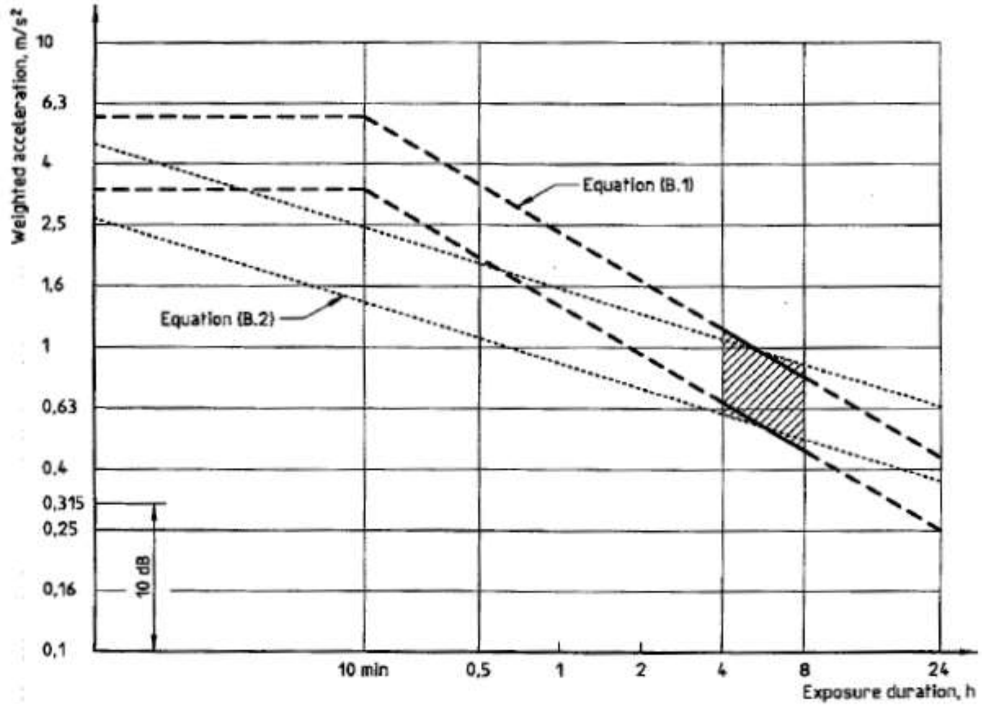

Figure 3. Equation (9) represents the (B,1) dashed line, and equation (10) represents the (B,2) dashed line. The area between the lines represents vibration exposure levels potentially harmful, and the upper boundary represents the vibration exposure level most likely to negatively impact one's health. Graph from the International Organization for Standardization (ISO), 1997c.

Another component of shock and vibration exposure is the frequency at which the exposure is occurring. Increased health risk due to shock and vibration exposure is in the frequency range from 4- 12 Hz. The range is near the natural frequency of seated humans therefore shocks and vibrations are more readily absorbed by the body (Cooper, R.A., 2003). At the resonant frequency, the amount of energy absorbed is maximized which potentially causes muscle fatigue, and damage to organs. The power spectral density shows the power disturbed over frequency (Cooper, R.A., 2003). For human exposure, the power spectral density (PSD) is looked at in frequency octaves. The octaves are calculated using:

$$f_2 = \left(2^{1/3}\right) * f_1 \tag{11}$$

starting with f_2 =500Hz, and then solving for f_1 would give 315 Hz. The next octave starts with f_2= 315, and then solving for f_1 would give 250 Hz. The octave calculation is repeated until reaching 1.25 Hz. The area under the curve in each octave approximates the total vibration power.

A few other methods have been used when assessing shock and vibration exposure in wheelchair users with suspension systems. Some studies have investigated changes in peak accelerations whereas others investigated differences in peak seat forces and head accelerations (Cooper, R.A., 2003; Kwarciak, 2008; Requejo, 2008; Requejo, 2009). If the seat force magnitudes and accelerations are small, then the shock-absorbing ability of the wheelchair and user is good. If the seat force magnitude is high whereas the head acceleration is low, then the user is absorbing most of the shock (Requejo, 2009). These authors think analyzing the forces and accelerations, rather than frequency domain accelerations (process described previously as advised by the ISO 2631-1 standards) , quantifies what is happening during mechanical shocks that will allow a better understanding of the relationship between shock, vibration, comfort level, and onset of injury (Requejo P. M.-G., 2009). However, this group is the only one utilizing this particular method, and have found results similar to those who utilize the ISO 2631-1 methods to analyze shock and vibration exposure.

VanSickle et al. (2001) reported wheelchair users have shock and vibration exposure exceeding the International Organization for Standardization (ISO) 2631-1 standards at the seat pan and head. The study had subjects traverse obstacles typically experienced by wheelchair users in their own personal chairs. The obstacles included truncated domes, industrial carpet, door threshold, climb up a 1.27 m long ramp and 5.0 cm drop, rumble strips, and three sinusoidal bumps. Shock and vibration exposure has been linked to muscular fatigue, through monitoring muscle activity using electromyography, and back injury in wheelchair users (Zimmermann, 1993; VanSickle, 2001; Requejo, 2008). Additionally, shock and vibration exposure decreases the rider's comfort, and limits the user's activity and participation (DiGiovine, 2000; Cooper, R.B., 2006). The effects of shock and vibrations vary from person to person depending on an individual's susceptibility and perception of thresholds, body composition, posture, and the frequency, direction, magnitude, and duration of the vibration (DiGiovine, 2003).

Rather than assessing exposure to vibration in a laboratory setting, Garcia-Medenz et al. (2013) used the ISO methods to evaluate the health risk associated with whole-body vibration exposure to wheelchair users in the community over a two week period. The results concluded that 100% of the subjects (a total of 37 wheelchair users with a combination of rigid and suspended chairs) were exposed to vibrations at the seat that were within or above the health-caution zone established by the ISO 2631-1 standards (Garcia-Mendenz, 2013).

How to Minimize Shock and Vibration Exposure

In industries exposed to high levels of vibration, seating systems (suspension, cushions, and back supports) along with posture have been investigated to determine the most efficient way to minimize vibration transmission (Lundstrom, 1998; Wilder, 1994). Similarly, a few different areas have been assessed in regard to shock and vibration exposure in wheelchair users: (1) transmission of vibration through different seat cushions (2) vibration exposure over various surface configurations and (3) suspension elements to reduce vibration exposure.

Seat Cushions and Surface Configurations

Medical implications associated with improper seating are pressure sores, pelvic problems and spinal deformities (Cooper, R.A., 1995). Proper seating should place all joints, (hip, knee, elbow), at or near the individual's neutral position or 90 degrees (Cooper, R.A., 1995). Seat cushions should provide good pelvis support to reduce the risk of pressure sores as well as poor pelvic alignment and the health associated consequences (Cooper, R.A., 1995). Wolf et al. (2004) and DiGiovine et al. (2003) investigated whether certain wheelchair cushions reduced the transmission of whole body vibrations to the wheelchair user more than other cushions. Wolf et al. (2004) found some seat cushions, Invacare Pindot and the Varilite Solo, were better at decreasing the transmission of vibrations to the user. DiGiovine et al. (2003) also found differences in the vibrations absorbed by cushions.

Wolf et al. (2007) assessed vibration exposure in manual and motorized power wheelchairs over different sidewalk surfaces (variations of concrete and brick configurations). They reported significant differences in whole body vibration exposure between surfaces for manual and motorized powered wheelchairs, and concluded some surfaces could produce levels of vibration exposure that may cause secondary injuries, such as muscle fatigue and pain, over time (Wolf, 2007).

Suspension Units

Vehicles have suspension systems to increase comfort and performance (Nielens, 2004); therefore, some manufacturers have begun to incorporate suspension systems within the wheelchair design. To reduce transmission of shock and vibration to the user, the addition of a suspension system can be included on the rear wheels, the front caster wheels, or both. Elastomers, springs, and spring and damper units are the most common suspension systems (Kwarciak, 2005). Elastomers (rubber springs) compress under a load, and are small as well as lightweight (Abbott, 1995). Springs support the weight of the frame but can continue to oscillate once the wheelchair has started to move, therefore, dampers are added to the spring system to eliminate the oscillations within a few cycles (Abbott, 1995).

Some commercially available self-propelled manual wheelchairs with suspension systems are Boing! ® and ShockBlade ®, Colours 'N Motion Inc. (Corona, California), Quickie XTR ®, Sunrise Medical (Carlsbad, California), and A4 ®, Invacare Corp. (Elyria, Ohio). Boing! ®, Figure 4**,** uses an A-arm suspension system (Colours Boing! Wheelchair). The A-arm suspension system is common in U.S. cars, and consists of a wishbone-shaped arm with two mounting points on the frame and one at the wheel that directly connects to a coil spring- damper (Wheelchair Suspension/Shock Absoprtion, 2009). ShockBlade ® also uses A-arm suspension but additionally is coupled with a front caster wheel suspension unit (Colours in Motion, 2010). Quickie XTR ®, Figure 4, uses a single spring-damper (hydraulic) supporting the seat (GT (TM)). The spring-damper unit combination is capable of reducing vibrations and oscillations (Kwarciak, 2005). The A4 ®, Figure 4, uses elastomer shocks, and Invacare states the clinical benefits of suspension are decreased pain, decreased reflexive spasms, improved stability, and greater comfort as well as endurance (Invacare, 2003). One opinion of wheelchairs with suspension is that they are harder to propel because the suspension on the chair causes more inertia to overcome. Therefore, the suspension unit must minimize weight and have configurations requiring minimal frame assembly (Requejo, 2008).

In addition to wheelchairs manufactured with suspension units, there are also after-market suspension units available for wheelchairs. One popular system currently on the market is the Frog Legs ™ suspension for front caster wheels (Figure 4). The suspension system uses elastomers made of polyurethane (Frog Lets Inc.). Frog Legs ™ claims the caster suspension eliminates 76% of all the vibration in the frame, and helps with spasticity, secondary injuries, fatigue, crystallization of body fluids, neuropathy, pressure sores, and longevity of the chair (Gregg, 1998; Frog Lets Inc.).

Figure 4. (a). Boing! ® (Colours Boing! Wheelchair) (b). Quickie XTR ® (GT) (c). A4 ® (Invacare Top End A-4 Wheelchair) (d). Frog Legs Big-Rig front caster (Big-Rigs)

Success of Suspension on Wheelchairs

Cooper et al. (2003) assessed vibration exposure in manual wheelchairs with and without suspension (rear-wheel and front caster forks) using test dummies. They found significant differences in peak accelerations at the seat and footrest between the wheelchairs when standard caster forks were used versus Frog Legs ™ (Cooper, R.A., 2003). Frog Legs ™ suspension decreased the peak accelerations, but still had peak frequencies occurring from 4 to 12 Hz which may elevate health risks. Rear-suspension reduced the total power per octave between 7.81 and 9.84 Hz, but concluded suspended chairs were not superior to the rigid frame wheelchairs (Cooper, R.A., 2003). One limitation within the study was the wheelchairs were categorized as either suspension or rigid even though there were three suspension wheelchairs (varying from elastomer to spring-damper suspension systems) and three rigid wheelchairs (i.e. cantileverframe ultralight, folding lightweight, and ultralight). Grouping all suspension types together when assessing the shock and vibration suppressing abilities may minimize the impact of certain suspension configurations.

Requejo et al. (2008) assessed the effect of rear suspension and speed on seat forces and head accelerations in self-propelled manual wheelchairs, (rigid frame, Boing! ®, A4 ®, and Quickie XTR ®), using load cells and an accelerometer at the head. All wheelchairs with suspension reduced the forces and accelerations but did not perform similarly in shock- and vibration- suppression (Requejo, 2008). The peak seat forces indicated the shock- and vibration-absorption performance of the wheelchair, and the larger difference in peak seat forces indicates a greater amount of damping the chair provides the users (i.e. presumably improving comfort and reduced chances of secondary injury). Boing! had the lowest forces and head accelerations suggesting a better shock- and vibration suppressing performance, but Boing! was the heaviest of the suspension systems tested therefore could lead to increased risk of upper-limb secondary injury from hand rim propulsions and could be difficult to transport (Requejo, 2008).

Kwarciak et al. (2008) looked at the effectiveness of vibration suspension performance of self-propelled suspension manual wheelchairs (Boing! ®, Quickie XTR ®, A4 ®) compared to rigid frames during a curb decent (i.e. high-impact task). They concluded that suspension manual wheelchairs provide some vibration suppression but that depends on the orientation of the wheelchair during the task (Kwarciak, 2008). Interestingly, the Quickie XTR ® performed better at suppressing vibrations during a curb decent task whereas, in the study performed by Requejo et al. (2008), Boing! ® had performed better at suppressing vibrations.

Requejo et al. (2009) also assessed the seat force and head accelerations manual wheelchair users experience during curb descents with rigid and rear suspension manual chairs.

They measured forces as well as head accelerations, rather than frequency content similar to Kwarciak et al. (2008), experienced during a curb descent to quantify single mechanical shocks since the ISO 2631 is for near-constant vibration exposure that does not distinguish vibration containing single shocks (Requejo, 2009). The authors state the person and/or wheelchair absorb the energy when exposed to shocks and vibrations (Requejo, 2009). If the seat force magnitudes and accelerations are small, then the shock-absorbing ability of the wheelchair and user is good. If the seat force magnitude is high whereas the head acceleration is low, then the user is absorbing most of the shock (Requejo, 2009). Requejo et al. (2009), as did Kwarciak et al. (2008), found the Quickie XTR ® had the least seat force and head accelerations. They concluded, similarly as did the study by Kwarciak et al. (2008), that suspension systems can reduce the magnitude of the force by reducing head accelerations, but depends on the type and design of suspension system (Requejo, 2009).

Wheelchair Wheels and Tires

Wheelchairs have a set of (usually) larger diameter rear wheels and a set of smaller front caster wheels. The lighter the wheels, the faster the user can accelerate (Medola, 2014). The front caster wheel diameter and position can increase rolling resistance thus making it more challenging to maintain constant velocity. Smaller diameter wheels and if the front and rear wheels are close together are examples of how to increase rolling resistance that will negatively impact the user by making it more challenging to maintain a constant velocity (Medola, 2014).

There are two types of tires for wheelchairs: pneumatic or solid. Pneumatic tires are good at absorbing shock and vibration which increases the users' comfort (Medola, 2014). Solid tires have a higher rolling resistance, (or they make it harder to maintain constant velocity), compared to pneumatic tires, but they are more durable and require almost no maintenance (Medola, 2014). To our knowledge, no research has been conducted on whether or not wheel diameter impacts shock and vibration absorption. Larger diameter wheels roll over obstacles more easily than smaller diameter wheels, and once up to speed, it is easier to maintain said speed (Steiner, 2016).

Summary

Manual tilt-in-space wheelchairs are not sold with suspension, and, until recently, there was no after-market rear wheel suspension available for use. The QuadshoX, LLC (Fort Collins, CO) suspension kit for tilt-in-space wheelchairs utilizes a spring damper unit at the rear wheel similar to Boing! ® and ShockBlade ®. Instead of using A-arm suspension though, QuadshoX has moment arm brackets that attach at the rear axle and carriage shaft, and the spring/damper unit attaches to the carriage shaft and moment arm bracket (Figure 5). The suspension kit replaces the manufactured axle bracket on Quickie IRIS ® (Sunrise Medical, Phoenix, AZ) tilt-in-space wheelchairs (Figure 6). Other rear wheel suspension types have shown capabilities of reducing some components of shock and vibration exposure but inconsistencies exist therefore, research needs to be conducted to evaluate the impact of QuadshoX after-market rear wheel suspension as well as the role of wheel diameter on suppressing shock and vibration. Based on the research reviewed, we hypothesize the wheelchair with large diameter wheel with suspension would suppress shocks and vibrations more compared with the rigid wheelchair with small diameter wheels.

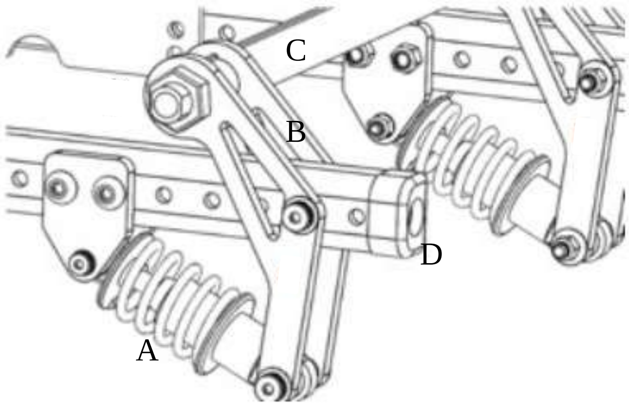

Figure 5. Image of QuadshoX, LLC rear wheel suspension installed on a Quickie IRIS ® tilt-in-space wheelchair. A. spring/damper unit B. Moment arm C. Rear axle D. Carriage shaft (Instruction Manual: QX-1L, 2017).

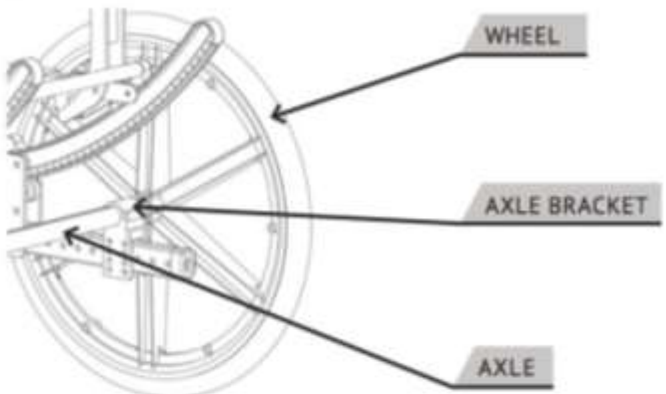

Figure 6. Image looking at the chair from the rear with manufacturing axle bracket on (prior to suspension installation) (Instruction Manual: QX-1L, 2017)

Chapter III: METHODS

Subjects

Ten non-wheelchair users (7 men, 3 women: age= 22.1 ± 3.36 yrs., height =1.75 ± 0.067 m, and mass= 73.9 ± 8.87 kg (mean $\pm$ SD)) voluntarily participated in this institutional review board-approved study (Appendix) after providing written informed consent (Appendix). Participants were healthy (pain and injury free at the time of data collection) age 18 years or older recruited from the student population.

Instrumentation

A single Quickie IRIS ® Tilt-in-Space manual wheelchair (17.69 kg, aluminum frame, maximum user weight capacity: 136 kg, 1.19 m L x 1.02 m W x 1.37 m H, front caster wheels: 0.20 m) manufactured by Sunrise Medical (Phoenix, AZ), was used by all subjects and for all trials. The tilt system was adjusted to a posterior tilt of 15° for all conditions. For the rigid trials, the wheelchair was kept as manufactured. For the suspension trials, the wheelchair was fitted with a QuadshoX, LLC (Fort Collins, CO) suspension kit (Figure 7). The suspension system was adjusted per manufacturer instructions for each subject. The preload on the coilover shocks were adjusted when the user was sitting in the chair. The preload adjustment collar was adjusted so there was a gap of 0.06 m between the lower spring perch and adjustment collar. Two solid Primo wheels of different diameter (Cheng Shin Rubber, Yuanlin, Taiwan), were also studied: a 0.381-meter (small) diameter wheel and a 0.508-meter (large) diameter wheel.

Figure 7. Posterior view of QuadshoX suspension kit installed on a Quickie IRIS ® Tilt-in-Space wheelchair. The suspension kit has moment arm brackets (green in image) that attach at the rear axle and carriage shaft, and the spring/damper unit attaches to the carriage shaft as well as moment arm bracket. The suspension kit replaces the manufactured axle bracket on Quickie IRIS ® (Sunrise Medical, Phoenix, AZ) tilt-in-space wheelchair.

A tri-axial accelerometer (Model339A31, PCB Piezotronics, Depew, NY) was mounted at the rear of the seat pan of the wheelchair (Figure 8). A custom data-collection program was written in LabVIEW software (National Instruments, Austin, TX) to interface with the data-acquisition assistant (DAQ, National Instruments, Austin, TX) sampling at 2000 Hz. Raw voltages were amplified with a gain of 10 by the power supply (Model 480B21, PCB Piezoelectronics, Depew, NY). The equipment (DAQ and power supply) was kept in a small pack fastened to the wheelchair, and connected to a hand held laptop computer.

Figure 8. The accelerometer and mount (circled) was securely attached to the rear seat pan of the wheelchair. Also visible is the QuadshoX suspension system on the right side of the wheelchair as well as the small diameter wheel.

Trials

The surfaces were designed by the investigators to represent common obstacles encountered by wheelchair users in their daily lives. An indoor course was made out of three sheets of plywood (laying over carpet/concrete, 1.22 meters wide x 2.44 meters long x 0.02 meters high). The first and last sheets of plywood remained in place, and the center sheet was interchanged between two surfaces with four different obstacles (Figure 9). The four obstacles included:

1. Simulated truncated domes (total length of 96 cm with 3.5 cm between each dome, domes: 1 cm high x 2 cm wide x 121.92 cm long)
2. Door threshold (1.27 cm high x 91.44 cm long (M-D Building Products, Inc., Oklahoma City, OK))
3. 2 cm descent made by overlapping one sheet of plywood over another
4. 2 cm ascent made by overlapping one sheet of plywood over another.

Figure 9. (a). Surface consisting of a door threshold (far) and simulated truncated domes (near). (b). Close up of the simulated truncated domes. (c). Close up of the 2 cm descent/ascent (descent if traversing from left to right, and ascent if traversing from right to left).

Data Collection

All rigid trials were collected first followed by the suspended trials on a separate day, separated by ~ 2 weeks. Subjects were pushed by the same trained investigator over all three obstacles. A minimum of three acceptable trials were recorded for each condition performed in the following order: rigid chair/small wheel (RS), rigid chair/large wheel (RL), suspended chair/small wheel (SS), suspended chair/large wheel (SL). Acceptable trials had to be within 0.2 seconds of each other as determined by hand timing. Each condition trial had to be within 0.2 s of the average as well as 0.2 s within the trials of the condition. The average for each surface was set by the RS trials. In order to achieve these requirements, usually 3-7 trials were recorded. During each trial subjects were instructed to stay as relaxed as possible, not reacting any more than necessary to the obstacle. The investigator pushed subjects as consistently as possible when traversing the obstacle.

Data Processing and Analysis

The accelerations were processed with a custom MATLAB code (version 8.4, The MathWorks, Inc; Natick, MA). Each channel was first zero-meaned to remove any potential DC offset and then converted from volts to m/s^2 by factoring out the gain and incorporating manufacturer supplied conversions. While each trial was hand timed as described above in an attempt to keep time spent over the obstacle in each condition the same, post-hoc analysis using the peak accelerations showed time over obstacles was significantly different. Therefore, the trials used for statistical analysis were selected based on this requirement instead. More specifically, first, the time over the trial for each of the four obstacles was found using a custom MATLAB code. The time over the obstacle was defined at the first peak acceleration to the last peak acceleration. Second, the rigid and suspended trial times were compared for each wheel diameter. The trials between the rigid and suspended conditions were selected for statistical analysis if the times were close, and all trials had a time difference less than 0.15 seconds between the rigid and suspended conditions. This selection process was different than originally planned (i.e. averaging the three timed trials), as now the analysis includes only a single trial per condition (RS, SS, RL, SL) and obstacle. However, the time over each obstacle was not significantly different between rigid and suspended trials using this modified approach, though it was not successful in removing time over obstacle differences between the large and small diameter wheels. After reviewing the data it was determined the gross measure of hand timing was not suitable to ensure consistency over each obstacle. Even if the trials were within 0.2 seconds of each other as well as the RS trials, this did not ensure the time spent over the obstacle was consistent for all the rigid and suspended trials.

The ISO 2631-1 states vibrations should be evaluated separately in each direction. When assessing the health effects of vibration at the seat, the vibration evaluated should be the highest acceleration. If the accelerations are comparable in two or more directions, they can be combined to evaluate (International Organization for Standardization (ISO), 1997c). Therefore, anterior/posterior and vertical directions were selected and combined for analysis.

Acceleration data was frequency weighted (FW), using a MATLAB algorithm adapted from Kwarciak et al. (2008), according to the standard vibration evaluation methods and parameters as stated in ISO 2631-1. The ISO 2631-1 recommends different frequency weighting for assessing vibrations' effects on health, comfort, and perception. The weightings put the most emphasis on the frequencies from ~ 2 – 12 Hz and the frequencies below and above this range gradually receive less weighting. Each direction is first weighted in isolation before being combined. The resultant FW acceleration equation used with x being the anterior/posterior and z being the vertical direction was:

$$a_w = (1.4a_{wx}^2 + a_{wz}^2)^{\frac{1}{2}} \tag{7}$$

where 1.4 and 1 are the multiplying factors kx and kz , respectively, as defined by ISO 2631-1.

As defined by the ISO 2631-1 standards, the crest factor is the ratio of the maximum peak value of the frequency-weighted acceleration to its root-mean-square (RMS) value (International Organization for Standardization (ISO), 1997c). For vibration crest factors below or equal to nine, the basic evaluation method is sufficient (i.e. using frequency-weighted RMS acceleration). Frequency-weighted RMS acceleration is expressed in meters per second squares (m/s^2).The frequency-weighted RMS acceleration, a_{rms}, is calculated using:

$$a_{rms} = \left[\frac{1}{T}\int_0^T a_w^2(t)dt\right]^{\frac{1}{2}} \tag{8}$$

where $a_w(t)$ is the instantaneous frequency-weighted acceleration calculated in (7) and T is the duration of the measurement. The RMS was calculated over the region of interest (Figure 11), which consisted of 500 hundred data points prior to the initial peak acceleration and 500 data points after the final peak acceleration. This ensured that all accelerations associated with the obstacle were incorporated into the calculation.

The fourth power vibration dose value (VDV) is typically more sensitive to peaks, and is in meters per second to the power 1.75 (International Organization for Standardization (ISO), 1997c). VDV is calculated using:

$$VDV = \left\{\int_0^T \left[a_{w(t)}\right]^4 dt\right\}^{\frac{1}{4}} \tag{9}$$

where $a_w(t)$is the instantaneous frequency-weighted acceleration calculated in (7) and T is the duration of the measurement. VDV, unlike RMS, is not time dependent. This allows for short durations of vibration exposure to be compared to the ISO 2631-1 Health Guidance Caution Zone. While it is unclear just how much more movement time a tilt-in-space wheelchair user typically has each day, the lower boundary of VDV for 8 hours of exposure is 8.5 $m/s^{1.75}$, and represents the minimum shock and vibration exposure potentially harmful. The upper boundary of VDV for 8 hours of exposure is 17 $m/s^{1.75}$, and represents the point where health risks are more likely. When vibration exposure consists of multiple isolated periods, the sum of the individual exposures to find the total exposure can be calculated by:

$$VDV_{total} = \left(\sum_i VDV_i{}^4\right)^{\frac{1}{4}} \tag{10}$$

Another component of shock and vibration exposure is the frequency at which the exposure is occurring. As previously described, increased health risk due to shock and vibration exposure is in the frequency range from 4- 12 Hz. The power spectral density (PSD) is a way to examine how power is distributed over frequency (Cooper, R.A., 2003). For human exposure, the PSD is examined in frequency octaves. The octaves were calculated using:

$$f_2 = \left(2^{1/3}\right) * f_1 \tag{11}$$

starting with f_2 =500Hz, and then solving for f_1 would give 315 Hz. The next octave starts with f_2= 315, and then solving for f_1 would give 250 Hz. The process continues until reaching 1.25 Hz. The area under the curve in each octave approximates the total vibration power in each.

Statistical Treatment

The data was analyzed for normality. Outliers were identified using box plots, and extreme outliers (greater than three box lengths) were removed. After removal of outliers, the data was determined to be normally distributed by assessing the ratios between skewness and its standard error, and kurtosis and its standard error.

The data was analyzed using a two by two (suspension by wheel size) repeated measures analysis of variance (ANOVA) to compare the mean differences between groups in time spent over the obstacle, un-weighted and weighted peak accelerations, weighted RMS, VDV, total VDV, and total power per octave. The independent variables were wheel diameter, and the use of suspension (rigid small diameter wheel (RS), rigid large diameter wheel (RL), suspended small diameter wheel (SS), and suspended large diameter wheel (SL)).

The variables analyzed over the door threshold were weighted and un-weighted peak accelerations A, B, C, and D, (Figure 10) as well as VDV. The variables analyzed over the simulated truncated domes were the FW RMS, and VDV (Figure 11). The variables analyzed over the 2 cm descent and ascent were weighted and un-weighted peak accelerations E, F, G, and H (Figure 12) as well as VDV. The total vibration power per octave was analyzed for all octaves and for all four obstacles (door threshold, truncated domes, 2 cm descent, and 2 cm ascent). All analyses were performed using IBM SPSS Statistics software Version 24.0 (IBM Corp., Armonk, NY, USA), with a significance level of $p < 0.05$. Main effects are reported unless otherwise indicated.

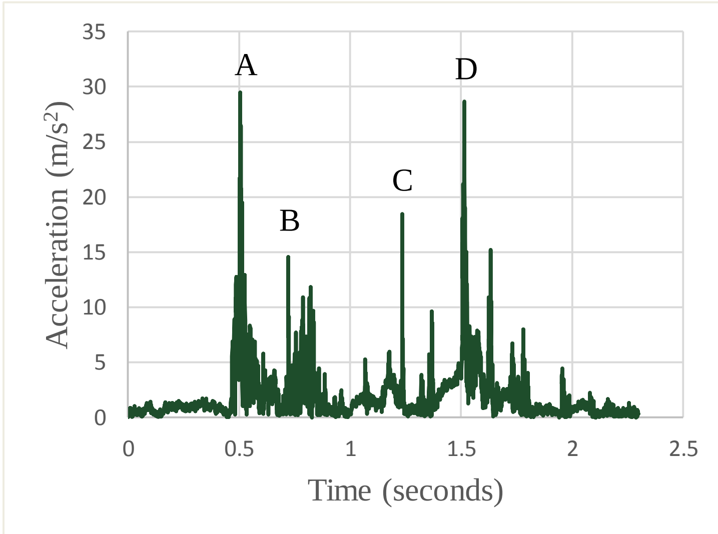

Figure 10. Raw un-weighted resultant acceleration profile of the wheelchair traversing the door threshold. Peak A= when the front caster wheel first hits the door threshold, Peak B= when the front caster wheel leaves the door threshold, Peak C= when the rear wheel first hits the door threshold, and Peak D= when the rear wheel leaves the door threshold.

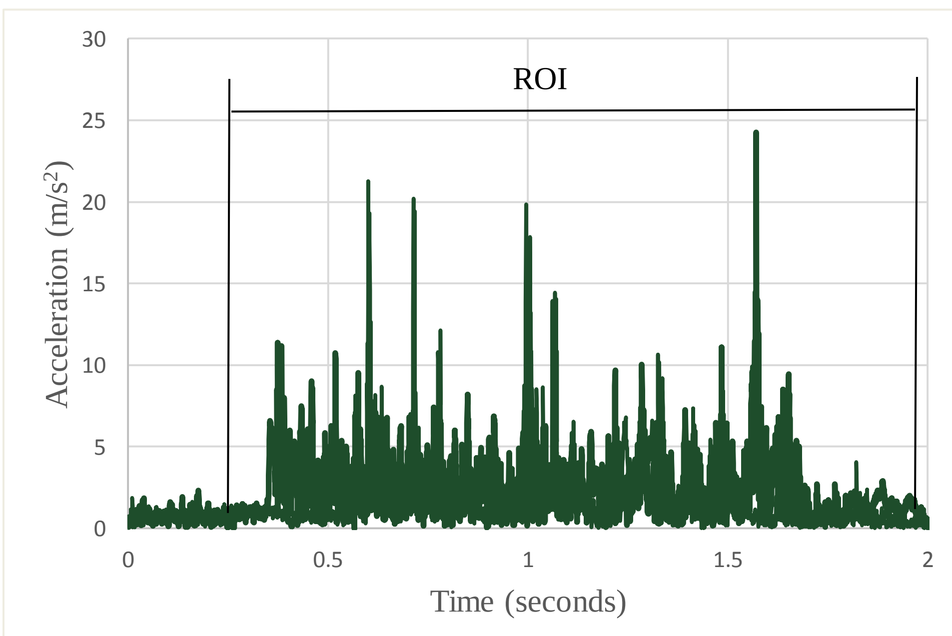

Figure 11. Raw un-weighted resultant acceleration profile of the wheelchair traversing the simulated truncated domes. ROI= region of interest where weighted RMS of the resultant acceleration was calculated.

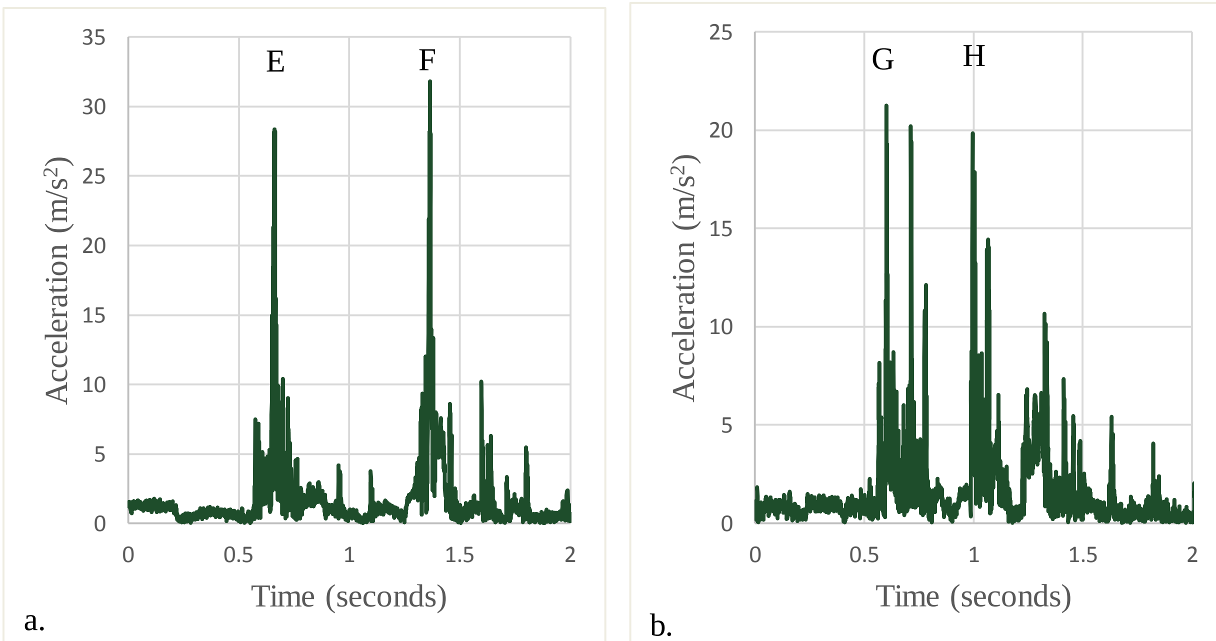

Figure 12. (a). Raw un-weighted resultant acceleration profile of the wheelchair over the 2 cm descent. Peak E= front caster wheel dropping off, and Peak F= rear wheel dropping off. (b). Raw resultant acceleration profile of the wheelchair over the 2 cm ascent. Peak G= front caster wheel ascending the obstacle, and Peak H= rear wheel ascending the obstacle.

Chapter IV: RESULTS

The time spent over the door threshold, simulated truncated domes, 2 cm descent, and 2 cm ascent between small and large diameter wheels was significantly different (p=0.0181, p=0.028, p=0.003, p=0.024, respectively). Therefore, the investigators deemed it inappropriate to report on any potential effects of wheel diameters since traversing obstacles faster or slower could affect accelerations independent of wheel diameter. The time over the door threshold, 2 cm descent, 2 cm ascent, and truncated domes between rigid and suspended chairs was not significantly different (p=0.660, p=0.508, p=0.252, 0.064, respectively). The group mean and standard deviations of the time over each surface are presented in Table 1. The door threshold and truncated domes were together on one surface (1.27 meters long from the start of the door threshold to the end of the truncated domes). The 2 cm descent and ascent were together on one surface (1.22 meters long from the start of the descent to the ascent).

A total of five outliers were removed from the door threshold obstacle, and no more than two outliers were removed from any single variable. For VDV over the door threshold, there was one outlier removed from the rigid large diameter wheel, and one outlier removed from the suspended large wheel condition. There were no outliers in the 2 cm descent obstacle, and there were two outliers removed for the VDV variable. One outlier was removed from the 2 cm ascent obstacle, and there was one outlier removed for the VDV variable.

There were a total of 24 variables in the frequency content analysis per condition (RS, RL, SS, SL). For a single condition, no more than 8 variables out of the 24 were affected by an outlier. For a single variable, there were no more than 2 outliers removed per variable for the door threshold, truncated domes, 2 cm descent, and 2 cm ascent.

Table 1. Group mean and standard deviations for the time spent over each surface.

	Time (s)			
	Rigid Small	Suspended Small	Rigid Large	Suspended Large
Door Threshold and Truncated Domes	4.12 ± 0.22	4.13 ± 0.23	4.18 ± 0.22	4.12 ± 0.24
2 cm Descent/Ascent	3.73 ± 0.18	3.74 ± 0.18	3.74 ± 0.17	3.72 ± 0.19

Peak Accelerations

Door threshold un-weighted mean peak accelerations calculated for each wheelchair condition and wheel diameter traversing the obstacles are presented in Figure 13. Un-weighted peak acceleration C (when the rear wheel first hits the door threshold) was significantly less with the use of suspension (p=0.043). Peak A (when the front wheel first hits the door threshold), B (when the front wheel leaves the door threshold), and D (when the rear wheel leaves the door threshold) were not significantly different between the rigid and suspended chair (p= 0.257, p= 0.950, p= 0.280).

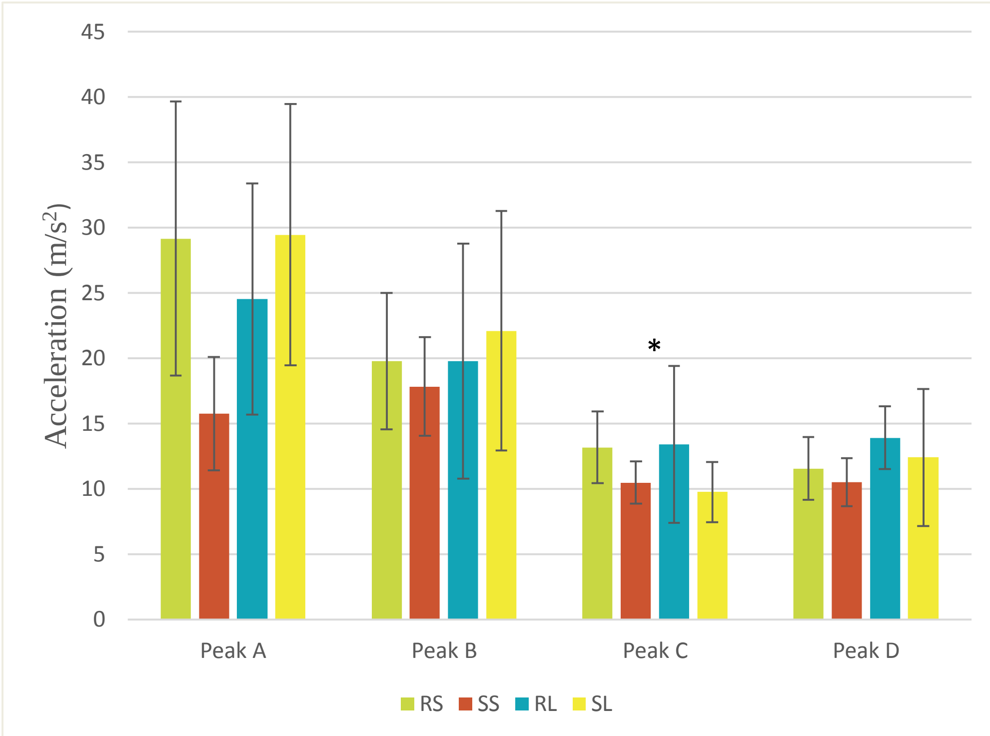

Figure 13. Group mean and standard deviation resultant un-weighted peak accelerations of the rigid small wheel (RS), suspended small wheel (SS), rigid large wheel (RL), and suspended large wheel (SL) conditions over the door threshold. Peak A= front wheel first contact with threshold, B= front wheel leaving the door threshold, C= rear wheel first contact with threshold, D= rear wheel leaving the door threshold. Significant differences denoted with *, $p \leq 0.05$

Door threshold mean frequency-weighted peak accelerations calculated for each wheelchair condition and wheel diameter traversing the obstacles are presented in Figure 14. The frequency-weighted peak accelerations C and D (when the rear wheel hits and leaves the door threshold) were significantly smaller with the use of suspension (p=0.049, p=0.001, respectively). For the small diameter wheels, the use of suspension resulted in a 17% and 25% decrease in weighted peak acceleration at peak C and peak D, respectively. For the large diameter wheels, the use of suspension resulted in a 10 % and 21% reduction in weighted peak acceleration for peak C and peak D, respectively. Again, the weighted peak accelerations A and

B (i.e. when the front wheel hits and leaves the door threshold) were not significantly different between the rigid and suspended chair (p=0.487, p=0.984).

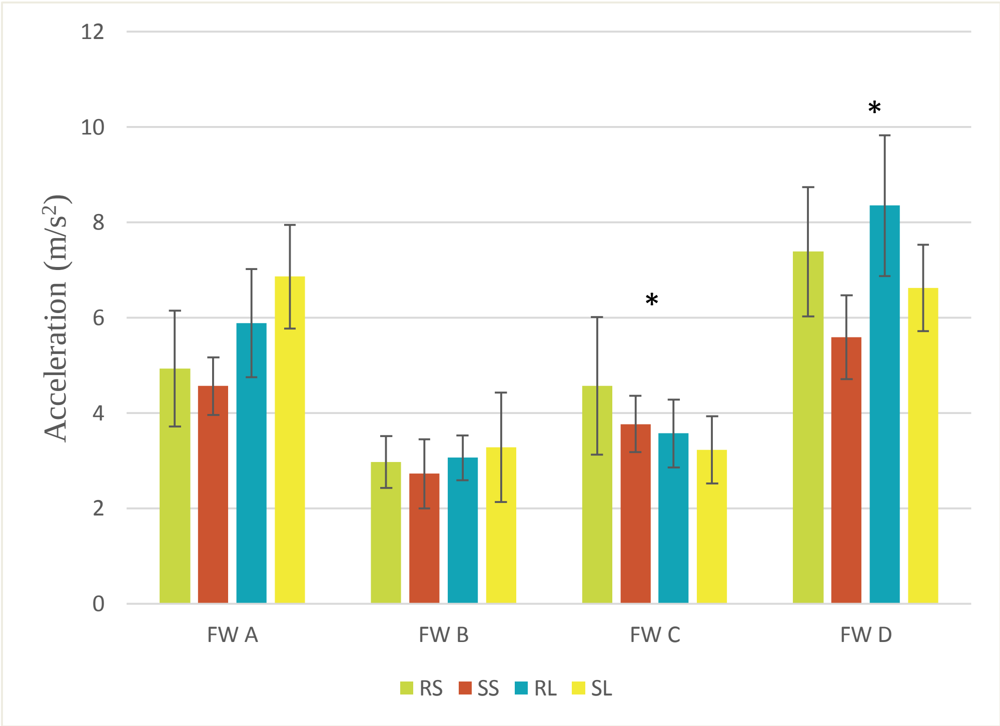

Figure 14. Group mean and standard deviation frequency-weighted (FW) resultant accelerations of the rigid small wheel (RS), suspended small wheel (SS), rigid large wheel (RL), and suspended large wheel (SL) conditions over the door threshold. FW A= front wheel first contact with threshold, FW B= front wheel leaving the door threshold, FW C= rear wheel first contact with threshold, FW D= rear wheel leaving the door threshold. Significant differences denoted with *. $p \leq 0.05$

The group mean and standard deviations for the un-weighted and FW peak accelerations over the 2 cm descent and ascent are in Figure 15 and Figure 16 respectively. There were significant decreases in un-weighted peak acceleration H (when the rear wheel was ascending 2 cm) with the use of suspension (p=0.001). There were not significant differences in un-weighted peak acceleration E, F, or G (p=0.263, p=0.693, p=0.533, respectively) during the 2 cm descent and ascent. There was a significant decrease in the weighted peak acceleration F with the use of suspension, but no significant differences in the weighted peak accelerations E, G, or H ($p=0.005$, $p= 0.133$, $p= 0.809$, $p= 0.086$, respectively).

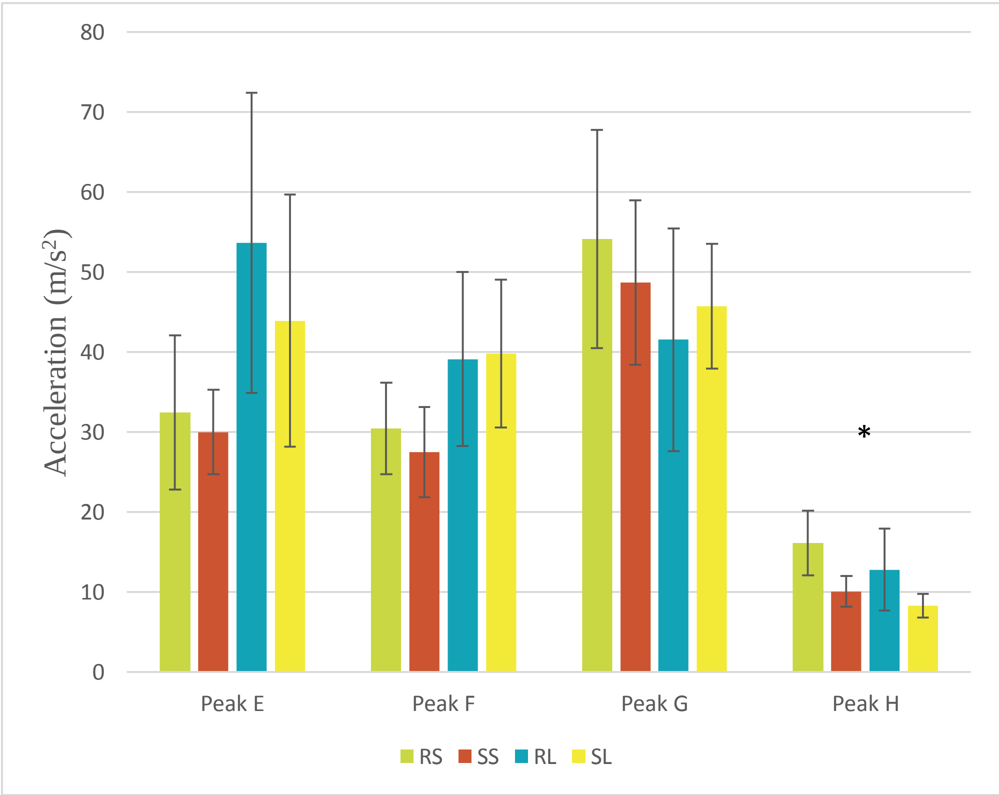

Figure 15. Group mean and standard deviations of un-weighted peak accelerations for the 2 cm descent and 2 cm ascent for rigid small wheel (RS), suspended small wheel (SS), rigid large wheel (RL), and suspended large wheel (SL). Peak E= when the front caster wheel drops off, Peak F= when the rear wheel drops off, Peak G= when the front caster wheel ascends, and Peak H= when the rear wheel ascends. Significant differences denoted with *. $p \leq 0.05$

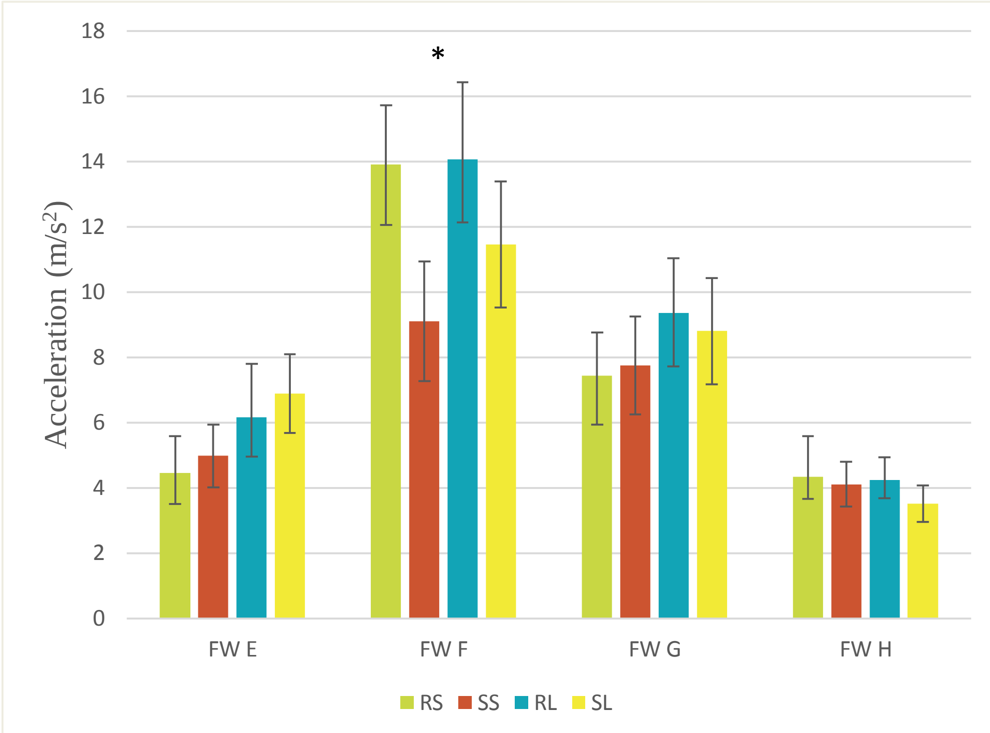

Figure 16. Group mean and standard deviations of the FW peak accelerations of the 2 cm descent and 2 cm ascent for rigid small wheel (RS), suspended small wheel (SS), rigid large wheel (RL), and suspended large wheel (SL). FW E= when the front caster wheel drops off, FW F= when the rear wheel drops off, FW G= when the front caster wheel ascends, and FW H= when the rear wheel ascends. Significant differences denoted with *. $p \leq 0.05$

RMS, VDV, and Total VDV

There were significant reductions in the weighted RMS over the truncated domes between the rigid and suspended wheelchair (p=0.011). Mean and standard deviations of frequency-weighted RMS of the resultant acceleration calculated for each wheelchair/wheel size are presented in Figure 17. The crest factors while traversing the truncated domes were not greater than 9, therefore, evaluating the vibration exposure with the RMS is sufficient. With the small and large diameter wheels, there was a 14% reduction when using suspension.

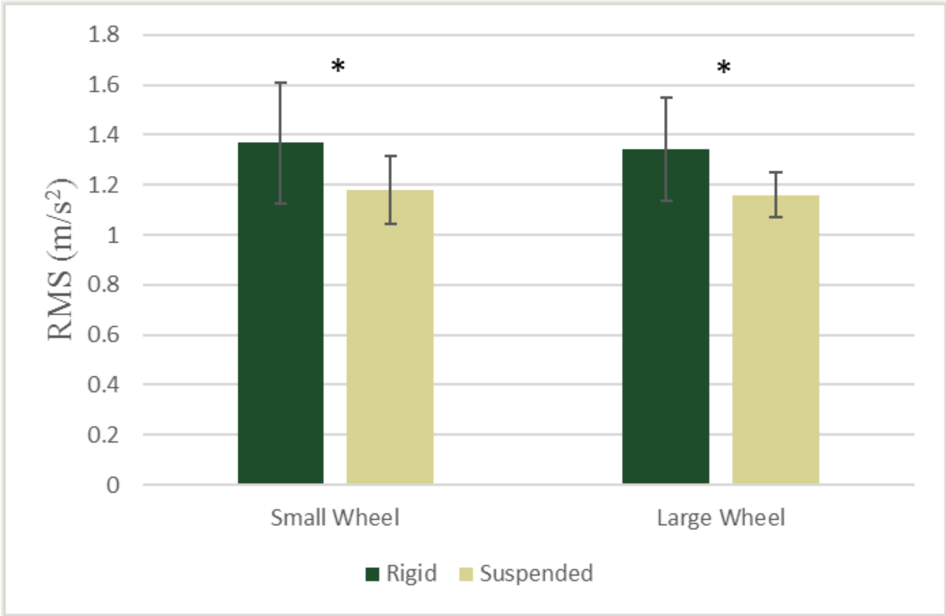

Figure 17. Group mean and standard deviation frequency-weighted RMS values of the resultant accelerations for each condition over the simulated truncated domes: rigid small wheel (RS), suspended small wheel (SS), rigid large wheel (RL), suspended large wheel (SL). Significant differences denoted with *. $p \leq 0.05$

The group mean VDV and total VDV values are presented in Table 2 and Figure 18, respectively. When vibration exposure consists of more periods, the total VDV (equation 11) can be calculated. There were significant decreases in VDV with suspension over the door threshold and 2 cm descent (p=0.041, p=0.016, respectively). There were not significant differences in VDV between rigid and suspended chairs over the truncated domes and 2 cm ascent (p=0.095, p=0.187, respectively). There were significant decreases in total VDV between the rigid and suspended chair (p=0.004). Using the small diameter wheels, the total VDV decreased 22%, and using the large diameter wheels, the total VDV decreased 10%. Table 3 provides the time to exceed the Health Guidance Caution Zone recommended by ISO 2631-1 in using total VDV. The time exceeding the upper boundary is greater when using suspension.

Table 2. Group mean and standard deviations of the VDV over each of the surfaces, and the Total VDV. Significant differences in VDV from the rigid wheelchair denoted with *. $p \leq 0.05$

	VDV ($m/s^{1.75}$)			
	Rigid Small	Suspended Small	Rigid Large	Suspended Large
Door Threshold	20.38 ± 3.51	16.56 ± 1.44 *	21.18 ± 2.97	20.24 ± 2.32 *
Truncated Domes	15.39 ± 2.62	13.47 ± 1.26	15.90 ± 2.97	13.36 ± 1.22
2 cm Descent	30.53 ± 4.13	22.31 ± 2.69 *	28.97 ± 5.92	25.65 ± 2.58 *
2 cm Ascent	20.48 ± 1.70	19.49 ± 2.15	24.33 ± 3.35	23.68 ± 2.39

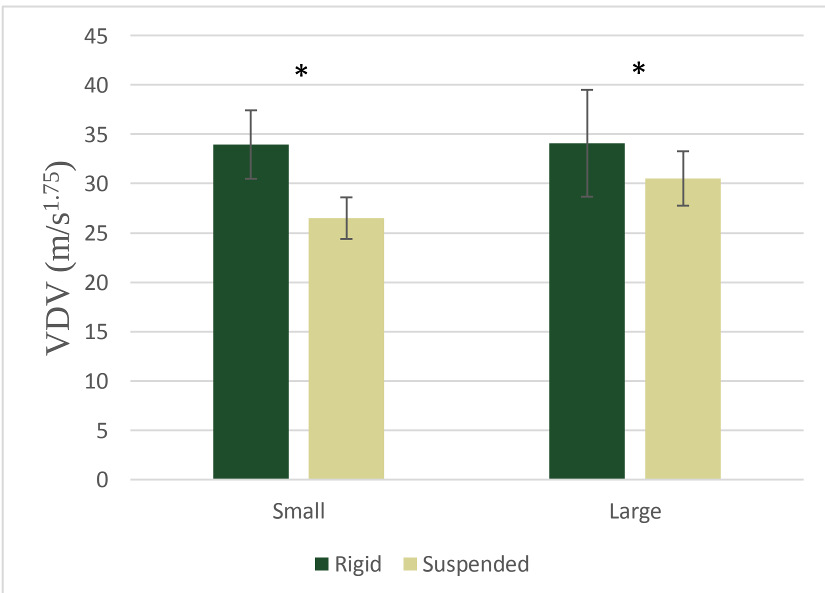

Figure 18. The group mean and standard deviations of VDV total of all surfaces (door threshold, truncated domes, 2 cm descent, and 2 cm ascent). Significant differences denoted with *. $p \leq 0.05$

Table 3. Time in minutes to cross the lower and upper boundaries of the Health Guidance Caution Zone for the rigid and suspended chairs with different wheel sizes.

	Lower Boundary Crossing (min)	Upper Boundary Crossing (min)
Rigid Small	2.07 ± 0.76	33.05 ± 12.13
Suspended Small	5.46 ± 2.37	87.37 ± 37.97
Rigid Large	2.38 ± 1.73	38.11 ± 27.71
Suspended Large	3.04 ± 1.17	48.68 ± 18.80

Total Power per Octave

The total power per octave group mean and standard deviations for the door threshold, truncated domes, 2 cm descent, and 2 cm ascent are presented in Table 4, Table 5, Table 6, and Table 7 respectively. There were significant increases in total power with the suspended wheelchair for the frequency ranges < 1.25, 49.61 – 99.21, 125- 157.49, and 198.43 – 500 Hz ($p \leq 0.033$) while traversing over the door threshold. There were significant increases in total power with the use of suspension in the frequency ranges < 1.25, 1.25 – 1.55, 62.5 – 198.43, 250- 315, and 315 – 500 Hz ($p \leq 0.024$) while traversing over the truncated domes. There were significant increases in total power with the use of suspension during the 2 cm descent in the frequency ranges < 1.25, 1.25 – 1.55, 1.55 - 1.95, 49.61 – 62.5, 62.5 – 78.75, 78.75 – 99.21, and 250 – 315 Hz ($p \leq 0.05$). There were significant decreases in total power with the use of suspension in the frequency ranges 15.63 – 19.69, 19.69 – 24.8 Hz ($p \leq 0.044$) while traversing the 2 cm descent. There significant increases in total power during the 2 cm ascent were in the frequency ranges 62.5 – 78.75, 125 – 157.49, and 250 – 315 Hz ($p \leq 0.05$). However, there were no differences within the typically identified hazardous region for seated humans from 4-12 Hz ($p \geq 0.086$).

Table 4. Total power per octave the door threshold. Significant differences denoted with *. $p \leq 0.05$, and indicate differences between the use of suspension.

Frequency (Hz)	Total Power m*Hz^{-1} *s^{-2}				
	Rigid Small	Suspended Small	Rigid Large	Suspended Large	p
< 1.25	6.44 ± 1.95	6.93 ± 1.73 *	6.18 ± 2.41	11.08 ± 5.86 *	0.012
1.25 – 1.55	0.29 ± 0.20	0.38 ± 0.24	0.37 ± 0.23	0.30 ± 0.16	0.466
1.55 – 1.95	0.26 ± 0.11	0.28 ± 0.09	0.25 ± 0.14	0.37 ± 0.35	0.27
1.95 – 2.46	0.38 ± 0.21	0.49 ± 0.23	0.45 ± 0.25	0.46 ± 0.25	0.539
2.46 – 4.92	2.92 ± 0.83	3.36 ± 0.95	2.58 ± 0.51	3.98 ± 3.06	0.24
4.92 – 6.20	0.62 ± 0.37	0.75 ± 0.42	0.52 ± 0.41	0.92 ± 0.73	0.097
6.20 – 7.81	0.72 ± 0.35	0.85 ± 0.37	0.52 ± 0.33	0.43 ± 0.11	0.298
7.81 – 9.84	0.62 ± 0.37	0.77 ± 0.43	0.39 ± 0.16	0.84 ± 0.40	0.327
9.84 – 12.4	0.65 ± 0.34	0.53 ± 0.32	0.37 ± 0.16	0.94 ± 0.49	0.633
12.4 – 15.63	0.68 ± 0.35	0.47 ± 0.33	0.47 ± 0.17	0.75 ± 0.43	0.584
15.63 – 19.69	0.80 ± 0.29	0.64 ± 0.46	0.57 ± 0.21	0.96 ± 0.36	0.788
19.69 - 24.8	1.15 ± 0.39	0.74 ± 0.49	0.65 ± 0.40	0.97 ± 0.41	0.339
24.8 – 31.25	1.05 ± 0.55	0.82 ± 0.36	0.90 ± 0.46	1.12 ± 0.66	0.947
31.25 – 39.37	1.15 ± 0.39	1.00 ± 0.52	1.00 ± 0.35	1.20 ± 0.58	0.84
39.37 – 49.61	1.39 ± 0.24	1.21 ± 0.52	0.93 ± 0.52	2.11 ± 1.30	0.078
49.61 – 62.5	1.63 ± 0.81	1.82 ± 0.83 *	1.28 ± 0.64	2.75 ± 0.73 *	0.001
62.5 – 78.75	1.30 ± 0.29	3.17 ± 1.96 *	1.26 ± 0.36	4.90 ± 2.96 *	0.007
78.75 – 99.21	1.83 ± 0.74	2.12 ± 1.22 *	1.28 ± 0.57	2.95 ± 1.94 *	0.029
99.21 – 125	2.28 ± 1.13	2.30 ± 1.20	1.81 ± 0.87	3.11 ± 1.20	0.103
125 – 157.49	2.40 ± 0.95	4.57 ± 3.20 *	2.15 ± 0.87	6.32 ± 3.58 *	0.025
157.49 – 198.4	2.95 ± 0.62	2.10 ± 0.89	2.56 ± 1.20	7.33 ± 4.78	0.066
198.4 – 250	4.05 ± 1.36	4.46 ± 3.62 *	4.27 ± 2.52	7.47 ± 2.92 *	0.033
250 – 315	6.62 ± 3.31	7.46 ± 3.25 *	7.35 ± 3.88	14.06 ± 7.09 *	0.03
315 – 500	39.41 ± 15.1	47.79 ± 22.37 *	30.43 ± 13.73	84.81 ± 33.89 *	0.003

Table 5. Total power per octave the door threshold. Significant differences denoted with *. p ≤ 0.05, and indicate differences between the use of suspension.

Frequency (Hz)	Total Power m*Hz^{-1} *s^{-2}				
	Rigid Small	Suspended Small	Rigid Large	Suspended Large	p
< 1.25	10.56 ± 3.42	13.29 ± 5.89 *	7.27 ± 3.51	10.63 ± 2.39 *	0.001
1.25 – 1.55	0.07 ± 0.04	0.08 ± 0.04 *	0.07 ± 0.04	0.15 ± 0.06 *	0.008
1.55 – 1.95	0.07 ± 0.04	0.07 ± 0.04	0.08 ± 0.05	0.10 ± 0.06	0.218
1.95 – 2.46	0.07 ± 0.04	0.07 ± 0.05	0.11 ± 0.07	0.10 ± 0.07	0.692
2.46 – 4.92	0.57 ± 0.15	0.71 ± 0.34	0.64 ± 0.34	0.91 ± 0.50	0.072
4.92 – 6.20	0.13 ± 0.05	0.18 ± 0.14	0.12 ± 0.04	0.20 ± 0.14	0.331
6.20 – 7.81	0.13 ± 0.05	0.32 ± 0.21	0.22 ± 0.11	0.20 ± 0.16	0.212
7.81 – 9.84	0.16 ± 0.06	0.45 ± 0.28	0.30 ± 0.19	0.25 ± 0.11	0.095
9.84 – 12.4	0.51 ± 0.15	0.58 ± 0.27	0.52 ± 0.14	0.72 ± 0.19	0.936
12.4 – 15.63	2.07 ± 1.44	1.25 ± 0.87	1.58 ± 1.16	1.55 ± 0.92	0.29
15.63 – 19.69	0.42 ± 0.17	0.40 ± 0.23	0.49 ± 0.28	0.47 ± 0.32	0.329
19.69 - 24.8	0.69 ± 0.15	0.58 ± 0.33	0.66 ± 0.32	0.42 ± 0.17	0.268
24.8 – 31.25	1.84 ± 0.86	1.57 ± 0.85	1.05 ± 0.61	1.03 ± 0.48	0.401
31.25 – 39.37	1.21 ± 0.50	1.39 ± 0.65	0.96 ± 0.39	0.82 ± 0.19	0.846
39.37 – 49.61	2.11 ± 0.87	2.75 ± 1.15	1.57 ± 0.72	1.60 ± 0.55	0.112
49.61 – 62.5	2.70 ± 0.82	3.42 ± 1.93	2.25 ± 1.15	2.88 ± 0.79	0.144
62.5 – 78.75	2.65 ± 0.55	4.62 ± 1.94 *	2.84 ± 1.34	5.09 ± 0.73 *	0.003
78.75 – 99.21	3.28 ± 1.85	4.89 ± 2.17 *	3.39 ± 1.80	5.37 ± 1.62 *	0.002
99.21 – 125	3.85 ± 2.04	6.72 ± 2.99 *	3.30 ± 1.16	9.24 ± 4.13 *	0.003
125 – 157.49	3.70 ± 1.60	11.61 ± 7.56 *	3.68 ± 1.97	12.18 ± 4.60 *	0.003
157.49 – 198.4	3.92 ± 1.64	5.24 ± 3.00 *	3.12 ± 1.14	6.39 ± 4.29 *	0.024
198.4 – 250	4.43 ± 1.36	5.27 ± 2.74	4.25 ± 1.79	6.56 ± 3.28	0.115
250 – 315	6.41 ± 3.15	10.28 ± 5.28 *	6.12 ± 1.74	11.49 ± 3.96 *	0.005
315 – 500	51.9 ± 16.0	65.74 ± 23.79 *	38.62 ± 14.07	63.13 ± 25.11 *	0.006

Table 6. Total power per octave the 2 cm descent. Significant differences denoted with *. p ≤ 0.05, and indicate differences between the use of suspension.

Frequency (Hz)	Total Power m^*Hz^{-1} $^*s^{-2}$				
	Rigid Small	Suspended Small	Rigid Large	Suspended Large	p
< 1.25	6.16 ± 1.36	7.80 ± 1.37 *	6.78 ± 1.32	11.30 ± 2.94 *	0.009
1.25 – 1.55	0.38 ± 0.11	0.53 ± 0.13 *	0.50 ± 0.08	0.97 ± 0.26 *	0.003
1.55 – 1.95	0.67 ± 0.14	0.79 ± 0.12 *	0.81 ± 0.17	1.42 ± 0.37 *	0.011
1.95 – 2.46	0.83 ± 0.18	0.96 ± 0.25	1.22 ± 0.48	1.71 ± 0.51	0.097
2.46 – 4.92	5.01 ± 1.51	5.15 ± 1.89	5.69 ± 2.31	9.87 ± 3.35	0.102
4.92 – 6.20	0.72 ± 0.35	0.69 ± 0.27	1.13 ± 0.76	1.52 ± 0.67	0.178
6.20 – 7.81	0.65 ± 0.22	0.65 ± 0.24	1.18 ± 0.48	1.79 ± 0.78	0.296
7.81 – 9.84	0.57 ± 0.25	0.53 ± 0.23	1.14 ± 0.69	1.42 ± 0.72	0.494
9.84 – 12.4	0.69 ± 0.25	0.49 ± 0.17	1.11 ± 0.48	1.15 ± 0.61	0.29
12.4 – 15.63	0.96 ± 0.31	0.51 ± 0.25	1.47 ± 1.08	1.18 ± 0.64	0.138
15.63 – 19.69	1.09 ± 0.49	0.53 ± 0.22 *	1.61 ± 0.81	1.04 ± 0.79 *	0.044
19.69 - 24.8	1.12 ± 0.54	0.73 ± 0.34 *	1.66 ± 0.75	0.80 ± 0.43 *	0.012
24.8 – 31.25	1.23 ± 0.65	0.84 ± 0.36	1.35 ± 0.34	1.76 ± 0.98	0.623
31.25 – 39.37	1.35 ± 0.56	0.98 ± 0.51	2.03 ± 1.02	2.21 ± 1.16	0.801
39.37 – 49.61	1.02 ± 0.30	1.29 ± 0.40	1.92 ± 0.63	2.62 ± 1.62	0.142
49.61 – 62.5	1.09 ± 0.62	2.02 ± 0.68 *	1.91 ± 1.15	2.95 ± 1.66 *	0.018
62.5 – 78.75	0.93 ± 0.28	3.83 ± 1.41 *	2.01 ± 0.94	3.78 ± 0.90 *	0.0001
78.75 – 99.21	1.16 ± 0.70	$3.45 \pm .45$ *	2.55 ± 1.43	4.79 ± 1.96 *	0.01
99.21 – 125	1.56 ± 0.66	2.17 ± 0.68	3.03 ± 1.82	3.40 ± 0.89	0.623
125 – 157.49	1.59 ± 0.75	2.98 ± 1.49	4.19 ± 3.07	6.07 ± 3.98	0.141
157.49 – 198.4	1.83 ± 0.91	2.71 ± 0.89	5.89 ± 4.11	4.72 ± 1.01	0.553
198.4 – 250	3.73 ± 1.54	5.40 ± 1.43	8.89 ± 7.72	9.96 ± 4.43	0.522
250 – 315	6.81 ± 2.57	8.82 ± 2.77 *	9.35 ± 3.51	16.80 ± 5.55 *	0.018
315 – 500	36.7 ± 11.3	53.91 ± 14.38	70.08 ± 33.4	95.81 ± 38.87	0.11

Table 7. Total power per octave the 2 cm ascent. Significant differences denoted with *. p ≤ 0.05, and indicate differences between the use of suspension.

Frequency (Hz)	Total Power m*Hz^{-1} *s^{-2}				
	Rigid Small	Suspended Small	Rigid Large	Suspended Large	p
< 1.25	9.94 ± 1.39	14.38 ± 4.65	14.86 ± 6.59	21.17 ± 13.75	0.1
1.25 – 1.55	0.92 ± 0.18	1.40 ± 0.55	1.63 ± 0.89	1.75 ± 1.02	0.29
1.55 – 1.95	1.06 ± 0.38	1.37 ± 0.72	2.11 ± 1.40	2.44 ± 1.62	0.514
1.95 – 2.46	0.85 ± 0.25	1.78 ± 0.92	1.34 ± 0.64	1.54 ± 1.13	0.077
2.46 – 4.92	4.23 ± 2.28	5.19 ± 2.49	5.56 ± 3.36	9.66 ± 8.27	0.086
4.92 – 6.20	0.83 ± 0.27	1.05 ± 0.51	0.71 ± 0.27	1.21 ± 0.77	0.095
6.20 – 7.81	0.93 ± 0.48	1.01 ± 0.60	0.82 ± 0.27	0.58 ± 0.27	0.561
7.81 – 9.84	1.12 ± 0.40	1.06 ± 0.42	0.70 ± 0.32	1.31 ± 0.66	0.131
9.84 – 12.4	1.40 ± 0.62	2.14 ± 1.60	1.42 ± 1.07	2.43 ± 1.72	0.15
12.4 – 15.63	1.33 ± 0.76	1.61 ± 1.05	1.51 ± 1.12	1.75 ± 0.78	0.366
15.63 – 19.69	1.29 ± 0.31	1.14 ± 0.74	1.71 ± 0.93	1.26 ± 0.47	0.351
19.69 - 24.8	1.93 ± 0.97	1.04 ± 0.58	1.27 ± 0.76	1.05 ± 0.47	0.857
24.8 – 31.25	1.65 ± 0.75	0.96 ± 0.20	1.08 ± 0.41	1.52 ± 1.24	0.255
31.25 – 39.37	1.60 ± 0.72	1.06 ± 0.55	1.16 ± 0.30	1.36 ± 0.83	0.466
39.37 – 49.61	2.26 ± 1.23	1.26 ± 0.61 *	2.60 ± 1.64	1.55 ± 1.20 *	0.027
49.61 – 62.5	2.19 ± 0.64	2.35 ± 0.64	2.28 ± 1.12	2.13 ± 0.74	0.492
62.5 – 78.75	3.10 ± 1.95	4.80 ± 2.24 *	3.17 ± 1.65	7.36 ± 5.81 *	0.049
78.75 – 99.21	2.70 ± 1.65	3.81 ± 1.18	3.03 ± 1.14	5.06 ± 4.21	0.238
99.21 – 125	3.47 ± 1.14	5.43 ± 2.15	4.04 ± 1.96	9.20 ± 7.61	0.059
125 – 157.49	3.76 ± 1.09	18.46 ± 18.60	5.46 ± 2.88	13.94 ± 12.83	0.036
157.49 – 198.4	4.69 ± 2.21	6.21 ± 2.27	6.17 ± 5.02	8.36 ± 6.55	0.268
198.4 – 250	6.78 ± 3.00	10.18 ± 5.75	10.98 ± 7.02	14.92 ± 13.08	0.128
250 – 315	11.57 ± 2.25	20.81 ± 7.22 *	12.72 ± 6.83	27.49 ± 21.32 *	0.038
315 – 500	52.10 ± 9.13	87.37 ± 27.06	89.25 ± 51.48	127.87 ± 67.11	0.053

Chapter V: DISCUSSION

Wheelchair users spend extended periods of time in their chair, and experience shock and vibrations exceeding the ISO 2631-1 recommendations for health (VanSickle, 2001; Garcia-Mendenz, 2013). According to the ISO 2631-1 standards, without an eight-hour recovery period between shock and vibrations exposure, the effect of the exposure is cumulative. Therefore, the longer the shock and vibration exposure time, the lower boundary, where health risks begin, decreases. Reductions in shock and vibration exposure could lead to increased comfort, decreases in muscle fatigue, and decreases in back and neck pain, although, there is no conclusive evidence to the percent reduction needed for a decrease in health risks. Therefore, it is recommended to reduce shock and vibration exposure as much as possible. Rear wheel suspension systems exist in a variety of different designs (elastomers, springs, spring/dampers). However, prior to QuadshoX, manual tilt-in-space wheelchairs did not have a rear suspension system option available. The purpose of the present study was to evaluate the impact of the QuadshoX spring- damper suspension kit on shock and vibrations over 4 different obstacles. The general hypothesis was the QuadshoX after-market suspension kit would decrease shock and vibration exposure compared to its rigid manufactured design. More specifically, the suspension would decrease un-weighted and weighted peak accelerations at both the front caster and rear wheel traversing the door threshold, 2 cm descent, and 2 cm ascent. Additionally, the weighted RMS of the resultant acceleration traversing truncated domes would decrease, and the VDV as well as total VDV would decrease with the use of suspension. From a frequency perspective, it was hypothesized the total power per octave would decrease with the use of suspension in frequencies most harmful for seated humans. Finally, a gap in the literature exists investigating the effect of wheel diameter on shock and vibration suppression. Therefore, the final hypothesis was a larger wheel diameter (solid tires) would decrease shock and vibration exposure compared to a smaller wheel.

Peak Accelerations

The hypothesis of a decrease in un-weighted and weighted peak accelerations at the rear wheel is partially accepted. The door threshold and 2 cm decent had significant decreases in weighted peak acceleration at the rear wheel with the use of suspension. The un-weighted peak acceleration C (when the rear wheel first impacted the door threshold) was significantly decreased with the use of suspension. The un-weighted peak acceleration at the rear wheel during the 2 cm descent was not significantly different with the use of suspension. The opposite was found during the 2 cm ascent, as there were significant decreases in un-weighted peak acceleration at the rear wheel with the use of suspension, but no significant decreases in weighted peak acceleration.

The ISO 2631-1 frequency-weightings put more emphasis on the frequencies most harmful to the human body; therefore, it is proposed that they are more useful in concluding the injury risk associated experiencing high magnitude shocks when traversing an obstacle (Kwarciak, 2003). When examining the weighted accelerations for the rigid wheelchair, the largest magnitude accelerations occurred at the rear wheel, not the front caster wheel. With the use of suspension, the largest weighted accelerations at the rear wheel (FW C and FW D) significantly decreased by 10-25% when traversing the door threshold. Cooper et al. (2003) examined just *un-weighted* peak accelerations, and found no significant differences between rear wheel suspension and no suspension. Somewhat similar results were found in the present study when examining the *un-weighted* peak accelerations (Peak D). But un-weighted peak C was significantly reduced with the use of suspension while traversing the door threshold. Both un-weighted peaks at the rear wheel while traversing the door threshold are of similar magnitude. . This potentially indicates collecting single peak accelerations, regardless of which wheel and the point the wheel is during the traverse, may also hide some of the benefits of rear wheel suspension as there are multiple high magnitude shocks when traversing obstacles. Although different from the methods in the present study, Requejo et al. (2008) investigated shock and vibration exposure of self-propelled manual wheelchairs using a technique looking at seat forces and head accelerations. They had subjects set up on treadmill with a vibration simulator to apply small repeated bumps. For one of the trials, the subjects selected their own speed. The subjects selected faster speeds when using a suspended chair suggesting the suspension provided a smoother ride and increased user comfort, and they also found suspension reduced the force and accelerations experienced by the user. Tilt-in-space wheelchair users typically have an attendant pushing their chair; therefore, do not have full control of their speed. This is especially true in non-verbal tilt-in-space users. Additionally, Requejo et al. (2008) compared subjects with lower level spinal cord injuries to subjects with higher-level spinal cord injuries, and found those with higher spinal cord injuries and less postural control saw greater reductions in accelerations and seat forces than those with lower spinal cord injuries. For individuals with less postural control, having the QuadshoX suspension reducing accelerations at the rear wheel may improve comfort as well as decrease health risk through minimizing accelerations and forces experienced by the rider. However, as previously indicated, more research is needed to determine the mechanisms contributing to discomfort, fatigue, and secondary injuries in this population.

Similar to the door threshold, the largest magnitude weighted peak accelerations during the 2 cm decent occurred at the rear wheel. The rear wheel suspension significantly reduced weighted peak accelerations (FW F) at the rear wheel from 10- 25%. Kwarciak et al. (2003) reported weighted peak accelerations comparable to those found in the present study. In their study, the peak accelerations collected were solely at the rear wheel during a “wheelie” curb decent. There were no significant differences in un-weighted peak accelerations (Peak E and F) during the 2 cm descent. There appears to be high variability in the accelerations at both the front caster and rear wheel which could have contributed to the lack of significance reported.

To our knowledge, no research has explored a 2 cm ascent when investigating the effectiveness of front caster or rear wheel suspension. However, we felt it was an important dimension for an abrupt rise and/or fall similar to what one might encounter on older sidewalks where the concrete has fractured and shifted. There were significant decreases in un-weighted peak acceleration at the rear wheel, (Peak H), when ascending the obstacle from 35 -37%, varying only slightly with wheel diameter. However, there were no significant decreases in the frequency-weighted peak accelerations at the rear wheel (FW H) when ascending the obstacle. Although the FW accelerations were not significantly different during the ascent, the un-weighted peak acceleration at the rear wheel did decreases, which could potentially improve the comfort of the users when traversing uneven ground.

We found no significant differences in un-weighted and weighted peak accelerations at the front caster wheel over any of the obstacles; therefore, we reject this hypothesis. There was high variability in the un-weighted and weighted peak accelerations experienced at the front caster wheel contributing to the lack of significance reported. One potential reason for the variability could be inconsistencies in the speed at which the obstacle was first hit. The time of greatest concern was the time spent over the obstacle, but not as much emphasis was put on the time spent going from rest to desired speed, which was typically just a couple of steps. The variability in speed could have resulted from the direction the front caster wheels at the start of motion. If the caster wheels were somehow near or actually perpendicular to the chair, this could impact the force used to accelerate the wheelchair, and potentially vary the time used accelerating to the obstacle.

Although large magnitude shocks occur at the front caster wheel, comparable or greater magnitude shocks occur at the rear wheel. This was opposite of what was reported in a study evaluating front caster wheel suspension. These investigators found the front caster wheel to experience much greater peak accelerations than the rear wheel (Gregg, 1998). Their subjects were traveling at 1.53 and 2 m/s when traversing a treadmill with repeated 2 mm bumps and ramp with a 3 cm drop at the end respectively. In the present study, the rear wheels sometimes exhibited much greater peak accelerations than the front caster wheels, and the subjects were traveling approximately less than 1 m/s when traversing the obstacles. In the Gregg et al. 1998 study, the subjects first hit the obstacles at higher speeds than the subjects in the present study. The higher speeds would result in higher peak accelerations at the front caster wheel. Additionally, the present study had large variability in the peak accelerations at the front caster wheel, which is potentially a result of differences in the initial speed the front caster wheel hit the obstacle. A ramp with a 3 cm drop was not investigated in the present study, but we speculate the results in peak acceleration at the front caster wheel would still be highly variable with a tilt-in-space wheelchair. Tilt-in-space wheelchairs are typically pushed by a caregiver. Therefore, when rolling down a ramp there could potentially be some resistance to the increasing speeds due to the caregiver (i.e. the caregivers will reduce the acceleration while self-propelled users are more likely to “fall” down the slope).

Cooper et al. (2003) found no significant differences in peak accelerations between no suspension and rear wheel suspension, but did not separate the peak accelerations into events happening at the front versus rear wheels, which could have masked any differences because occasionally front caster wheels have higher accelerations and at other times the rear wheels. Additionally, Cooper et al. (2003) had placed all tested rear suspension types (elastomer, spring/damper) into a single group, therefore, the benefits of a particular type of rear wheel suspension was not directly examined. Kwarciak et al. (2003) and Requejo et al. (2008) found spring-type suspension wheelchairs had the lowest accelerations and seat forces when compared to a rigid or polymer-based rear suspension. Therefore, grouping all suspension types together when assessing the shock and vibration suppressing abilities, as the research by Cooper et al. (2003) and Garcia-Mendez et al. (2013) did, may minimize the impact of a particular suspension configuration.

Prior to the present study, limited research has separated peak accelerations into categories based on where the front caster or rear wheel was when traversing over different obstacles. Studies that have isolated the impacts to solely at the rear wheel have found decreases in shock and vibration exposure with the use of rear wheel suspension (Kwarciak, 2003; Requejo, 2008; Requejo, 2009). It is advantageous to separate the peaks into individual events to understand where the changes in accelerations happen due to the use of suspension. Through qualitative measures, a study found vibrations from the wheelchair affect the comfort of the rider at the neck, lower back, and buttocks (Maeda, 2003). The areas with the highest levels of discomfort are in vertically in line with the rear wheel, therefore seeing a reduction in the weighted peak accelerations at the rear wheel may improve rider comfort, as well as decrease health risks, when traversing a door threshold, 2 cm descent, and 2 cm ascent.

Proper seating for wheelchair users should place all joints, (hip, knee, elbow), at or near the individual's neutral position or 90 degrees (Cooper, R.A., 1995). Poor posture, hips, knees, and elbows not near neutral position or 90 degrees, and improper support causes spinal deformities, abnormal pelvic tilts, and decreases in range of motion potentially causing discomfort and pain (Cooper, R.A., 1995). Tilt-in-space wheelchair users have less postural control than someone without a mobility impairment, which results in bumps, small and large, slowly displacing them from an optimal posture. Additionally, they do not always have the ability to readjust themselves when displaced from their optimal seating position, nor is it easy for a caregiver to see if they have been displaced when pushing from behind; therefore, minimizing the accelerations of the chair could assist in maintaining proper posture when exposed to high magnitude shocks. This could potentially decrease the user's risk of the pain and discomfort associated with improper postures.

RMS, VDV, and Total VDV

With the use of rear wheel suspension, there was a statistically significant 14% reduction in weighted RMS over the truncated domes; therefore, we accept this hypothesis. The crest factor for the obstacle was not greater than 9, thus, according to the ISO 2631-1 standards, using weighted RMS for vibration analysis is appropriate. The weighted RMS values reported in the present study are slightly higher than reported in studies by Garcia-Mendez et al. (2013) and Wolf et al. (2007). The present study had subjects propelled over obstacles creating high magnitude shock exposure; whereas Wolf et al. (2007) had subjects riding over different sidewalk surfaces that were relatively level in comparison. Furthermore, Garcia-Mendez et al. (2013) collected accelerations over a two-week period, and because RMS is time dependent, some of the shock exposure during that time could have been lost.

VDV is more sensitive to peaks than weighted RMS, and is calculated through the 4^{th} power rather than the 2nd power as in RMS:

$$\mathrm{VDV} = \left\{ \int_0^{\mathrm{T}} \left[\mathrm{a}_{\mathrm{w(t)}} \right]^4 \mathrm{dt} \right\}^{\frac{1}{4}}$$

The present study found similar magnitude VDV as reported within other studies (Kwarciak, 2003; Garcia-Mendez, 2013). When looking at the VDV over each obstacle separately, VDV significantly decreased over the door threshold and 2 cm descent with the use of suspension. The present study found significant differences in the RMS over the truncated domes, but did not find significant differences in the VDV over the truncated domes. Similar to the present study, Kwarciak et al. (2003) had wheelchairs with and without suspension (a combination of front caster and rear wheel suspension) traverse over truncated domes. They only analyzed the VDV, (whereas we had looked at RMS and VDV over the truncated domes), and found significant differences with the use of suspension. The lowest VDV reported in their study was the wheelchair with an A-arm suspension configuration at the rear wheel compared to a single spring-damper and elastomer suspension (Kwarciak, 2003).

Altogether, there were four different obstacles traversed commonly encountered daily by someone who utilizes a wheelchair as their main means of transportation, and each obstacle accounted for short instances of shock and vibration exposure. Therefore, we also chose to compute the total VDV. With the use of suspension, the total VDV decreased 10% with the use large diameter wheels, and 22% with the use of smaller diameter wheels. Garcia-Mendez et al. (2013) found no significant differences in weighted RMS or VDV between using a rigid chair and a suspended chair during a two week period, but only had eight subjects out of 37 who used suspension (three with rear suspension of different types, and five with suspension at the front caster wheels). In the present study only comparing a rigid chair to a suspended chair, we saw reductions in vibration exposure that could potentially lead to decreases in health risks such as neck pain, back pain, muscle fatigue, and discomfort.

Total Power per Octave

The frequencies at the natural frequency of seated humans, 4 - 12 Hz, are amplified when experiencing shock and vibration which can cause muscle fatigue and damage to organs (Cooper, 2003). The goal of suspension systems is to either reduce the amplitude (i.e. power), or to shift the frequency so it is outside of the ranges causing the most risk (Cooper, 2003). There were no significant changes in the total power per octave in the 4-12 Hz range when comparing the rigid to the suspended chair; therefore, this hypothesis was rejected.

The seat vibration power found for the frequency octaves from 1.25 to 157.49 Hz were comparable to those found in the study by Cooper et al. (2003). The frequencies greater than 157.49 Hz were greater than reported in the study performed by Cooper et al. (2003). However, the frequencies greater than 150 Hz are far greater than the frequencies of interest in human vibration exposure as these frequencies are not the resonant frequencies of organs and tissues (Cooper, 2003). One possible explanation for the large discrepancy is Cooper et al. (2003) collected acceleration data at 1000 Hz, whereas the present study collected data at 2000 Hz. As such, the higher sampling rate will more faithfully measure higher frequencies. Cooper et al. (2003) found using rear wheel suspension, (a combination of spring/damper and elastomer suspension types), resulted in significant decreases in vibration power at octaves between 7.81 and 9.84 Hz and increases in the 12.4 – 15.63 Hz octave. They concluded although the vibration seemed to shift to a higher frequency, the higher octave was still in the natural frequency range of humans. As a result, they were not convinced frequency content was altered enough to reduce injury risk.

In the lowest octaves (less than 1.25 and 1.25 – 1.95 Hz), we found some significant increases in total power with the use of rear wheel suspension while traversing the door threshold, truncated domes, and 2 cm descent and ascent. The ISO 2631-1 document does not put as much emphasis on the weightings less than 1.25 Hz, as they deem those frequencies negligible when assessing the health risks of vibration exposure in the vertical direction (International Organization for Standardization (ISO), 1997c). The significant increases found in the lowest octaves do not pose nearly as great of a risk as do the frequencies from 4 – 12 Hz.

Significant differences to note during the 2 cm descent were at the ranges of 15.63 – 19.69 and 19.69 – 24.8 Hz. At these octaves, the rigid chair had significantly greater total power than the suspended chair. There was a range of higher octaves with significant increases in total power with the use of rear wheel suspension. However, the frequencies of interest when evaluating vibration exposure for humans do no exceed 50 Hz (Cooper, 2003). We conclude, similarly as Cooper et al. (2003), limited changes in total power with the use of the QuadshoX rear wheel suspension.

Limitations

One major limitation in the present study was the subjects did not use tilt-in-space wheelchairs. Because all the subjects had no limitations with postural control, head accelerations were not collected and evaluated. Those with less postural control could have greater levels of discomfort and increased health risks with shock and vibration exposure. They may have also responded differently to the impacts. Subjects were told to stay relaxed. However, they could not suppress reflex responses. Similar to most of the research, non-wheelchair users were utilized because wheelchair users are much more challenging to recruit and study in large numbers. Furthermore, it is undesirable to expose wheelchair users to additional shock and vibration.

Overall, we had collected data on a relatively small sample size in this exploratory project, though p values of non-significant findings were typically far from 0.05 and would not change unless a large number of subjects were added. Additionally, the rigid chair data were collected separately from the suspension data. Even though the same trained investigator pushed the chair for the rigid and suspended conditions, there was a two week period in between the time the conditions were collected and the investigator did not push wheelchairs during this time. Therefore, the time between collection periods for the rigid and suspended conditions could have potentially changed how the investigator approached and traversed the obstacles. Also, there is a notable difference between the rigid and suspended chairs in how it feels to push someone over the obstacles. This makes it difficult for a person to push the wheelchair over the obstacle consistently between the rigid and suspended chairs in a comparable manner. Similarly, there is a different feel to pushing someone in a wheelchair with larger rear wheels compared to a smaller rear wheel, which is most likely why we were not able to match speeds over the obstacles close enough to perform this comparison. A treadmill designed obstacle system may be better than manually pushing a wheelchair over obstacles to account for variability in speeds.

Another limitation were the obstacles chosen. We replicated obstacles often experienced by wheelchair users within the lab. These obstacles are a small fraction of the shock inducing obstacles a person in a wheelchair might traverse on a daily basis. Furthermore, the time of exposure was much less than the 4 to 8 hours in which the ISO 2631-1 Health Guidance Caution Zone was derived. In the future, longer duration studies, as performed by Garcia-Mendez et al., 2013, could be more representative of the impact QuadshoX rear wheel suspension kit has on tilt-in-space wheelchair shock and vibration suppression. Additionally, other literature has proposed the ISO 2631-1 does not adequately represent the vibration exposure in individuals who use wheelchairs as they have different mass distribution and trunk control than subjects in an occupational setting as the ISO 2631-1 standards were derived.

The research finding wheelchair users experience shock and vibrations exceeding the ISO 2631-1 standards have used manual self-propelled wheelchair users. It is unknown whether or not tilt-in-space wheelchair users are as active as individuals who are not pushed by an attendant or caregiver. Again, a longer duration study would reveal if the vibration exposure of tilt-in-space wheelchair users is comparable to the self-propelled manual wheelchair users previously reported on.

The time spent over each obstacle was significantly different between the small diameter and large diameter wheels. More specifically, less time was spent over the obstacles with the larger diameter wheels than the smaller diameter wheels. Larger wheels are much easier to keep up at a set speed than smaller diameter wheels, and because of this, it may have been much easier to traverse the obstacle at a higher speed with the larger wheel. The hypothesis of a larger wheel diameter having better shock and vibration suppressing ability is inconclusive. Future research should control for time discrepancies. We also were limited to only one trial for analysis rather than a three trial average as initially planned.

Lastly, research claims decreasing shocks and vibrations improves rider comfort, but rider comfort was not qualitatively assessed in the present study because the subjects did not utilize tilt-in-space wheelchairs on a daily basis. However, anecdotally, most subjects freely commented on a much more comfortable ride in the suspended chair. Future work should not only include shock and vibration exposure evaluations, but should also include an assessment of comfort of people who use tilt-in-space wheelchairs with and without suspension. Very limited research assesses the comfort levels of those who utilize rear wheel suspension.

Conclusions

This study found that the QuadshoX after-market rear wheel suspension provides some level of shock and vibration suppression at the rear wheel, but little change in frequency content. To our knowledge, no other research has investigated shock and vibration exposure when using tilt-in-space wheelchairs as all previous literature focuses on self-propelled manual wheelchairs or motorized power wheelchairs. Additionally, this is the first research investigating the newly available rear wheel suspension kit for tilt-in-space wheelchairs. The finding is comparable to the conclusions of other research completed on rear wheel suspension (Cooper, 2003; Kwarciak, 2003; Requejo, 2008; Requejo, 2009). In the present study, the spring-damper unit decreases vibration exposure at the rear wheel over a variety of different obstacles. Especially for wheelchair users with limited postural control, the use of rear wheel suspension may improve rider comfort (Requejo, 2008; Maeda, 2003). There is no conclusive evidence on the amount of shock and vibration exposure needed for a decrease in health risks, therefore we cannot definitively state the reductions found in the present study reduces the health risks associated with shock and vibration exposure. However, tilt-in-space wheelchair users spend large periods of time in their chair that even small reductions in shock and vibration most likely have positive health benefits. Additionally, large magnitude shocks may displace the person in the tilt-in-space wheelchair resulting in them having improper posture, which may lead to spinal deformities, pain, and discomfort. Tilt-in-space wheelchair users typically do not have the trunk innervation to readjust themselves to an optimal posture or even be aware they are not in an optimal posture; therefore, minimizing the shock exposure could impact the pain and discomfort associated with improper posture.

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APPENDIX

Colorado State University
Knowledge to Go Places

Research Integrity & Compliance Review Office
Office of the Vice President for Research
321 General Services Building - Campus Delivery 2011 eprotocol
TEL: (970) 491-1553
FAX: (970) 491-2293

NOTICE OF APPROVAL FOR HUMAN RESEARCH

DATE:	November 28, 2016	
TO:	Reiser, Raoul, 1582 Dept Hlth & Exer Sci	
	Braun, Barry, 1582 Dept Hlth & Exer Sci, Hischke, Molly	
FROM:	Swiss, Evelyn, CSU IRB 1	
PROTOCOL TITLE:	Effect of rear wheel size on shock absorption in manual wheelchairs	
FUNDING SOURCE:	NONE	
PROTOCOL NUMBER:	16-6933H	
APPROVAL PERIOD:	Approval Date: November 17, 2016	Expiration Date: November 16, 2017

The CSU Institutional Review Board (IRB) for the protection of human subjects has reviewed the protocol entitled: Effect of rear wheel size on shock absorption in manual wheelchairs. The project has been approved for the procedures and subjects described in the protocol. This protocol must be reviewed for renewal on a yearly basis for as long as the research remains active. Should the protocol not be renewed before expiration, all activities must cease until the protocol has been re-reviewed.

Important Reminder: If you will consent your participants with a signed consent document, it is your responsibility to use the consent form that has been finalized and uploaded into the consent section of eProtocol by the IRB coordinators. Failure to use the finalized consent form available to you in eProtocol is a reportable protocol violation.

If approval did not accompany a proposal when it was submitted to a sponsor, it is the PI's responsibility to provide the sponsor with the approval notice.

This approval is issued under Colorado State University's Federal Wide Assurance 00000647 with the Office for Human Research Protections (OHRP). If you have any questions regarding your obligations under CSU's Assurance, please do not hesitate to contact us.

Please direct any questions about the IRB's actions on this project to:

IRB Office - (970) 491-1553; RICRO_IRB@mail.Colostate.edu
Evelyn Swiss, Senior IRB Coordinator - (970) 491-1381; Evelyn.Swiss@Colostate.edu
Tammy Felton-Noyle, Assistant IRB Coordinator - (970) 491-1655; Tammy.Felton-Noyle@Colostate.edu

Evelyn Swiss

Swiss, Evelyn

Approval to recruit 12 adult participants with the approved recruitment and consent procedures. The above-referenced project was approved by the Institutional Review Board with the condition that the approved consent form is signed by the subjects and each subject is given a copy of the form. NO changes may be made to this document without first obtaining the approval of the IRB. Approved documents include: Consent Form, version 1; Wheel

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Size Recruitment Flyer, dated 10/24/2016; Wheel Size Recruitment Script, dated 10/24/2016.

Approval Period:	November 17, 2016 through November 16, 2017
Review Type:	EXPEDITED
IRB Number:	00010468

Consent to Participate in a Research Study

Colorado State University

TITLE OF STUDY: **Rear axle shock reduction in manual wheelchairs**

PRINCIPAL INVESTIGATOR: Raoul F. Reiser II, PhD. Department of Health and Exercise Science. Director of the Clinical Biomechanics Laboratory. Contact at (970) 491-6958 or Raoul.Reiser@Colostate.edu

WHY AM I BEING INVITED TO TAKE PART IN THIS RESEARCH? *You are being asked to volunteer for this research because you are an adult wheelchair user aged 18 years or older. More specifically, you have* a *Quickie Iris or Invacare Solara Tilt-in-Space wheelchair equipped with Quadshox rear suspension. You must also be willing to use an identical wheelchair without Quadshox for a short period of time, be willing to wear a snuggly fit helmet for a short period of time, have a caregiver that is able to accompany you to the CSU Main Campus and willing to push you during the course of this project.*

WHO IS DOING THE STUDY? *This research is being performed by Raoul F. Reiser II, Ph.D. of the Health and Exercise Science Department. Dr. Reiser is interested in clinical biomechanics. His work looks at both performance and injury aspects of movement.*

WHAT IS THE PURPOSE OF THIS STUDY? *Powered wheelchairs are equipped with shock absorbers on the rear axle. To save weight, they are not common equipment on manual wheelchairs. A new lightweight, rear-axle shock system has been developed for manual wheelchairs (Quadshox, Fort Collins, CO). However, shock reduction has not been quantified. The goal of this investigation is to examine the shock reducing capabilities of this new system.*

WHERE IS THE STUDY GOING TO TAKE PLACE AND HOW LONG WILL IT LAST? *This research project will take place on the CSU main campus. Your involvement will last roughly 2 hours during a single visit. You will need to have transportation to and from campus on the day of your visit.*

WHAT WILL I BE ASKED TO DO? *If you agree to participate you will be pushed by your caregiver at a comfortable pace over four surfaces that you might encounter on the CSU Campus. These surfaces range from a smooth indoor floor to a bumpy older outdoor sidewalk. You will be pushed over these surfaces both in your Quadshox equipped wheelchair as well as in an identical rigid wheelchair. For half of the trials you will need to wear a helmet snuggly on your head so that we can measure head motion. During the other half of the trials we will be measuring the motion (acceleration) of the wheelchair.*

ARE THERE REASONS WHY I SHOULD NOT TAKE PART IN THIS STUDY? *You should not volunteer for this study if you do not meet the criteria outlined above. Additionally, if you are a woman, you should not participate if you are pregnant. Regardless of gender, you should also not participate if you are uncomfortable with what you will be asked to do.*

WHAT ARE THE POSSIBLE RISKS AND DISCOMFORTS? *The risks associated with participation are no greater than daily living. You will have a familiar caregiver push you over the selected surfaces. You will not be pushed faster than what you are comfortable with. However, as you are aware, using a wheelchair does have inherent risks. There is the very low possibility that your chair could tip over. However, all surfaces will be flat and we will have your caregiver or trained personnel around you at all times. It is not possible to identify all potential risks in research procedures, but the researchers have taken reasonable safeguards to minimize any known and potential, but unknown, risks.*

WILL I BENEFIT FROM TAKING PART IN THIS STUDY? *While this study should provide useful information regarding the efficacy of Quadshox, there are no direct benefits to you. However, we will inspect your wheelchair to ensure it is in proper, working order. Assuming our investigation verifies a significant reduction in vibration and bumps transmitted to the user, our results should help facilitate their use and improve the wheelchair experience of those unable to walk.*

DO I HAVE TO TAKE PART IN THE STUDY? *Your participation in this research is voluntary. If you decide to participate in the study, you may withdraw your consent and stop participating at any time without penalty or loss of benefits to which you are otherwise entitled.*

WHAT WILL IT COST ME TO PARTICIPATE? *There are no costs to participate in this study. However, if you are injured during the course of involvement, you will be responsible for medical costs beyond the emergency treatment. You must provide your own transportation to and from the laboratory.*

WHO WILL SEE THE INFORMATION THAT I GIVE? *We will keep private all research records that identify you, to the extent allowed by law.*

Your information will be combined with information from other people taking part in the study. When we write about the study to share it with other researchers, we will write about the *combined information we have gathered. You will not be identified in these written materials. We may publish the results of this study; however, we will keep your name and other identifying information private.*

We will make every effort to prevent anyone who is not on the research team from knowing that you gave us information, or what that information is. For example, your name will be kept separate from your research records and these two things will be stored in different places under lock and key. You should know, however, that there are some circumstances in which we may have to show your information to other people. For example, the law may require us to show your information to a court. We may be asked to share the research files for audit purposes with the CSU Institutional Review Board ethics committee, if necessary. *The files containing information about you will be identified with a code, such as "WCS01," where WCS is short for Wheelchair Study and 01 is a subject number. Upon completion of data collection and verification of results, the list linking your name to the code will be destroyed.*

CAN MY TAKING PART IN THE STUDY END EARLY? *Your participation in the study may end early if you or your caregiver are unable to perform the tasks required of the study.*

WILL I RECEIVE ANY COMPENSATION FOR TAKING PART IN THIS STUDY? *There is no monetary compensation for your involvement in the study.*

WHAT HAPPENS IF I AM INJURED BECAUSE OF THE RESEARCH? *The Colorado Governmental Immunity Act determines and may limit Colorado State University's legal responsibility if an injury happens because of this study. Claims against the University must be filed within 180 days of the injury.*

In light of these laws, you are encouraged to evaluate your own health and disability insurance to determine whether you are covered for any physical injuries or emotional distresses you might sustain by participating in this research, since it may be necessary for you to rely on your individual coverage for any such injuries. Some health care coverages will not cover research-related expenses. If you sustain injuries, which you believe were caused by Colorado State University or its employees, we advise you to consult an attorney.

WHAT IF I HAVE QUESTIONS? *Before you decide whether to accept this invitation to take part in the study, please ask any questions that might come to mind now. Later, if you have questions about the study, you can contact the investigator, Raoul F. Reiser II, Ph.D. at 970-491-6958. If you have any questions about your rights as a volunteer in this research, contact the CSU IRB at: RICRO_IRB@mail.colostate.edu ; 970-491-1553. We will give you a copy of this consent form to take with you.*

Your signature acknowledges that you have read the information stated and willingly sign this consent form. Your signature also acknowledges that you have received, on the date signed, a copy of this document containing 3 pages.

____ ________________________

Signature of person agreeing to take part in the study Date

____

Printed name of person agreeing to take part in the study

____ ________________________

Name of person providing information to participant Date

____

Signature of Research Staff

PARENTAL/GUARDIAN SIGNATURE

As parent or guardian I authorize ___________________________ (print name) to become a participant for the described research. The nature and general purpose of the project have been satisfactorily explained to me by ________________________ and I am satisfied that proper precautions will be observed.

____

Parent/Guardian name (printed)

____ _____________________

Parent/Guardian signature Date