Good Vibes Only: An In-Depth Analysis on the Implementation of In-Wheel Suspension in Manual Wheelchair Users

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This dissertation aimed to investigate the potential of using in-wheel suspension to reduce harmful whole-body vibration (WBV) and improve comfort and mobility for manual wheelchair users (MWUs). The LoopWheels Urban is designed with three C-shaped carbon fiber springs to absorb vibration and provide a smoother ride and increased comfort to the MWU. Root Mean Square (RMS) for vibration and Vibration Dose Value (VDV) for more transient shocks are used as measures for WBV. LoopWheels was found to reduce harmful vibrations and shocks experienced by MWUs with spinal cord injury (n= 26) across various indoor/outdoor surfaces and obstacles by 10% at the backrest and 7% at the footrest (all $p < 0.05$) compared to standard spoked and Spinergy CLX wheels in a lab setting. Neck/back pain, fatigue, and WBV exposure were further analyzed through a 12-week community-based intervention with the LoopWheels. Participants experienced a median reduction of 15% in neck pain, 8% decrease in median perceived fatigue, and reported one fewer median pain problems (all $p < 0.05$) after the trial period. Community sensor data shows MWUs propelled an average of two hours and were exposed to WBV levels below hazardous thresholds defined by ISO 2631. LoopWheels shows a 35% reduction in vibration and a 50% reduction in shock when compared to previous community-based studies with standard wheels. Users indicated a smoother, more comfortable ride experience, but felt the wheels were harder to push. Rolling resistance (RR) testing indicated that LoopWheels had 118% more RR on linoleum and 44% more on carpet than standard spoked wheels. LoopWheels exhibited significantly higher static deformation (max 0.27 inch) compared to standard and CLX

wheels over all loading conditions ($p < 0.05$). Propulsion testing showed larger oscillatory amplitudes in the anterior-posterior and vertical direction that varied in phase from the standard and CLX wheels. Periodic behavior could be contributing to the sensation of reduced propulsion efficiency. While LoopWheels show promise in reducing vibration and improving health, challenges such as increased RR and dynamic behavior necessitate further exploration.

Table of Contents

List of Tables

List of Figures

Preface

The author would like to dedicate this dissertation to Allison Michalowski (fiancé); Sreeramachandra, Sreevani, and Satvika Neti (family); Moxie, Devika, Hazel, Coco (pets); friends; and loved ones. This process would be impossible without the consistent and great support provided by everyone around. The author would also like to give significant acknowledgement to the following undergraduate students who assisted on the conduction of this study: Celia Davis, Emma Vukovich, Allison Brunswick, Richard Nott, and Logan Marsalko. The following staff members are also noted for their significant contributions: Benjamin Gebrosky, Jonathan Nocket, and Chloe Shearer. Finally, the author would like to thank all members of the Human Engineering Research Laboratories for their support. This research was funded by Craig H. Neilsen Foundation Project 651619. The findings and views presented in this dissertation do not represent the view of the United States of America or the Department of Veterans Affairs.

1.0 Introduction

1.1 Manual Wheelchair Users and Daily Activities

Individuals who rely on wheelchairs constitute a significant portion of the population, currently estimated at approximately 65 million individuals according to the World Health Organization [1]. The need for wheelchair use can arise from a variety of medical conditions, including spina bifida, cerebral palsy, and spinal cord injury (SCI). The selection of a manual or power wheelchair is typically based on factors such as the individual's specific needs, functional capabilities, and environmental considerations. For instance, individuals with advanced cerebral palsy may lack the dexterity required to maneuver a manual wheelchair effectively. In such cases, a power wheelchair may be prescribed to accommodate their mobility needs. Conversely, individuals with a low-level complete SCI who retain trunk control and upper extremity function often find manual wheelchairs suitable for their mobility requirements. The choice between manual and power wheelchairs is tailored to each individual's abilities and circumstances, aiming to optimize independence and quality of life.

Despite the diverse range of conditions leading to wheelchair use, individuals engage in a variety of daily activities aimed at maintaining independence and social participation [2]. These activities encompass both indoor and outdoor tasks, including essential self-care activities such as grooming and household chores, as well as work-related tasks and recreational pursuits [1, 2]. Outdoor mobility presents unique challenges for wheelchair users, as they navigate diverse environments such as urban streets, sidewalks, and recreational areas. Accessing these spaces is essential for social integration and participation in community life. However, environmental barriers, including uneven surfaces and lack of accessibility features, can impede mobility and limit opportunities for engagement [2, 3].

1.2 Natural and Built Environment Causes Whole-Body-Vibration

The natural and built environment presents numerous challenges for manual wheelchair users (MWUs), impacting their daily mobility and health. MWUs navigate diverse terrains and surfaces, encountering various obstacles such as curbs, thresholds, and grassy areas, which contribute to significant levels of whole-body vibration (WBV) exposure [4-6]. WBV is a known risk factor for several adverse health outcomes, particularly among MWUs who spend extended periods in their chairs and encompasses both long-term vibrations and short-term shocks [4, 6-9]. Prolonged exposure to WBV has been linked to increased prevalence of back and neck pain, disc degeneration, and musculoskeletal disorders [4, 6-9]. Studies have shown that MWUs experience elevated levels of WBV that surpass safety standards set by the International Standards Organization (ISO), putting them at heightened risk for pain and injury [6-8, 10-12]. The detrimental effects of WBV are further compounded by a range of risk factors prevalent among MWUs [7, 8, 13]. Prolonged seating, awkward postures, and vibration exposure exacerbate discomfort and contribute to musculoskeletal disorders. Additionally, individuals with SCI often exhibit spinal deformities, posterior pelvic tilt, and lack of trunk control, predisposing them to unnatural body positions that increase susceptibility to WBV-related injuries.

Efforts to mitigate WBV exposure through various seating systems and cushion characteristics have yielded mixed results. While some studies have explored the efficacy of different suspension systems in minimizing vibration transmission, such systems have sometimes been found to amplify harmful vibrations, particularly those occurring within the frequency range most detrimental to human health [4, 6, 14-16]. The challenging terrain and surfaces encountered in the natural and built environment highlight the urgent need for effective solutions to minimize WBV exposure for MWUs. The pain felt by MWUs is typically focused on the upper back, neck, and shoulders. As most MWUs rely on the function of their upper extremities, this pain can cause a significant barrier to completing activities of daily living (ADLs) [7, 8]. The completion of ADLs independently has a critical role in the quality of life for MWUs. Research has shown that this pain and discomfort can also lead to sedentary lifestyles, decreasing overall activity and cardiovascular health [7, 8, 17, 18]. Furthermore, individuals report that being in a wheelchair was the number one limiting factor in community participation; more than their disability, pain, or other factors [19]. It is possible that the lack of access to specific areas of the environment or the presence of unavoidable obstacles that could be causing this feeling of limitation among MWUs.

1.3 Current Methods to Reduce Vibration

The pursuit of mitigating whole-body vibration (WBV) for manual wheelchair users (MWUs) has spurred the development of various suspension systems, including frame material, accessories, front caster suspension, in-frame suspension, and in-wheel suspension. However, traditional methods \ have faced challenges in significantly reducing WBV exposure for MWUs [4, 14, 16, 20]. A short table highlighting some previous work and the results is shown in Table 1.

Different materials used from the wheelchair frame have been investigated by Chenier et al and found that between carbon based wheelchair frames had the greatest reduction in vibration transmission compared to aluminum and titanium frames, which showed comparable results [15].

Furthermore, their results revealed that there was a direct negative relationship between vibration transmissibility and work required during propulsion; the more a material reduces vibration the more work is required to move the same distance [15]. Other methods that hope to reduce vibrations are by introducing accessories such as seat cushions and backrests. Cooper et al. found that the effect of these accessories in reducing vibration is greatly dependent on the material composition and highly individual between MWUs [4].

In-frame suspension systems, characterized by shock-absorbing spring coils integrated into the wheelchair frame, initially held promise as a solution. However, studies have revealed their limitations in adequately attenuating WBV transmission. For instance, research conducted by Kwarciak et al. compared various suspension manual wheelchairs and found that they showed very little promise in reducing vibration and shocks over uneven terrain, revealing that in-frame suspension systems often fall short in providing sufficient vibration absorption during propulsion [21]. Alternatively, in-frame suspension systems integrate springs or shock absorbers into the wheelchair frame, deflecting under rough surface conditions. However, their uni-directional operation and limited effectiveness in reducing WBV have been noted in studies, emphasizing the need for more advanced suspension solutions [16, 21, 22]. In summary, while traditional suspension methods offer some reduction in WBV for MWUs, their limitations highlight the need for further innovation and research to enhance vibration mitigation in manual wheelchairs.

Similarly, front caster suspension systems, exemplified by products like Frog Legs, aim to absorb shocks encountered by the wheelchair's front casters using carbon fiber spring levers and polymer dampening elements. While laboratory tests have shown reduced peak accelerations [22], real-world effectiveness remains limited. Misch and Sprigle demonstrated that wheelchairs fitted with suspension casters are exposed to nearly 100% larger vibrations at the seat panel and 43% larger vibrations at the front caster compared from a default configuration [20].

Table 1: Summary of previous work in vibration attenuation

	Investigation	Outcomes
Frame Material	Frame material on vibration transmisability and work per meter. Alluminium, titanium, and carbon.	Carbon had lowest vibration. No difference between titanium and aluminium. Found that the more a material reduced vibration exposure, the more energy was required to move the same distance [15].
Cushions and Backrests	Cushions and backrest effects on vibration during propulsion	Mixed results on vibration attenuation. Material and configuration play a large role on effect. Determined that most harmful vibrations were from shocks and not oscilations or posture correcting.[4]
In Frame	Investigated wheelchair frames with rear-suspension compared to ones without suspension.	Rear-suspension showed to reduce some exposure to shock and vibration, but not superior to traditional frames. [22]
	Determined functionality of different suspension wheelchairs in reducing vibration exposure over uneven terrain and on curb decents.	While one wheelchair outperformed others, results indicate that overall suspension wheelchairs are not suiable to reduce vibration or shock exposed to MWUs. [21]

Front Caster	Tested the effect of front castor suspension over four different surfaces on reducing vibration and on propulsion cost.	Suspension casters showed to increase vibrations at the footrest and seat by nearly 100% compared to a defulat configuration.[20]

1.4 In-Wheel Suspension

In the realm of manual wheelchair technology, innovative solutions are being developed to mitigate the adverse effects of whole-body vibration (WBV) on users. Among these advancements, in-wheel suspension systems have garnered attention for their potential to enhance comfort and reduce vibration transmission during wheelchair use. One prominent example of such technology is the LoopWheels Urban [23], a product engineered by Jelly Products Ltd. LoopWheels Urban features an aluminum rim integrated with a central hub connected via three C-shaped carbon fiber springs. These springs are strategically designed to absorb and deflect shocks and vibrations encountered during wheelchair propulsion, thereby minimizing the transmission of WBV to the user. The Spinergy CLX [24] wheel employs a sophisticated design, incorporating a carbon fiber rim and spokes made of a specialized synthetic fiber known for its vibration-dampening properties. By leveraging these materials and construction techniques, Spinergy aims to offer MWUs a smoother and more comfortable ride while reducing the impact of WBV on their bodies.

1.5 Investigation Of In-Wheel Suspension in Reducing WBV Among Experienced MWUs

This dissertation constitutes a comprehensive exploration into in-wheel suspension systems for MWUs, with an examination of their short-term and long-term effects, as well as an in-depth analysis of their mechanical characteristics. To this end, this specific aims of this project were to:

1.5.1 Specific Aim 1: Investigate The Effect of In-Wheel Suspension Systems on Reducing Vibration and Shock And Improving Comfort In MWUs

The effects of in-wheel suspension systems on vibration, shock, and overall comfort were first analyzed in a controlled in-laboratory study with experienced MWUs. By conducting a controlled experiment replicating real-world conditions, this study endeavored to quantify the reduction in WBV transmission achieved by in-wheel suspension systems, offering valuable insights into their efficacy in enhancing the comfort and well-being of MWUs during wheelchair use.

1.5.2 Specific Aim 2: Assess The Longitudinal Effect of In-Wheel Suspension Systems on Reducing Pain and Fatigue And Improving Mobility In MWUs

To understand the impact that in wheel suspension had on reducing pain and fatigue and enhancing community participation, MWUs were provided the LoopWheels to use for 3 months. Community WBV and activity levels (propulsion and non-propulsion time) were also recorded via sensors on the wheelchair to determine if the amount of WBV exposure was within safe limits.

The association between participant demographics, WBV, activity levels, and perceived improvements in quality of life from wheel use were further investigated to identify factors that may provide insight into specific MWUs who would be more likely to benefit from the technology.

1.5.3 Specific Aim 3: Characterize The Rolling Resistance and Propulsion Mechanics of In-Wheel Suspension Systems in MWUs

Feedback received from the MWUs in the longitudinal study prompted additional investigation into design factors that may adversely affect the performance of the LoopWheels. This investigation focused on characterizing design attributes including rolling resistance, axle deformation/displacement and the periodic oscillatory behavior of the springs in an effort to explain reports of perceived propulsion inefficacy.

With these aims, the dissertation seeks to advance the understanding of in-wheel suspension systems for MWUs and their potential to mitigate WBV exposure, alleviate pain and discomfort, and enhance overall well-being. By elucidating both the short-term and long-term effects, as well as delving into the mechanical aspects of these systems, the dissertation aims to provide a holistic perspective on their efficacy, potential benefits and shortcomings. Ultimately, the goal is to contribute to the advancement of wheelchair technology and the improvement of overall health and quality of life for MWUs, thereby fostering more informed decision-making and facilitating enhanced mobility and independence.

2.0 Effects of In-Wheel Suspension on Whole Body Vibration and Comfort in Manual Wheelchair Users

This chapter is currently under review for publication in the MDPI Vibrations Journal

2.1 Chapter Summary

Frequent and prolonged exposure to high levels of vibration and shock can cause neck and back pain and discomfort for many wheelchair users. Current methods to attenuate the vibration have shown to be ineffective and in some cases detrimental to health. Novel in-wheel suspension claim to offer a solution by replacing traditional spokes of the rear wheels with dampening elements or springs. The objective of this study was to investigate the effects of in-wheel suspension on reducing vibration and shock and improving comfort in manual wheelchair users. Twenty-four manual wheelchair users propelled over nine different surfaces using a standard spoked wheel, a Spinergy CLX and LoopWheels while accelerometry data was collected at the footrest, seat and backrest. LoopWheels lowered vibrations by 10% at the backrest compared to the standard and CLX wheels (p-value < 0.001) and by 7% at the footrest compared to the CLX (p-value <0.05). It also reduced shocks by 7% at the backrest compared to the standard wheel and CLX (p-value <0.001). No significant differences were found in comfort between wheels. Results indicate that LoopWheels are effective at reducing vibration and shock, but more long-term testing is required to determine effects on health.

2.2 Introduction

Manual wheelchair users (MWUs) propel in many environments and over a wide variety of surfaces every day, including; linoleum, carpet, thresholds, sidewalks, asphalt, gravel, and grass [2, 10]. Many of these surfaces have the potential to inflict large amounts of whole-body vibration (WBV) to the MWU. WBV is typically characterized by both low impact, long-term vibrations and high impact, short-term shocks (e.g. driving over a gravel road versus hitting a large pot hole on a smooth road, respectively). Given the frequent and pervasive nature of these surfaces in the home, natural, and built environments, MWUs are unable to avoid many of these surfaces while participating in their daily activities [4, 6, 10]. For example, when at an intersection, many MWUs are forced to choose between propelling over the rumble strips on the curb-cuts or performing a curb drop; both cases would induce large vibrations and/or shocks. Exposure to high levels of WBV has been linked to several adverse health outcomes including neck and back pain/discomfort and neuromuscular fatigue in several populations including commercial truck drivers, factory workers, and wheelchair users [7, 10, 13, 25].

2.2.1 Whole Body Vibration and Impact on Health

Given the risk of health consequences associated with WBV, the International Standards Organization (ISO) has published several guidelines on levels of vibration exposure and their relationship to health outcomes, specifically ISO 2631 defines WBV exposure from a seated position [11, 12]. Typically, vibrations and shocks in the vertical direction are investigated to determine effects on health, but ISO 2631 also outlines what to do to analyze other directions [11, 12]. Typically, the acceleration in the vertical direction is analyzed to determine vibration with two values, root-mean square (RMS) measure long-term, low amplitude and the vibration dose value (VDV) measure short-term, high amplitude [6, 11, 12, 26]. ISO 2631 has also released documentation on the Health Guidance Caution Zone (HGCZ), a region of vibration intensity that has the potential to cause injury [12, 27]. This zone is determined for both RMS and VDV values under different lengths of exposure time [12, 27]. For general purposes, ISO 2631 defines the HGCZ for both RMS and VDV by the 10-minute exposure values, as anything shorter than this has the same threshold values for the HGCZ [10, 12]. Researchers across many disciplines utilize the ISO 2631 standards as a framework for analyzing exposure to vibration.

In studies that have investigated vibration exposure among MWUs, it was found that current manual wheelchair frames are not sufficient in reducing WBV [4, 21, 22]. Using novel materials in the frame, such as carbon, have shown to significantly reduce vibration transmissibility compared to aluminum and titanium frames, but it was not investigated whether this improved health outcomes in MWUs [15]. One study investigated vibration exposure in MWUs during two weeks of community and home use. The re-searchers found that all 37 participants experienced RMS and VDV values that were within or exceeded the HGCZ [7]. They also found that MWUs are exposed to unsafe levels of WBV around 8-13 hours each day and that wheelchair frames with suspension elements did not attenuate vibration significantly [7].

2.2.2 Current Methods to Reduce WBV

Current methods to reduce WBV in MWU include in-frame, front castor, or in-wheel suspension systems. In frame suspension systems have shock absorbing spring coils connected between the seat panel and the axle housing. These springs are designed to compress and expand when exposed to vibration and reduce the amount transmitted to the seat [21, 22]. As the springs have a fixed orientation within the frame, they are not optimized to handle vibrations and shocks coming from multiple directions, such as when ascending or descending curbs and door thresholds. Other drawbacks include more moving parts in the frame which typically leads to more repairs or maintenance required and added weight to the wheelchair which can increase propulsion forces. Front caster suspension systems include products like the Frog Legs, which consist of a carbon fiber spring lever and polymer dampening element that enables the castor to move vertically when encountering an obstacle and then return to its rest position [20, 22]. One study using test dummies and an American National Standards Institute/Rehabilitation Engineering Society of North America (ANSI/RESNA) double drum tester found that wheelchairs with suspension castors has significantly lower peak accelerations than those with the original equipment manufactured caster at both the seat and footrest [22]. Another study utilized a robotic propulsion system to simulate a MWU propelling with and without the suspension casters over more common everyday surfaces (brick, aluminum grates, sidewalk, and smooth tile) [20]. This study found that when using the suspension casters, the peak accelerations at the seat increased by nearly 100% depending on the surface type while at the footrest, there was an average of 43% increase in accelerations over all surfaces [20]. Unfortunately these tests were only conducted with a surrogate for the MWU, which fails to take into account biological factors [5, 9, 20, 22, 28]. Additionally, WBV values while using suspension systems are still within or exceed the HGCZ, indicating that further efforts are required to reduce vibration to a safe level [21, 22]. Other accessories such as seat cushions and back supports have shown to have some benefit in reducing vibration based on materials used, but their performance is highly dependent on the person’s posture and how much muscle tension is present [4].

2.2.3 In-Wheel Suspension

In-wheel suspension systems replace the mags or spokes of the rear wheel with multiple springs that are designed to absorb shock and vibration in all directions as opposed to just one. When the wheelchair encounters an obstacle, the springs allow the axle of the wheel to deflect while keeping the wheel level with respect to the ground (Figure 1). As it is very common for a MWU to perform a wheelie when performing a curb drop or traversing a threshold, the in-wheel suspension systems are more effective than front caster suspension in these scenarios. Additionally, these wheels have a quick-release axle housing which is common to many wheelchair make and models, enabling them to be retrofitted to an existing quick release wheelchair axle frame. There is limited research on how well these systems do at attenuating shock and vibration and improving rider comfort. The LoopWheels Urban (LoopWheels Urban, Jelly Products Ltd, Nottinghamshire, United Kingdom), a design encompassing three “C” shaped carbon fiber springs [23], and SoftWheels (SoftWheels, SoftWheels Ltd, Tel Aviv, Isreal), a similar design using gas springs, has been studied along with several other rigid rear wheel types using a robotic wheelchair that simulates MWU propulsion [20]. In this study, the same surfaces were tested as described earlier for the front caster suspension. Researchers found that LoopWheels increased vibrations by 12-26% at the seat and the footrest; however, SoftWheels showed to be effective at reducing vibrations by 11% over some surfaces at the frame under the seat panel [20]. Another study, focusing specifically on SoftWheels, aimed to investigate rider comfort when using in-wheel suspension [29]. This study tested 24 new wheelchair users (recently admitted to a rehabilitation facility) and found that the participants noticed significant improvement when using the suspension wheel over a non-suspension wheel in two specific areas: decreased the amount of bumps felt and feeling more confident when riding in the chair [29].

Unfortunately, this study failed to test wheel-chair users accustomed to propulsion and did not measure the amount of vibration transmitted to the wheelchair. To our knowledge no vibration related studies on in-wheel suspension have been conducted with experienced MWUs; for example, those having more time to acclimate to their disability and with more wheelchair skills experience.

2.2.4 Study Objectives and Hypothesis

Therefore, the goal of this study was to determine the effects of in-wheel suspension on reducing WBV and improving ride comfort in experienced MWUs. As certain materials used to construct wheelchair wheels may exhibit dampening properties, we also included a Spinergy CLX (CLX, Spinergy, San Marcos, California, USA) as another condition in our test protocol, which uses carbon fiber on the rim and Zylon (PBO, poly(p-phenylene-2,6-benzobisoxazole)) spokes [24]. The LoopWheels was also selected for this study as SoftWheels was currently not on the market when this study was conducted. A secondary goal of this study was to compare the WBV values from using suspension wheels to the HGCZ provided by ISO 2631.

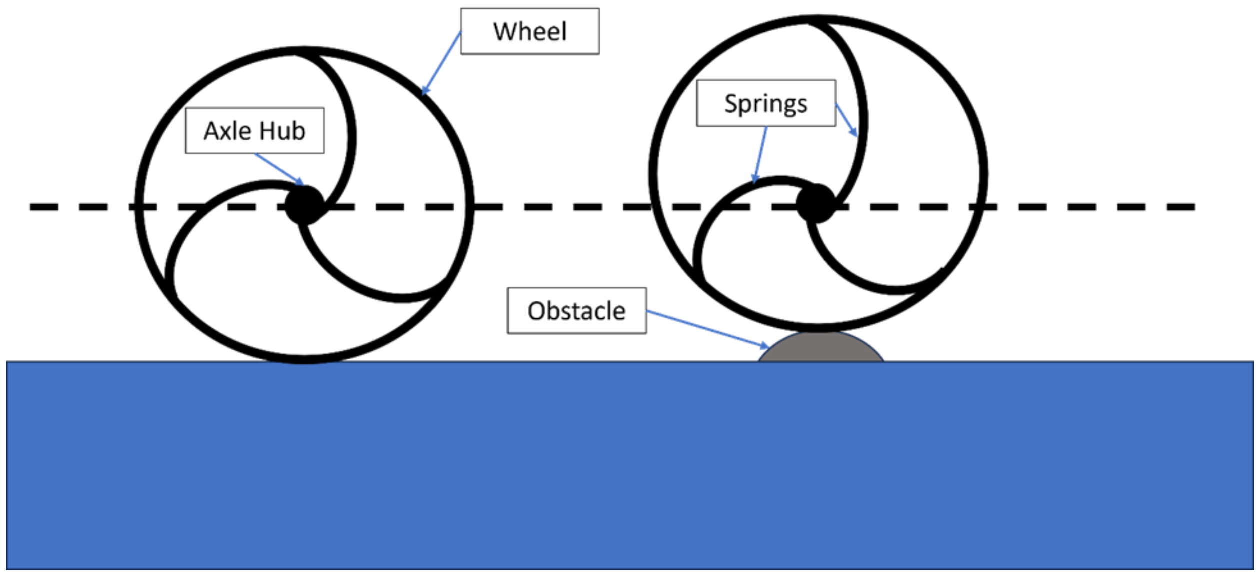


Figure 1: How the in-wheel suspension system works when it encounters an obstacle. The axle hub of the wheel is able to deflect relative to the rim of the wheel and stay level relative to the ground.

We hypothesized that the RMS vibration and VDV shock values experienced by the wheelchair frame at the backrest, under the seat panel, and on the footrest would be highest for a pair of standard spoked wheels (Quickie, SouthwestMedical LLC, Phoenix, Arizona) followed by the CLX wheels and lastly the LoopWheels. We also hypothesized that using the LoopWheels over different surfaces will be most comfortable followed by the CLX and lastly the standard wheels.

2.3 Materials

A mobility course composed of nine different surface types designed to simulate a variety of surfaces that MWUs are exposed to was used in this study [1-3] (Figure 2). The surfaces and obstacles were set up similar to other studies using a simulated road course where the surfaces and obstacles were performed in succession instead of in isolation [6]. The mobility course was split into three portions by level of intensity. Factors such as surface roughness, texture, and incline were evaluated when determining which sur-faces and obstacles should be in each category. The low intensity portion contains a high-pile carpet (HPC), low-pile carpet (LPC), and a textured rubber mat (RBR). The moderate intensity portion contains a slat-board with randomly spaced non-uniform gaps (SLT), a small bump (BMP), and a foam wobble board (FBD). The high intensity portion involves a ramp up (RMP), 4-inch curb drop (CRB), and a gravel board (GRV). Three wheel types were tested: standard spoke wheels, Spinergy CLX wheels, and LoopWheels Urban wheels (Figure 3). The standard spoked wheel, manufactured by Quickie, has a rim made of aluminum and 30 aluminum spokes (15 equally spaced on each side of the wheel hub). The Spinergy CLX wheel is made with a carbon fiber rim and 18 Zylon PBO polymer spokes. The LoopWheels is made with an aluminum rim and three C-shaped "spokes" made of ultralightweight carbon fiber. LoopWheels C-shaped springs are designed by the manufacturer in three different stiffnesses: soft, regular, and hard. The specific stiffness given to a MWU is based on their weight. For this in lab testing, all subjects used the standard stiffness springs. All wheels were outfitted with standard ½" (1.27 cm) diameter anodized push-rims and Marathon Plus tires filled to the recommended 125 psi. Furthermore, custom wheel covers were applied to both sides of the wheel to blind the participant to wheel type during testing. Three Shimmer3 IMU (Shimmer3, Shimmer Sensing, Dublin, Ireland) sensors were attached to three points on the participants' wheelchairs: at the footrest, under the seat panel, and on the backrest. Data collection was facilitated wirelessly by a custom Matlab (Matlab 2023a, MathWorks Ltd, Natick, Massachusetts, USA) application built from documentation provided by Shimmer. Sensors were secured to the wheelchair frame at these locations with zip-ties to ensure no movement artifact during testing. A diagram showing example placements of the sensors and the coordinate system is shown in Figure 4.

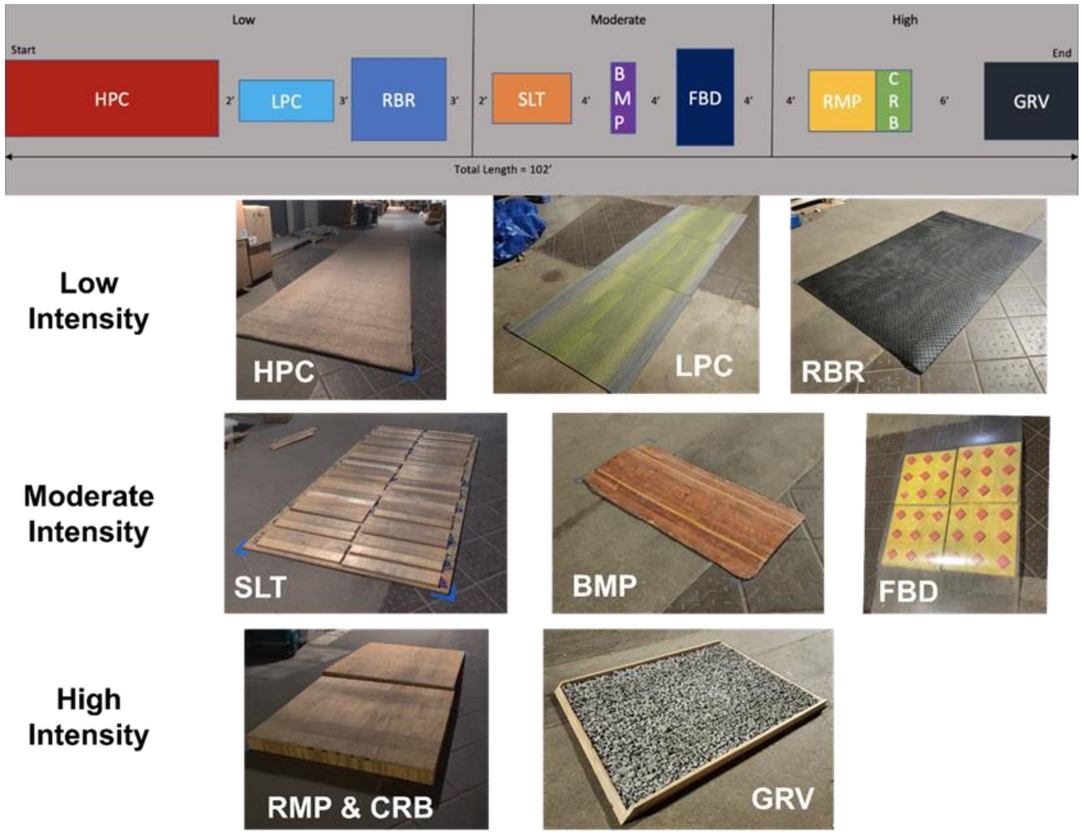


Figure 2: The surfaces, obstacles and course layout used for this study. The course was split into three intensity regions and three surfaces/obstacles in each region. These regions were physically marked off with tape on the floor.

Figure 3:The three different wheel types used in this study.

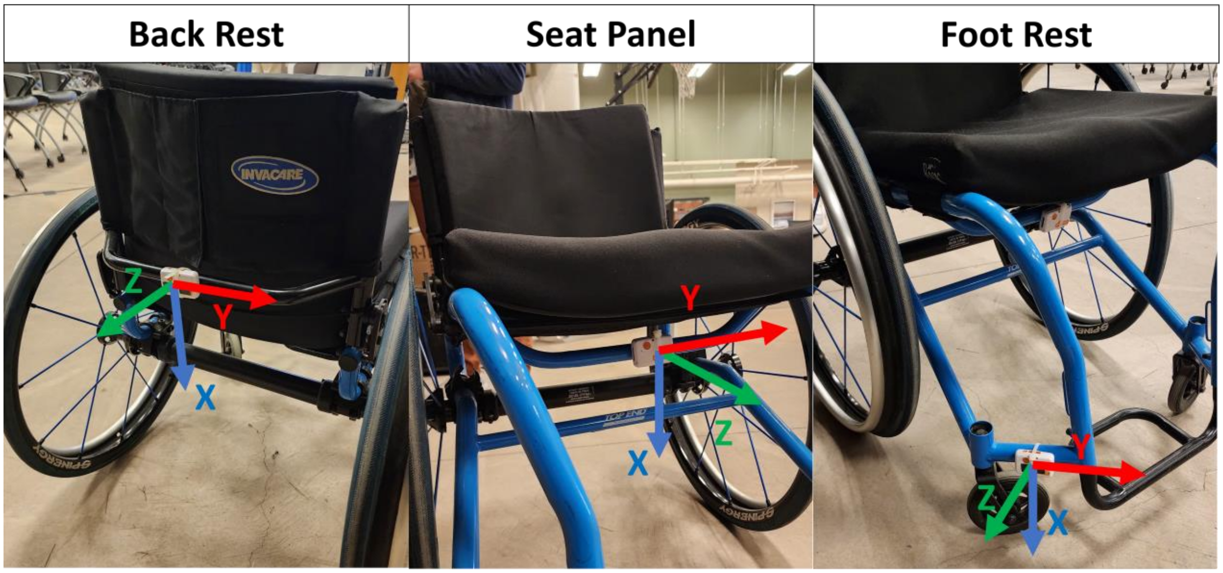


Figure 4: Approximate locations Shimmer3 sensors placed on the wheelchair along with the respective coordinate system.

2.4 Methods

2.4.1 Participant Recruitment:

This study was approved by the University of Pittsburgh Internal Review Board (ID: 20020199). Participants were recruited to the study through research registries, local disability community gatherings and events, and word of mouth. The inclusion criteria for participation were: (1) be at least 18 years old, (2) have a spinal cord injury diagnosis at least one year prior to the start of the study, (3) use a manual wheelchair as primary mode of mobility, (4) use 24 inch diameter wheels with quick-release axle pins, (5) no other suspension elements in frame or castors, (6) weigh under 265 pounds (maximum weight limit of the suspension wheels), (7) self-reported moderate chronic neck and/or back pain, and (8) be proficient in verbal and written English.

Individuals were excluded from the study if they had: (1) a history of fractures or dislocations in the upper extremity which has not fully recovered, (2) upper limb pain that interferes with propulsion, (3) severe cognitive limitations, or (4) current or recent history of pressure sores.

2.4.2 Study Protocol:

After informed consent, the participant completed a socio-demographic questionnaire and was given their first set of wheels to test. Wheels were retrofit onto the participants' wheelchair frames via their quick release axle. Sensors were placed at the three locations on their wheelchair frame: on the backrest, under the seat, and on or near the footrest. The sensor on the backrest was attached to the grab bar or a horizontal piece of the frame on the backrest at the midway point. The sensor under the seat panel was secured to the frame directly under the seat panel and centered under the buttock region to the extent possible. The last sensor was placed on the right side of the frame next to the foot plate or footrest if applicable. All sensors were positioned and oriented in a way that the -X axis was the closest to pointing up and the -Z axis in the anterior direction. In some circumstances it was not possible to position the sensors at these locations and with the -X axis pointing up due to the variations in wheelchair frame geometry. Sensors were positioned as close to the ideal location as possible provided the given geometry. An example of sensor placement for two different manual wheelchairs is shown in Figure 5. Acceleration data in all three axis will be calculated independently according to the ISO 2631 and compared to determine the direction of the most vibration exposure. The axis with the largest magnitude of acceleration will be used to calculate the WBV for that sensor. In the event of two axis with similar levels of acceleration, the ISO 2631 outlines how to combine multiple directions with individual weighting factors.

The order of the wheels was randomized per participant. Before testing began, a random order of wheels (generated using a random function in Excel) was assigned to each participant ID number. After placing the test set on each wheelchair they propelled over the entire mobility course three times with rest in between trials. They were asked to propel over the course at a self-selected speed consisting of a comfortable and casual pace. Practice runs were used to familiarize themselves with the wheels, course and the speed. Participants were also asked to pause for five seconds between each intensity region. Raw acceleration data in three directions was collected by the Shimmer3 sensors at 100 Hz consistent with the vibration standards [11, 12] during the entire trial and stored into a CSV file for later processing. If participants finished the course too soon or too fast (e.g. within ± 1 sec) of baseline the trial was repeated. In between wheel types, the participant completed a study-specific survey modeled after a prior study on ride comfort [30]. The survey asked for the participant's general comfort going over each of the nine surfaces and obstacles on a 10-cm visual analog scale, where a higher number indicates greater comfort. A comfort rating was provided for each obstacle independently. Additional questions regarding stability, security, pushrim comfort, maneuverability, and efficiency while using the wheels were also asked where the answer choices were on a 5 point Likert scale ranging from Not at all (1) to Extremely (5) for each factor evaluated. A copy of this survey can be found in Appendix A.

2.4.3 Data Analysis:

To correct for differences in sensor position and placement between wheelchair frames, each sensor's coordinate system was rotated so that the z-axis represents the vertical with respect to gravity [11, 12]. First, an original coordinate system (O) was established with an identity matrix (Equation 2-1). The average accelerations in each axis were used to establish a gravity vector (G), since gravity was the only acceleration present when the participant was not moving (Equation 2-2). A new z-axis (NZ) was established using the normalized gravity vector (Equation 2-3). Crossing it with the original z-axis (the axis oriented in the anterior-posterior direction) yields the new x-axis (NX) (Equations 2-4 & 2-5). The last axis, the y-axis (NY), was calculated by normalizing the cross product of the new z and new x axis (Equations 2-6 & 2-7). Finally, to establish the rotation matrix for the rotation of the sensor to gravity (Rsensor), the new coordinate system was multiplied with the inverse of the original coordinate system (Equation 2-8). This rotation matrix was then used to rotate the acceleration data collected during the trial (Equation 2-9). Analysis revealed that the acceleration data in the vertical direction (z-axis) had the largest magnitude compared to the other axes for all sensor locations, therefore all vibration calculations were performed with the new z-axis (Equation 2-10). The rotated acceleration values were passed through a 4th order Butterworth bandpass filter of frequencies within 0.4 to 100 Hz (Equation 2-11); values taken from ISO 2631[11, 12]. Next, a powerband filter was applied to extract the first third octave of frequency domain data, as these are considered to impact whole body vibration [26] (Equation 2-11). The total trial acceleration data was then split into each intensity region manually by visual analysis of the raw acceleration data. Since the participants were asked to start from rest and end with a complete stop, the boundaries could be easily identified. Starting and ending effects from each intensity region were trimmed to avoid the effects of starting and stopping propulsion on vibration. Additional post-processing details for the rotation of sensor coordinate systems to gravity, filtering of raw acceleration data from ISO 2631 guidelines, and separation of mobility course trial into three intensity regions can be found in Appendix B.

$$O = \begin{bmatrix} 1 & 0 & 0 \\ 0 & 1 & 0 \\ 0 & 0 & 1 \end{bmatrix} \tag{2-1}$$

$$G = \left[\sqrt{{a_{x_{raw}}}^2} \; \sqrt{{a_{y_{raw}}}^2} \; \sqrt{{a_{z_{raw}}}^2} \right] \tag{2-2}$$

$$N_z = \frac{G}{|G|} \tag{2-3}$$

$$N_x = O_z \times N_z \tag{2-4}$$

$$N_x = \frac{N_x}{|N_x|} \tag{2-5}$$

$$N_y = N_z \times N_x \tag{2-6}$$

$$N_y = \frac{N_y}{|N_y|} \tag{2-7}$$

$$R_{sensor} = N * O^{-1} \tag{2-8}$$

$$A_{corrected} = R_{sensor} * A_{raw} \tag{2-9}$$

$$A_{corrected} = [a_{x_{corrected}}, a_{y_{corrected}}, a_{z_{corrected}}] \tag{2-10}$$

$$a_{z_{weighted}} = bandpower\left(bandpass\left(a_{z_{corrected}}\right)\right) \tag{2-11}$$

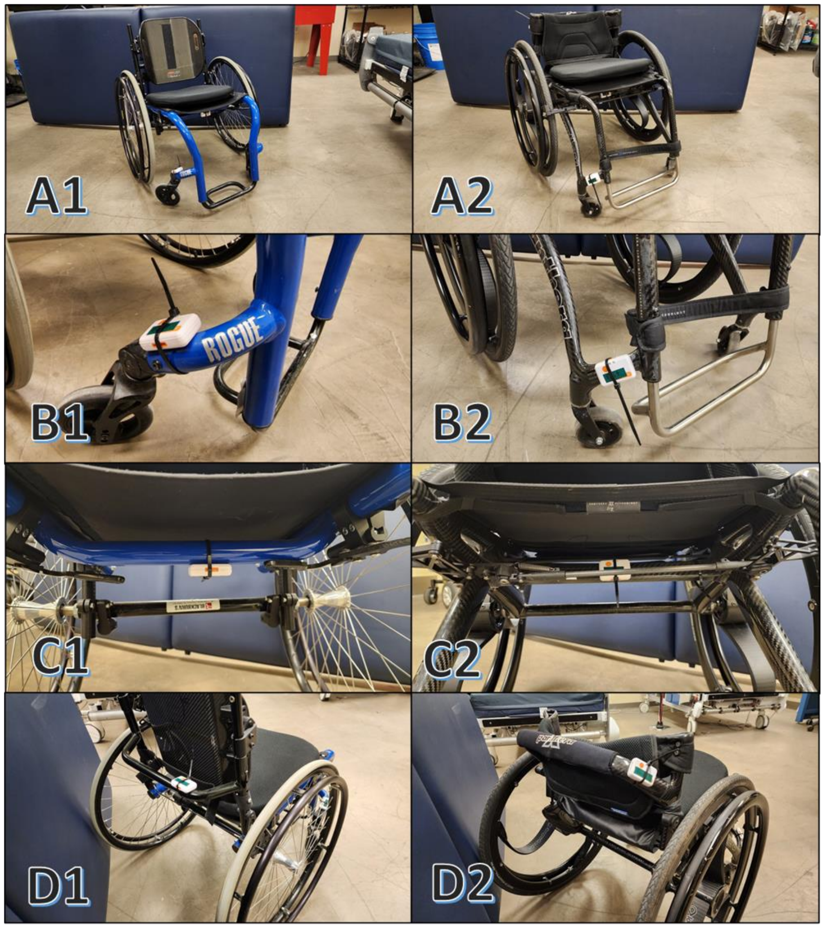


Figure 5: The image above shows differences in sensor position between two different manual wheelchairs , a ROGUE (1) and a PantheraX (2). Overall (A), foot rest (B), seat panel (C), and back rest (D) differences are shown along with sensor orientation.

2.4.4 Vibration Calculations:

Two metrics of shock and vibration in accordance with the methods and procedures set by the ISO 2631 were computed [11, 12]. Root-mean square (RMS, Equation 2-12) is a validated calculation that represents the high frequency, low amplitude vibrations from accelerometer data over a specified period of time (T); on the other hand, vibration dose value (VDV, Equation 2-13) is used for describing low frequency, high magnitude shocks from accelerometer data [11, 12].

Previous studies investigating WBV have included both RMS and VDV values when propelling over a simulated road course [6]. Filtered and weighted accelerometer data were processed through Matlab 2023a using methods derived from ISO 2631 and used to calculate RMS and VDV for each wheel type and for each intensity region. The crest fac-tor was determined to evaluate the validity of the RMS data for the trials. A higher crest factor indicates the presence of low-frequency, high amplitude shocks, suggesting VDV values should be analyzed in addition to RMS for describing the WBV [7]. The crest factor is calculated by taking the ratio of the peak filtered acceleration and the RMS value for that trial and the threshold is given at 9 [11]. If the crest factor is over the threshold for the ISO 2631 analysis, then both RMS and VDV values should be used for describing WBV. The final filtered, rotated, and frequency weighted z-axis acceleration was used to calculate the RMS (Equation 2-12) and VDV (Equation 2-13) for each trial and is oriented in the same direction as gravity, the axis specified by ISO 2631 for vibration analysis. Additionally, the RMS and VDV values were compared against the ISO 2631 HGCZ intervals: RMS: 0.43 – 0.86 m/s2 and VDV: 8.5 – 17 m/s1.75 [11]. These intervals are defined for an average vibration exposure time of 4-8 hours as these standards were originally designed for occupational use, values below this range have not been clearly investigated in regards to their effect on health [11]. Values of WBV exposure within these intervals should be treated with caution as they have a potential to cause health risks and values above this region are highly likely to cause health risks [11]. Final RMS and VDV values were aver-aged over the three trials per wheel type per intensity region.

$$RMS_{weighted} = \sqrt{\frac{1}{T}\sum a_{z_{weighted}}^2} \tag{2-12}$$

$$VDV_{weighted} = \sqrt[4]{\sum a_{z_{weighted}}^4} \tag{2-13}$$

2.4.5 Ride Comfort Survey:

The results from the comfort survey were compiled for each wheel type and averaged over all the participants. Comfort for each obstacle was calculated from the VAS by measuring the distance in centimeters from the zero point to the mark indicated by the participant for that specific obstacle. An overall average comfort score across all surfaces was calculated for each participant and wheel type.

2.4.6 Statistical Tests:

All statistical tests were conducted in R Studio 2023.09.0 (RStudio, R-Tools Technology Inc, Boston, Massachusetts) [31]. A 2-way repeated measures MANOVA was used to investigate the main and interaction effects of wheel type and obstacle intensity on WBV values at each sensor location. Post-hoc tests were performed on specific pair-wise differences between the standard, LoopWheels and CLX wheels. Another post-hoc on pair-wise differences between intensity regions was also conducted. Comfort scores were assessed for normality. A one-way repeated measures ANOVA was used to investigate main effects of wheel type on comfort scores by surface type and overall. In the event that the comfort scores were not normally distributed, a non-parametric Kruskal Wallis test was performed. All analyses were computed with a 95% confidence level and a p-value of less than 0.05 indicates significant differences between wheel type. Effect size was estimated for each test with a partial Eta squared (small effect: 0.01, medium effect: 0.06, and large effect: 0.14).

2.5 Results

2.5.1 Participants:

A total of 24 MWUs (18 men and 6 women) with spinal cord injury/disorder participated in this study. Participants were on average 41.9 ± 10.6 years old, 170.9 ± 9.2 centimeters tall and weighed 83.4 ± 19.1 kilograms on average. Participants were on average 20 ± 13.5 years (a range of 2.4 to 45.9 years) post-injury.

2.5.2 RMS and VDV in Relation to HGCZ:

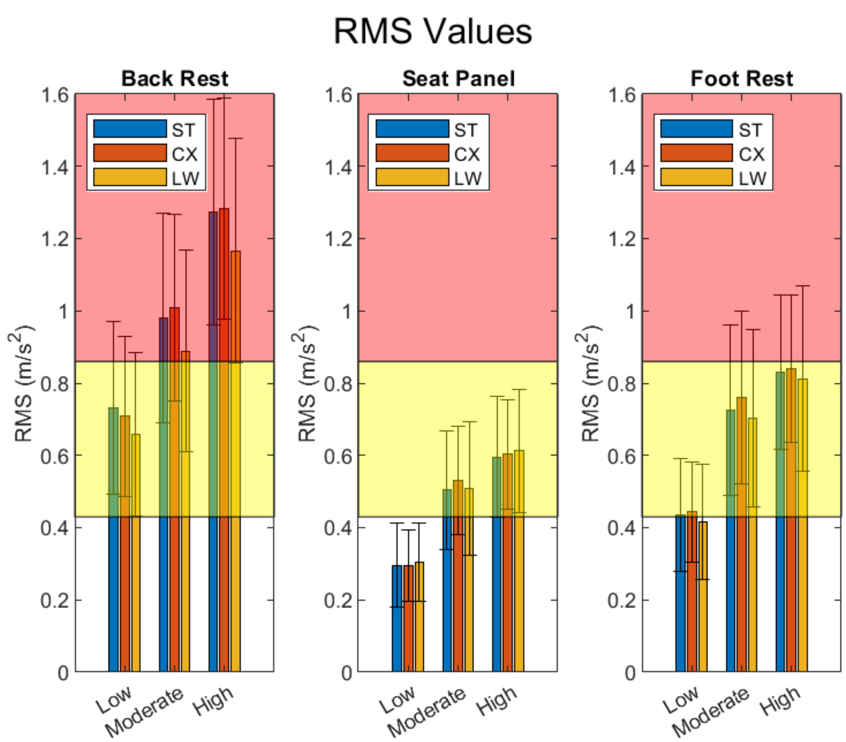


Figure 6: Average RMS values over all participants for each wheel, sensor location, and intensity region. The yellow region indicates the HGCZ defined from ISO 2631 for RMS (0.43 – 0.86). The red region is exceeding this zone. Error bars show standard deviation.

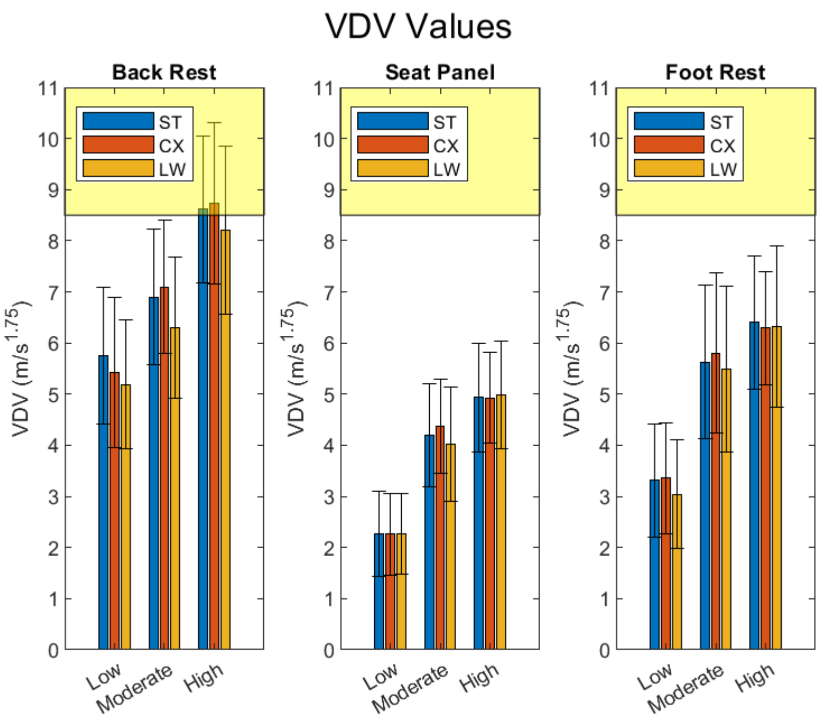


Figure 7: Average VDV values over all participants for each wheel, sensor location, and intensity region. The yellow region indicates the HGCZ defined from ISO 2631 for VDV (8.5 – 17). Error bars show standard deviation.

Crest factors were calculated for each trial, intensity region, and wheel type separately and averaged over the three trials. All calculated crest factors were above 9 (average crest factor across all participants, trials, and conditions was 15.9), indicating that several low-frequency, high amplitude shocks were observed. Therefore, both the RMS and VDV values were analyzed to evaluate the WBV.

2.5.3 RMS and VDV: Effects of Wheel Type and Intensity:

RMS values were significantly different across wheel type in the backrest and the footrest sensor positions (p-value < 0.001), whereas at the seat panel location there was no significant difference in RMS (p-value = 0.204). In general, the WBV values for the seat panel were the lowest out of all the locations for all wheel types; the backrest position had the highest WBV values (Figures 8 and 9). For VDV values, significant differences were found at the backrest (p-value < 0.05 and eta-squared=0.05) and not at the seat panel or footrest (p-value 0.58 and 0.17 respectively).

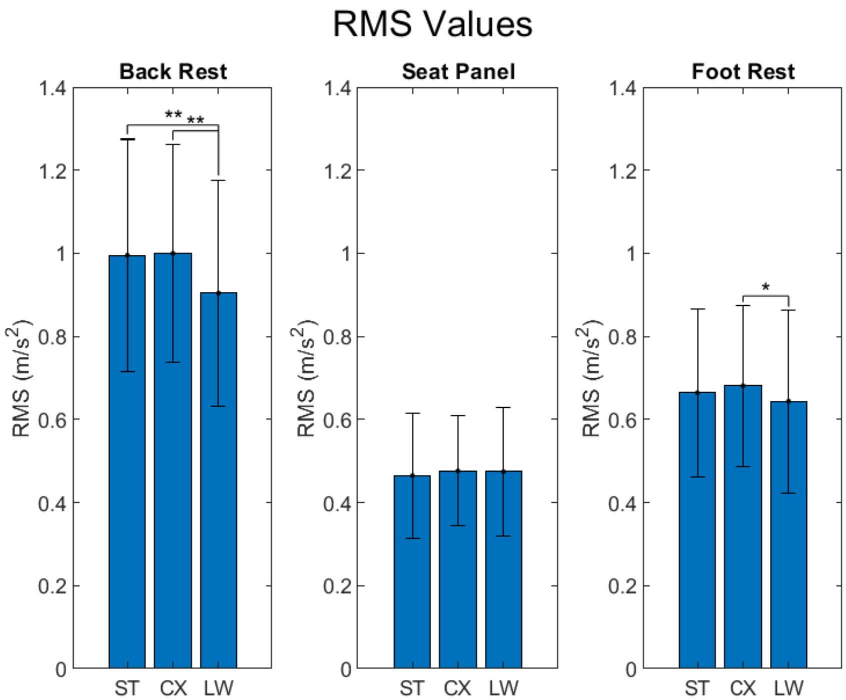


Figure 8: RMS mean values calculated for each wheel type at each sensor location. Error bars signify the standard deviation and statistical significant differences are reported by * and ** (p-value <0.05 and 0.001 respectively). (ST: standard, CX: CLX, and LW: Loopwheels)

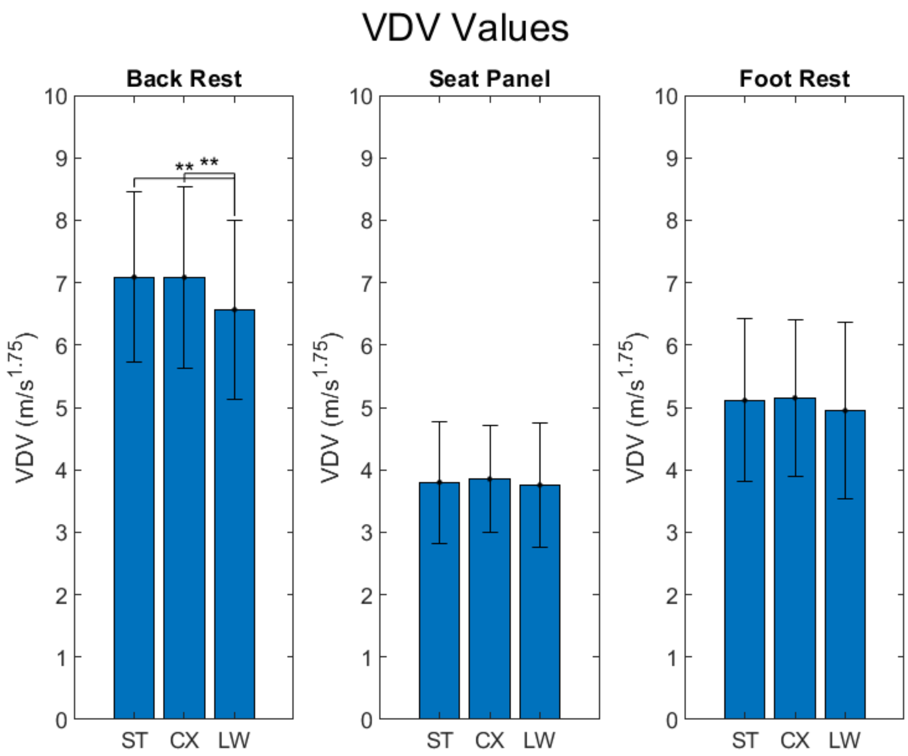


Figure 9: The VDV mean values calculated for each wheel type at each sensor location. Error bars show the standard deviation and statistically significant differences are reported by ** (p-value< 0.001). (ST: standard, CX: CLX, and LW: LoopWheels)

At the back rest the LoopWheels had significantly lower RMS and VDV than the CLX and standard wheel with a medium effect size ($p<0.005$ and eta-squared=0.08); no significant differences were found between the CLX and standard wheel. The RMS values at the footrest were significantly lower for the LoopWheels than CLX with a small effect size (p-value < 0.001 and eta-squared=0.01). RMS at the seat panel showed no significant difference between wheel types ($p=0.16$ and eta-squared=0.005). VDV at the seat panel and foot rest also showed no significant difference between wheel types ($p=0.6$, 0.2 and eta-squared=0.001, 0.005 respectively). Pairs of wheel types that showed significant differences were further analyzed to determine range of change between wheels. Percent changes between the standard spoke or CLX wheel and the

LoopWheels were calculated per subject. At the backrest and footrest, most participants experienced around a 10-20% decrease in RMS and VDV compared to standard and CLX wheels, with some experiencing as much as 40-50% decrease in WBV. A few participants experienced increased vibrations by around 10%, but the majority experienced a decrease in WBV when using the LoopWheels. Specific histograms can be found in Appendix C.

Both RMS and VDV showed significant differences across intensity regions (all p-values < 0.001). A post-hoc analysis revealed that both the high and medium intensity regions produced significantly higher WBV values than the low intensity region ($p<0.005$); high intensity region produced significantly higher WBV than the medium intensity region. This was the case for all sensor locations. There were no significant interaction effects between wheel type and intensity (all p-values > 0.2 across WBV values and sensor locations).

2.5.4 Rider Comfort:

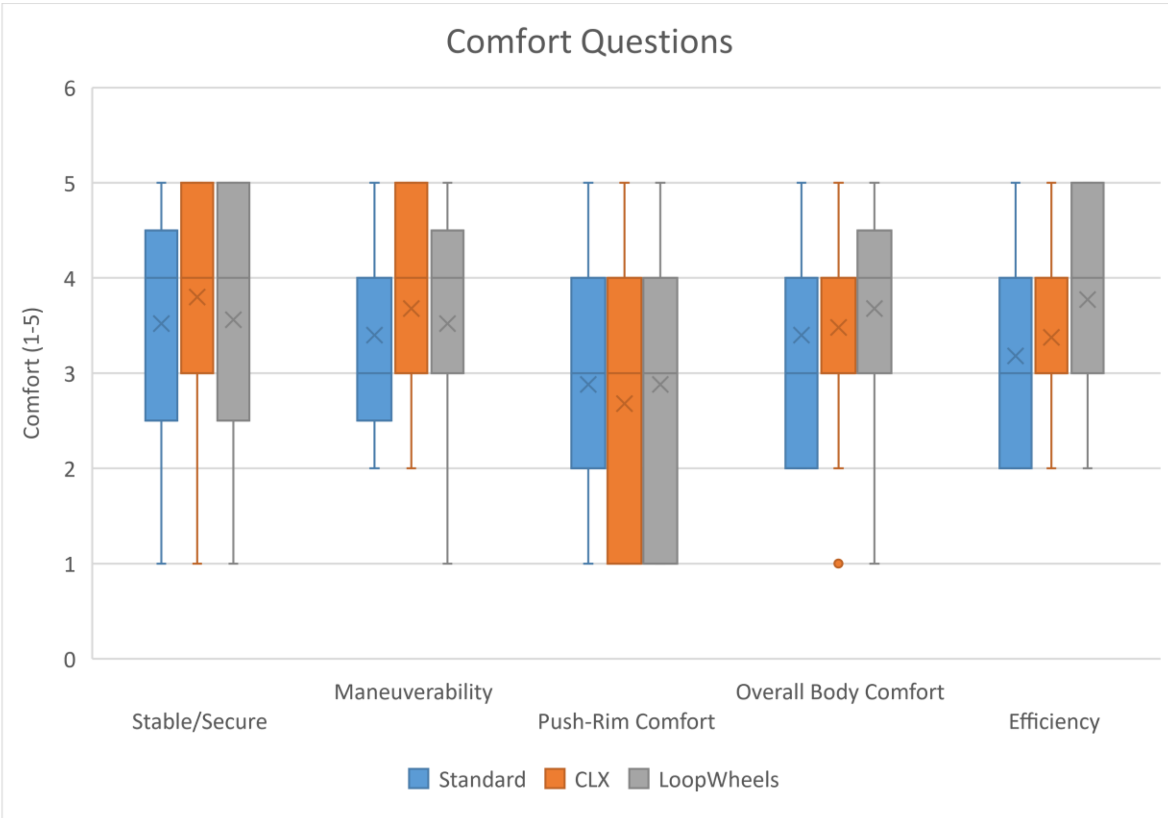


Figure 10: Median self-reported comfort scores for each surface and wheel type along with interquartile range.

Table 2: Median and mode of the overall experience questions for each wheel type.

Wheel Type	Statistic	Stable/Secure	Maneuverability	Push-Rim Comfort	Overall Body Comfort	Efficiency
Standard Spoke	Median	4.0	3.0	3.0	3.0	3.0
	Mode	4.0	3.0	3.0	4.0	2.0
CLX	Median	4.0	4.0	3.0	3.0	3.0
	Mode	4.0	4.0	1.0	3.0	3.0
LoopWheels	Median	4.0	4.0	3.0	4.0	4.0
	Mode	5.0	3.0	3.0	3.0	4.0

Results from the Shapiro-Wilks test for normality revealed that all the comfort scores were not normally distributed. Therefore, a non-parametric Kruskal Wallis test was performed. There were no significant differences in self-reported comfort for each surface and across all surfaces between wheel types (p-values > 0.6) (Figure 10).

No significant differences were found between wheel types for questions related to overall experience (p-values > 0.6). Values for pushrim comfort were the lowest out of all questions asked (2.78 averaged over all three wheel types) whereas all other questions showed above moderate experiences for stability, maneuverability, overall body comfort and efficiency.

2.6 Discussion

In-wheel suspension systems are a potential convenient solution to reducing the large amounts of WBV that MWUs are exposed to as they can be retrofitted to many existing wheelchair frames. In this study, we aimed to quantify to what extent a LoopWheels can reduce harmful vibrations and shocks in MWU when traversing sur-faces of varying roughness and when descending a curb. By involving experienced MWUs, we hoped to incorporate several elements that were not accounted for in other studies investigating the effect of suspension systems. In addition to involving experienced MWUs with spinal cord injury we also tested them in their own wheelchairs and setup which they were intimately familiar with. The results of this study suggest that LoopWheels Urban has the potential to lower WBV values when compared to the CLX and a standard wheel at the backrest and footrest. However, there was no differences at the seat and in perceived comfort or overall experience between the three wheels.

We found that nearly all conditions had RMS values within or exceeding the HGCZ. The exceptions were the low intensity region seat panel and footrest values. The VDV values were below the threshold for the HGCZ with the exception of the high intensity region for the backrest position. Furthermore, the average RMS values are within (seat panel: ~0.47 m/s2 and foot rest: ~0.65 m/s2) and exceed (back rest: ~0.94 m/s2) the HGCZ. However, almost all VDV values are below the HGCZ for all locations and wheel types. These results indicate that the vibrations (e.g. RMS) measured in this study could be very harmful to MWCs who traverse similar surfaces and obstacles in the real-world while the shocks may be of lessor concern. However, this is assuming that MWUs are exposed to these surfaces and obstacles for 4-8 hours per day which may or may not be the case. Community-based studies have found that MWUs only propel for an average of one hour a day [2, 7] while occupancy time is much longer. Furthermore, the maximum RMS value calculated across all wheel types, sensor locations, and surface intensities was around 1.2 m/s^2; indicating that MWUs would need to propel over the surfaces in the mobility course for around 2 hours before being exposed to hazardous levels of WBV. Our results for vibration and shock values are different than what was found previously in laboratory studies, where both vibration and shock were reported to be within or exceed the HGCZ [7, 14, 20, 22]. However, it should be noted that these studies investigated surfaces and obstacles in isolation, and did not report both RMS and VDV or WBV for multiple surfaces and obstacles in succession.

This study used a simulated road course approach (e.g. combined obstacle/surface type) to evaluate WBV [6] as opposed to a single surface or obstacle analysis where either a RMS or a VDV is determined based on the nature of the surface or obstacle [6]. Given that our course design consisted of sections of differing surfaces and obstacles, crest factors were calculated to determine if the basic evaluation method (RMS) was valid for describing the vibration effects. All in-tensity sections were found to be above the cut off threshold (15.86 > 9) implying that traversing the sections induced occasional or transient shocks in addition to the high frequency low magnitude vibrations that would be expected by some of the surfaces. These results are similar to other studies using a simulated road course where crest factors were high and both RMS and VDV were analyzed [4, 15, 32]. These shocks are likely coming from interfaces between surfaces, preparing before an obstacle, performing a wheelie, and/or moving abruptly in their chair and thus these aspects which are typical of ‘re-al-world’ propulsion may be important to consider in a WBV analysis with wheelchair users. However, since most all VDV values were under the HGCZ implies that the occasional or transient shocks are not enough to cause health concerns.

As expected, the RMS and VDV values increased with increasing intensity region providing a level of validity for our approach and measures. RMS and VDV values for the backrest and footrest showed significant differences for wheel type, but not at the seat panel. This may be due to the way the sensor needed to be attached to the participant’s wheelchair which differed chair to chair based on the model and from the method defined by ISO 2631. The ISO standard recommends that the accelerometer be placed under the seat cushion and securely fixed to the seat panel directly below the sacral joint of the MWU to minimize any artifact motion [11]. Other studies investigating vibration in MWC have either used the same wheelchair in a controlled environment [6, 14, 21, 22], or designed specific apparatuses to hold the sensor in place for each subject [4, 7]. In this study, we utilized the participant’s existing wheelchair to preserve their unique setup rather than provide a wheelchair for the MWU to use. The seat panel sensor was attached to part of the frame under the seat panel as close to directly under the sacral joint as possible. This method was also used in other vibration studies in MWUs [6, 20]. The seat panel position also produced the lowest levels of WBV compared to the backrest and the footrest, which is consistent with results from other studies investigating WBV at the seat and footrest [20].

The LoopWheels had significantly lower RMS values than the CLX and standard wheel at the backrest with a difference around 0.1 m/s2 for each. This indicates a 10% reduction in vibration exposure when using the LoopWheels. RMS was also significantly lower for the LoopWheels compared to the CLX at the footrest position by 0.05 m/s2 (5 %). Similarly, VDV values at the backrest were significantly lower for LoopWheels compared to CLX and standard wheel by about 7%, with differences of about 0.5 m/s1.75. This indicates that the LoopWheels was able to significantly reduce the amount of WBV exposure at the backrest and has some effect at decreasing RMS at the footrest. Partial eta-squared values from 0.05 to 0.08 indicate a medium effect size for the reduction in WBV when using LoopWheels. To put these results into perspective when considering time of exposure if a MWU occupies their wheelchair for an average of 13 hours a day [33], a 7-10% reduction in backrest and footrest RMS would reduce the amount of exposure to harmful WBV by around an hour, and potentially increase the amount of time a MWU can safely push before it becomes hazardous to health. This assumes again that MWUs would be experiencing RMS vibrations at similar levels to those measured in this study for the entire exposure period. Lastly, the results showed no statistically significant interaction effects between the wheel type and the intensity region meaning that the LoopWheels appears to reduce vibration and shock more than standard spoked and CLX wheels across all intensity region levels.

Our results are different from Misch et al., who reported a 12% increase in vibration (RMS) at the seat panel when using LoopWheels [20]. Differences could be explained by the different types of surfaces and the approaches used for analysis (e.g. simulated road course vs. single surface type) [20]. The nine surfaces used in this study may represent a broader range of the surfaces MWUs commonly propel over compared to the four tested in the previous study. Carpet is a very common household flooring and curb drops, while not routinely performed, are encountered in outdoor areas; both of these surfaces were not tested in previous studies evaluating WBV and in-wheel suspension systems. Furthermore, the main outcome measure reported by Misch et al. was

RMS and not VDV [20], meaning they were not able to determine the effect of suspension systems on shock with those specific surfaces. Another factor that could account for differences in results could be the human participants used in this study. Misch et al. utilized a robotic propulsion system which is unable to perform many actions a MWU would use when traversing various surfaces in their environment such as wheelieing and upper limb and trunk postural adjustments [20]). When wheelieing, or lifting the casters off the ground (e.g. a common practice used for curb descent and when navigating high roughness surfaces), the rear wheels should be able to more effectively absorb the vibration and shock. A future study is needed to validate this assumption.

This study also found that the CLX wheels appear to have no vibration reducing advantages over the standard wheels and produced shocks and vibrations significantly higher than the LoopWheels. This indicates that the claims made by Spinergy that the CLX wheels reduce vibrations due to the Zylon (PBO) spokes and the carbon fiber rim are not valid in these circumstances. The method of vibration attenuation in LoopWheels is more effective at reducing vibration than the novel material used for the spokes. However, the suspension system offered by LoopWheels adds additional weight to the system compared to the CLX. LoopWheels was about 0.5 and 1.0 kg heavier than the standard spoke and the CLX wheels respectively. This increased weight can make not only propulsion but other aspects of use more difficult, such as transportation, inclines, maintenance, etc. However, the extra weight may be necessary for effective vibration suppression. Interestingly, the Misch et al. study found that LoopWheels increased propulsion cost by 12-16% compared to a standard wheel when using the robotic propulsion system. So, while the LoopWheels in our study showed potential in reducing vibration and shock, there may be a trade-off for propulsion cost as is similar for other suspension systems found in wheelchairs and bikes [20]. It is a known effect in bicycles that suspension systems cause a decrease in efficiency and an increased feeling of weight [34, 35], so it would make sense that these effects translate to wheelchairs.

There was no significant difference in perceived comfort across wheel types (comfort scores of standard wheel: 7.88±2.53, CLX: 7.83±2.38, and LoopWheels: 8.19±2.10). The standard deviations were also fairly small, indicating that there is not much difference in perceived comfort between different participants. However, all the comfort values are high, reflecting that there might be possible ceiling effects preventing an accurate comparison of comfort. The comfort scores for the LoopWheels were trending towards higher than the standard wheel and CLX, which resembles what was seen in another study investigating comfort with wheelchair suspension systems [29]. No difference between comfort scores may also be beneficial, indicating that the Loopwheels despite being heavier and potentially energy absorbing had no adverse affect on the comfort scores.

2.7 Limitations and Future Directions

One of the main limitations in this study was that we were unable to measure exactly how much vibration was transmitted to the MWU instead of the wheelchair frame. All of the accelerometers in this study were fixed to various points on the wheelchair frame directly against the rigid surface. This means that several factors such as backrest supports and seat cushions could impact the amount of WBV the MWU actually experiences; a soft seat cushion could reduce vibrations at the seat, or a firmer backrest could amplify shocks. Previous studies have found that the material properties of the seat cushion and the postural stability of the back supports have significant effects on vertical vibration transmissibility due to changes in posture and muscle activity [4]. While the methods used in this study are similar to other studies investigating vibration and following the guidelines in ISO 2631, it is not fully representative of the WBV experienced by the MWU. Previous studies have utilized a bite-bar accelerometer along with one mounted to the seat panel to measure vibration transmissibility from the seat through the spine [4]. Similarly, another study used an accelerometer mounted onto a bicycle helmet [14]. Head mounted sensors introduce their own limitations such as delayed response to seat forces and increased motion during propulsion [4, 14]. Future studies should focus on measuring and monitoring the accelerations at these same positions but directly on the MWU to determine vibration transmissibility. Another limiting factor of this study was the use of standard aluminum ½" (1.27 cm) diameter circular pushrims to control the propulsion interface between wheels and participants. Many of the participants used ergonomic pushrims which have a larger surface area or additional rubber grips. These pushrims have be-come more prevalent among MWUs as they assist with gripping and have been shown to reduce pain in the wrists and arms by 80-85% [36]. The results from the comfort survey indicated that participants found the study pushrims not very comfortable (average 5-point Likert scale score of 2.78) which could have impacted their overall comfort scores. It is unclear how the in-wheel suspension systems impact pro-pulsion efficiency with human MWU and whether it shows a similar increase in energy cost as found when using a robotic propulsion system [20]. Future efforts should focus on analyzing the propulsion efficiency of in-wheel suspension systems with actual MWUs, and determine other factors related to real-world and long-term use (e.g. impact on transfers, transportability, durability, pain, fatigue, etc.).

2.8 Conclusions

MWUs are exposed to high levels of WBV throughout their lives; caused by propelling over various surfaces that induce shock and vibration with limited technology to reduce the effects of the environment. In-wheel suspension systems present a potential solution to reducing the WBV and are compatible with most wheelchair models on the market today. One particular in-wheel suspension system, the LoopWheels Urban, was able to significantly reduce RMS vibrations experienced by the wheelchair frame by 7-10% at the backrest and at the footrest. On the other hand, the Spinergy CLX wheel was not able to significantly reduce vibration. Additionally, both systems were unable to reduce WBV levels to below the HGCZ considering an average exposure time of 4-8 hours; in fact, levels of RMS for all wheel types across the majority of surfaces tested showed levels either within or exceeding the HGCZ. This means that the efforts to reduce WBV in MWU may need to be investigated further. Furthermore, there was no perceived difference in comfort experienced between wheel types.

3.0 Longitudinal Effects of In-Wheel Suspension Systems on Pain and Fatigue in Manual Wheelchair Users and Their Relationship to Sociodemographic and Activity Levels

3.1 Chapter Summary:

This study examined the effects of in-wheel suspension systems on neck/back pain, fatigue, participation and whole-body vibration (WBV) exposure among manual wheelchair users (MWUs). Twenty-four MWUs participated in a 12-week intervention using LoopWheels Urban suspension wheels. An accelerometer measured average distance, time during propulsion, and community levels of WBV for about two weeks midway into the intervention period. A battery of surveys were administered at baseline and post intervention, with the aim to determine changes in specific domains of pain and fatigue and community levels of participation. Results demonstrated a significant reduction in median fatigue scores (0.25 out of 3), reduction in median neck pain scores (1.5 out of 10), median weekly pain interference decrease (2.5 out of 10), and a 1.5 decrease in median pain problems (all $p < 0.05$). Furthermore, participants reported a smoother ride over bumpy surfaces and improved shock absorption compared to traditional wheels. Moreover, community WBV levels fell below hazardous thresholds defined by International Standard Organization (ISO) 2631 for an average daily propulsion time of two hours. MWUs were exposed to an average of 0.29 ± 0.14 m/s^2 root mean square (RMS) vibration and 8.67 ± 4.07 m/s^1.75 vibration dose value (VDV) while propelling in the community, representing a 35% reduction in vibration and a 50% reduction in shock compared to previous community based studies. In-wheel suspension systems show promise in improving health outcomes and reducing WBV exposure for MWUs.

3.2 Introduction:

The natural and built environment causes significant levels of whole body vibration and shock (WBV) to manual wheelchair users (MWUs) [3, 6]. Surfaces such as sidewalks, gravel, pavement, curb-cuts, grass, and thresholds are often unavoidable and propelling over them can cause significant levels of vibration and shock. When exposed to high levels of WBV for long periods of time, MWUs often face increased prevalence of back and neck pain and increased levels of fatigue [4, 7, 13, 18, 27].

Research has demonstrated a significant association between WBV exposure and adverse health outcomes, particularly in the context of back pain and secondary injuries. A study found that full-time wheelchair users, who spend an average of 11 to 13 hours per day in their chairs, are at heightened risk for WBV-induced secondary complications such as back pain, disc degeneration, and musculoskeletal disorders [7, 33]. Moreover, neck and back pain prevalence rates among wheelchair users surpass those of the general population, with increased activities, poor propulsion technique, and poor body posture exacerbating the discomfort [6, 7, 27]. Moreover, prolonged seating, awkward postures, and vibration exposure have been identified as significant occupational risk factors contributing to back pain [6, 7, 27]. Additionally, individuals with SCI often exhibit spinal deformities, posterior pelvic tilt, and lack of trunk control, predisposing them to unnatural body positions that increase susceptibility to WBV-related injuries [33]. Notably, wheelchair users may spend more than 16 hours a day seated in their wheelchairs, heightening their vulnerability to WBV-induced health issues [7]. Efforts to mitigate WBV exposure through various seating systems and cushion characteristics have yielded mixed results. While some studies have explored the efficacy of different seating systems and accessories, such as seat cushions, in minimizing vibration transmission, some systems have been found to amplify harmful vibrations due to cyclic vibratoins, particularly those occurring within the frequency range most detrimental to human health [4]. Despite these challenges, wheelchair users continue to face elevated levels of WBV that exceed safety standards established by the ISO [11, 12]. These values are defined as the Health Guidance Caution Zone (HGCZ) in ISO document 2631. These thresholds are determined by both the amount of accelerations experienced and the length of time of exposure. As the duration of vibration increases, the threshold for hazardous levels of WBV decrease. A standard exposure time of 8 hours provides a hazardous range of 0.43 to 0.86 m/s^2 for vibration and 8.5 to 17 m/s^1.75 for shocks [10-12]. HGCZs are provided for other ranges of exposure time as well, with lower exposure time raising the limit considered hazardous [11].Values experienced within these ranges are considered to be unsafe and anything exceeding these levels are known to be very detrimental to health.

Several laboratory based studies have been conducted that focus on measuring WBV in MWUs and have found that the surfaces they are they are frequently exposed to are associated with dangerous levels of vibration and shock [6, 7, 10, 21, 22]. Community levels of WBV in MWUs have also been found to exceed the limits set by ISO 2631. Garcia-Mendez et al. found that during a two week period of measuring the vibrations and shock to the wheelchair, users were exposed to 0.83 ± 0.17 m/s^2 and 0.55 ± 0.13 m/s^2 RMS vibration and 17.26 ± 3.23 m/s^1.75 and 12.06 ± 2.37 m/s^1.75 VDV shock at the seat and backrest respectively while occupying their wheelchair for around 13.07 ± 3.85 hours a day [7]. However, the occupancy time considered periods of motion (e.g. propulsion) and non-motion. Thus, the findings from this previous study may not be fully representative of the actual exposure time to vibration and shock which is more likely to occur when the MWU is only in motion (e.g. pushing, being pushed, on riding in a motor vehicle). Since many surfaces and obstacles that cause WBV are often unavoidable, every effort must be made to reduce the amount of vibration and shock that is transmitted to the MWU.

Wheelchair suspension systems have been developed to address the discomfort caused by prolonged wheelchair riding and WBV [15, 16, 21]. These systems typically include coil springs attached to the wheelchair frame or single spring-damper units supporting the seat. While studies have shown that these suspension systems can reduce some aspects of shock and vibration, particularly when traversing obstacles, they have limitations [16, 21]. For example, they are limited in their ability to absorb shock and vibration due to their fixed location and orientation within the wheelchair frame [21]. Suspension casters can significantly reduce peak accelerations but have mixed results compared to traditional frame designs in reducing harmful vibrations [4, 20]. Misch et al. found that FrogLegs suspension casters increased vibrations exposure by nearly 100% over different surfaces (tile, brick, sidewalk, and grates) [20]. Similarly, front caster suspension systems are constrained by their design and may not effectively mitigate vibrations [4, 20]. In contrast, in-wheel suspension systems offer a novel approach by providing a 360-degree capability to absorb shocks and vibrations during wheelchair use [23, 24, 29]. To our knowledge, in-wheel suspension systems have undergone limited research in the lab and community. Laboratory tests conducted with a robotic wheelchair propulsion system suggest that LoopWheels in-wheel suspension systems increase WBV exposure by around 12% over brick and grates at the seat panel [20]. A separate in-wheel suspension wheel, SoftWheels, was also tested and showed marginal improvements over tile surface; however, both LoopWheels and Softwheels increased propulstoin costs by 12-17% over select surfaces, like sidewalk and brick, compared to a default configuration [20]. A qualitative study conducted with novice MWUs found that participants reported a smoother ride and increased confidence when using the Softwheel suspension wheels over standard wheels

[29]. It is still unknown how in-wheel suspension systems affect community levels of WBV exposure or how they will impact the lives of experienced MWUs. A study conducted to examine the factors that limited MWUs participation in community and recreational events reported that wheelchairs were ranked as the number one barrier preventing them from participating in activities [19]. Using a wheelchair was a greater limitation than their physical impairment, pain, fatigue, or other complications when considering standard wheelchairs are not designed to traverse over rough or uneven terrains commonly found in the community (unpaved roads, pot-holes, curbs). In-wheel suspension systems seek to allow MWUs to more easily propel over these surfaces, thereby improving independence and psychological health [19]. Ideally, successful implementation of in-wheel suspension should decrease the avoidance of community participation by increasing the number and type of obstacles that can be traversed. Therefore, this study aimed to:

1. Determine the amount of WBV exposure to wheelchair users in their real-world environment when using in-wheel suspension and compare to the limits of exposure as defined by the ISO 2631-1 standard on Mechanical Vibration and Shock (4). We hypothesized that the daily WBV exposure to wheelchair users in their real-world environment when using in wheel suspension will be within the safe acceptable levels as described in the ISO-2631 standard.

2. Evaluate the effects of in-wheel suspension on pain, fatigue, and participation in MWUs with three months of routine use. We hypothesized that MWUs would report having significantly less neck pain, back pain and fatigue after the trial period of using in-wheel suspension wheels measured by a number of psychosocial surveys. We also expected MWUs to report a higher number of encounters and fewer numbers of avoidances with environmental features in the community after the trial period of using in-wheel suspension wheels as measured by the Environmental Aspects of Mobility Questionnaire (EAMQ).

3. Explore the association between perceived improvements in quality of life after the trial period as measured using the Patient's Global Impression of Change Scale (PGIC) and participant characteristics (socio-demographics and activity levels) to gain insight into sub-groups that may benefit most from in wheel suspension.

3.3 Materials and Methods:

3.3.1 Participants:

This study was approved by the University of Pittsburgh Internal Review Board (ID: 20020199). Twenty-six participants were recruited into the study through research registries, local disability community gatherings and events, and word of mouth. The inclusion criteria for participation were: (1) be at least 18 years old, (2) have a spinal cord injury diagnosis at least one year prior to the start of the study, (3) use a manual wheelchair as primary mode of mobility, (4) use 24 inch diameter wheels with quick-release axle pins, (5) no other suspension elements in frame or castors, (6) weigh under 265 pounds (maximum weight limit of the suspension wheels), (7) self-reported moderate chronic neck and/or back pain, and (8) be proficient in verbal and written English. Individuals were excluded from the study if they had: (1) a history of fractures or dislocations in the upper extremity which has not fully recovered, (2) upper limb pain that interferes with propulsion, (3) severe cognitive limitations, or (4) current or recent history of pressure sores.

3.3.2 Study Procedures:

Once the participant completed the informed consent documents, demographic and baseline pain and fatigue surveys were administered. Since pain and fatigue are subjective measures, a number of surveys were used to ensure we captured all aspects and perspectives. Surveys administered included: Brief Pain Inventory, Daily Questionnaire for Pain and Fatigue, Environmental Aspects of Mobility, International SCI Pain Basic Data Set, Neck Disability Index, Wheelchair User Shoulder Pain Index, Fatigue Severity Scale, and the Iowa Fatigue Scale [17, 37-45].

The Modified Brief Pain Inventory (BPI) (Appendix D) is a tool designed to assess the impact of pain on physical and psychosocial functioning. Its purpose is to evaluate pain intensity and interference across various domains of life, providing insights into the extent to which pain affects daily activities and quality of life. Using a numerical rating scale ranging from 0 (no pain/interference) to 10 (extreme pain/interference), the BPI assigns separate scores that reflect the level of pain intensity and interference, with higher scores indicating more significant pain interference [43]. Sub scores are calculated by averaging the four intensity questions and seven interference questions independently to obtain a score out of 10 for both. A separate validation study investigating the efficacy of the BPI for low back pain found that the minimal detectable change (MDC) for these scores were 2.57 and 2.34 for pain and interference respectively [46].

The Daily Questionnaire for Pain and Fatigue (Appendix E) serves to assess physical fatigue and pain levels over the past 24 hours in wheelchair users during various activities and on different surfaces. It combines two individual surveys, the Numerical Rating Scale for pain (NRS) and the Fatigability Index for individuals with SCI [37, 47]. Its purpose is to track daily fluctuations in fatigue and pain levels in relation to wheelchair use, offering insights into the impact of wheelchair use on fatigue and pain and helping tailor interventions accordingly. Scores range from 0 to 3 for four fatigue questions (0 = no fatigue, 3 = extreme fatigue) and 0 to 10 for pain (0 = no pain, 10 = worst pain imaginable) in different locations on the body [37]. Pain is scored separately in individual areas as well as the average of the neck and back areas with a maximum score of 10 and an MDC of 2. Fatigue is calculated as the average of four fatigue questions with a maximum score of 3 [48, 49].

The Environmental Aspects of Mobility Questionnaire (EAMQ) (Appendix F) is designed to assess encounters and avoidances of environmental barriers during mobility. Its purpose is to evaluate the environmental challenges faced by individuals with mobility impairments, identifying factors affecting mobility. Using a five-point scale to assess the frequency of encounters and avoidances, the EAMQ assigns scores separately to encounters and avoidance of obstacles by summing the total score for each domain [45]. Encounters is scored from a range of 21 to 105 and avoidance is scored from 15 to 75, with higher scores indicating greater encounters or increased avoidance respectively.

The International SCI Pain Basic Data Set (Appendix G) version 2.0 is a standardized scale for characterizing and reporting pain in individuals with spinal cord injury (SCI). It covers pain types, locations, intensities, and frequencies, facilitating consistent pain assessment across studies. Its clinical significance lies in its ability to classify pain types, subtypes, and locations, aiding in pain management and treatment evaluation. Number of pain problems is measured on a scale from zero to more than five. Numerical responses on a 0 to 10 scale are used, with lower numbers indicating lower pain interference and higher numbers suggesting higher pain interference [40]. Pain interference is measured by averaging three questions rated from 0 (no interference) to 10 (extreme interference) on pain interfering with activities, mood, and sleep.

The Neck Disability Index (NDI) (Appendix H) measures neck-related disability and pain levels, comprising items related to daily activities, pain, and concentration. Its purpose is to assess the impact of neck pain on daily functioning, providing insight into the degree of neck-related disability. Scores range from 0 to 5 for each item, with higher scores indicating greater neck-related disability [42]. The final score is calculated as a percentage of the maximum score (50). A percentage score of 40% or greater indicates severe disability.

The Wheelchair User's Shoulder Pain Index (WUSPI) (Appendix I) is a self-report instrument that measures shoulder pain intensity during functional activities in wheelchair users. Its purpose is to assess the severity of shoulder pain and its impact on daily tasks, providing information on the level of shoulder pain experienced during specific activities. Using a 15-item questionnaire, the WUSPI assigns scores from 0 to 10, with lower scores indicating lower shoulder pain intensity during activities [50]. The final score is the sum of all questions and has a maximum score of 150.

The Fatigue Severity Scale (Appendix J) asks 9 questions related to feelings of fatigue. Questions include motivation being lower when fatigued, exercise inducing fatigue, being easily fatigued, frequent problems from being fatigued, fatigue being the most debilitating symptom, and fatigue interfering with work, family, and social life. Each question is rated on a 7 point Likert scale from 1 (strongly disagree) to 7 (strongly agree) [39]. The final score is summed and on a range from 9 to 63. A threshold of 36 was established, with lower values indicating no fatigue and high values indicating the possibility of fatigue.

The Iowa Fatigue Scale (Appendix L) is a scale to quantify fatigue felt over the past month. Eleven questions surround aspects of feeling worn out, energetic, slowed down by thinking, trouble concentrating, feeling drowsy, having low output, having trouble with memory, being unable to concentrate well, feeling rested, and feelings of being busy. Each question is scored on a scale from 1 (not at all agree) to 5 (extremely agree) with each statement. Total scoring is calculated using a custom equation based on the positive or negative tone of each question. Overall fatigue scores from 30 to 39 indicate the presence of fatigue and scores from 40 to 55 indicate severe fatigue. Fatigue sub scores for cognitive, overall fatigue, energy, and productivity can also be calculated by analyzing specific sets of questions [8].

After completing the surveys participants were then fitted with a pair of LoopWheels Urban (LoopWheels Urban, Jelly Products Ltd, Nottinghamshire, United Kingdom) (Figure 3) onto their existing wheelchair with care to ensure the correct length of the axle pin and working wheel locks [23]. LoopWheels, a design encompassing three “C” shaped carbon fiber springs, were selected for the longitudinal portion of the study due to their demonstrated reduction of vibration and shock during in-lab assessment with MWUs (see Chapter 2). LoopWheels lowered vibrations by 10% at the backrest compared to the standard spoked and Spinergy CLX wheels and by 7% at the footrest compared to the CLX. It also reduced shocks by 7% at the backrest compared to the standard wheel and CLX. All wheels provided to participants were outfitted with Marathon Plus tires and filled to the recommended 125 psi. Instructions on how to ensure optimal pressure and fill up the tires were also provided.

The type of pushrim provided was based on the participant’s preference and what they were currently using. Pushrim options included standard ½” (1.27 cm) diameter anodized pushrims or an ergonomic style: Gecko (CarboLife, CarboLife technologies, Dresden, Germany), Tetra (CarboLife, CarboLife technologies, Dresden, Germany), supplied by LoopWheels, or Natural Fit (Quickie, SouthwestMedical LLC, Phoenix, Arizona). Individuals were instructed to use the LoopWheels as they would their original wheels for 12 weeks. Around six weeks into the intervention, two Verisense IMU sensors (VerisenseIMU, Verisense Health, Cambridge, MA), containing tri-axial accelerometers and gyroscopes were placed onto the spoke of the LoopWheel (Figure 11B) and on the wheelchair frame under the seat panel (Figure 11A). Sensors were attached with strong adhesive tape to secure and protect from any moisture in the environment. Sensors were set to record data at 52 Hz for both accelerometer and gyroscope functions. Sampling rate was chosen based on preliminary testing on battery life of sensors lasting around 10 days. A higher sampling rate decreased the length of battery life greatly. The sensor placed under the seat panel was used to collect the accelerations experienced by the wheelchair frame and was used to calculate amount of exposure to vibration and shocks. The sensor on the wheel monitored the rotations of the wheel and was used to calculate distance, speed, and bouts of propulsion. For all participants, the sensor under the seat panel was oriented so the Z axis was in the vertical, and the sensor on the wheel was oriented so the X axis was normal to the wheel (lateral) (Figure 11). After about two weeks, the sensors were removed from the frame and wheel. Sensors recorded data passively and data collection started once registering it with the cloud service provided by Verisense. Accelerometer and gyroscope data were stored in on-board memory with space up to 512 MB. After sensors were retrieved from participants, they were connected to the base station (an Android tablet, provided by Verisense) wirelessly and synced with the cloud service. Once paired, the sensor would send each packet of data to the base station, which would then upload it to the cloud service. Finally, all sensor data was retrieved from the cloud service through their online platform.

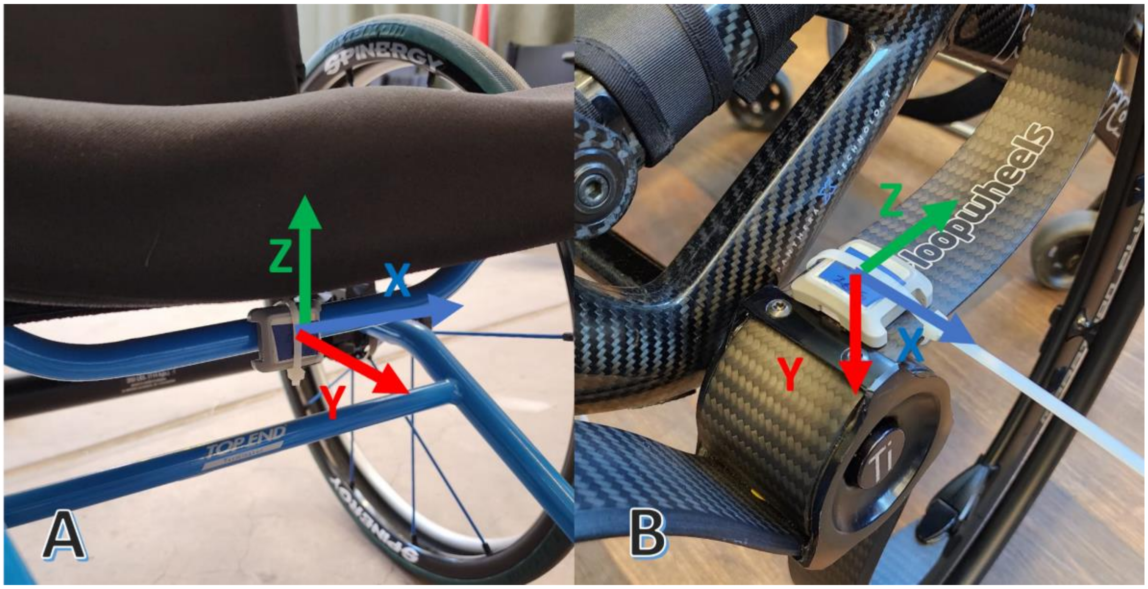


Figure 11: Sensor placement for the accelerometer (A) and the gyroscope (B). Coordinate systems are also shown.

At the end of the 12 week intervention, participants completed the same pain and fatigue surveys as from the baseline visit. Additionally, the Patient Global Impressions of Change scale was administered to determine the overall impressions of using the suspension wheels [51]. The Patient Global Impression of Change (PGIC) (Appendix K) scale is a self-reported measure commonly used in clinical research to assess patients' perceptions of change in their condition over time. It typically involves patients rating their overall improvement or deterioration since the start of treatment on a scale ranging from "very much worse" to "very much improved." The PGIC scale provides valuable insights into patients' subjective experiences and satisfaction with their treatment or intervention.

In addition to the PGIC scale, an exit survey inquired about participants' general experiences with the suspension wheels, focusing on functionality, likes and dislikes, maintenance or repair requirements, and their likelihood of recommending the wheels to others. The effects of in-wheel suspension on spasticity symptoms was also recorded. Participants were given the opportunity to provide detailed feedback on any limitations they encountered while using the wheels, as well as aspects they appreciated most and least about the devices. Scores for the two PGIC questions and the three Exit Survey questions were counted for each response option, averaged over all participants, and examined along each questions' scale.

3.3.3 Data Processing and Analysis:

Mobility characteristics were calculated from gyroscope data; a threshold of 1.5 degrees per second was used to determine a period of propulsion to eliminate the effect of sensor drift. This cut off value was chosen based on the sensor specifications provided by Verisense. Angular velocity data (from the gyroscope) was converted from degrees to radians and multiplied by the radius of the wheel in meters (0.30 meters) to obtain linear displacement. Distance and number of minutes in motion per hour in a day were summed and averaged over all the days. Accelerometer data were processed using the method outlined in ISO 2631 to filter and extract the appropriate frequency bandwidth for vibration and shock [11, 12]. Accelerations in the vertical direction were determined based on the direction of gravity. Raw vertical acceleration values were passed through a 4th order Butterworth bandpass filter of frequencies within 0.4 to 100 Hz and a powerband filter to extract the first third octave of frequency domain data. Final frequency weighted acceleration in the vertical direction was separated into periods of propulsion and no propulsion (obtained from gyroscope data analysis). Daily average vibration RMS (Equation 12) and shock VDV (Equation 13) values were calculated over all periods of propulsion for the entire day and then averaged over all days the sensor recorded data. RMS is a measure of the overall magnitude of vibration over time, providing an average value that reflects the intensity of vibration experienced by the body. It considers the amplitude of vibration across the entire frequency spectrum and is commonly used to assess the potential physiological effects of vibration exposure on the human body. Clinically, RMS values are indicative of the general vibration intensity experienced by individuals and are useful for understanding the overall impact on musculoskeletal health and comfort. On the other hand, VDV takes into account both the magnitude and duration of vibration, providing a measure that emphasizes the transient or shock-like nature of vibration exposure. It specifically focuses on the repetitive nature of high-intensity vibration events and is often used to assess the potential for musculoskeletal fatigue and injury due to repeated shocks. Clinically, VDV values are valuable for understanding the cumulative effects of vibration exposure over time, especially in occupations or activities where individuals are subjected to frequent and repetitive shocks or jolts. Average daily RMS and VDV for all participants were compared against the HGCZ outlined by ISO 2631. All data processing was completed in Matlab (Matlab 2023a, MathWorks Ltd, Natick, Massachusetts, USA).

Ordinal socio-demographic data was converted to dummy variables for each possible response. For example, the level of urbanization was categorized into four groups (rural, small town, small city, big city) where levels were defined from 0 to 3 respectively. Similar principles were applied to American Spinal Injury Association Impairment (ASIA) Scale for scale A through D. Gender was coded as ordinal. To test for effects of demographics on community mobility outcomes, a Spearman's Correlation test was conducted on continuous scalar data (age, BMI, and time using a wheelchair). Linear regression on ordinal and significant scalar variables was performed to determine relationship between demographics and mobility outcomes.

Survey data was first tested for normality on differences between baseline and post using the Shapiro-Wilk test for normality. In the event of significance reported by the test, the specified survey was analyzed using the Wilcoxon Ranked Sign test. All other surveys demonstrating normality underwent a T-test between baseline and post intervention. PGIC and Exit survey results were analyzed with raw scores (average and standard deviation) and recoded as a one (1) for improved outcome and zero (0) for stable or worsened outcome. Logistic regression on bivariate outcomes with demographic variables was performed to determine effect of demographics on experience with using the in-wheel suspension. All statistical significance was calculated at the 95% confidence level and were performed in IBM SPSS 29 (SPSS, IBM, Armonk, New York, USA).

3.4 Results:

A total of 26 MWUs were recruited for this study, but only 24 completed all phases and surveys. One subject dropped out due to the weight of the wheels causing too much interference with daily life (transfers to cars, pushing uphill, etc.) and one dropped out because the equivalent pushrim provided began pinching their fingers between the pushrim and the wheelrim during propulsion causing discomfort. Community accelerometer and gyroscope data from the Verisense sensors was unable to be obtained for four participants. This was due to technical issues with with the cloud service and the transfer of data from the sensor to the cloud. The remaining participants' demographics are shown in Table 3.

Table 3: Participant demographics for longitudinal and community WBV. Averages, counts, and percentages for categorical varibales are shown.

Demographics		Mean(STD)/Count(%)
Community WBV Sensor N		20
Pain and Fatigue Survey N		24
Age		43± 11.6
Height		1.72 ± 0.1
Weight		83.4 ± 18.0
Time using WC		18.73 ± 14.9
Gender	Male	18 (75%)
	Female	6 (25%)
ASIA Scale	A	12 (50%)
	B	4 (17%)
	C	2 (8%)
	D	6 (25%)
Area of Residence	Big City	9 (38%)
	Small City	10 (42%)
	Small Town	4 (17%)
	Rural	1 (4%)

Table 3 shows the average mobility characteristics of the MWUs (N=20) from the community sensor data collection. The Verisense sensors recorded data for an average of 10.14 ± 3.4 days. The MWUs traveled 1208.11 ± 969.5 meters per day with 114.18 ± 80.45 minutes of time in motion. Average daily WBV levels during bouts of propulsion was 0.29 ± 0.14 m/s^2 for RMS and 8.67 ± 4.07 m/s^1.75 for VDV (Table 3). The value obtained for the community levels of VDV during propulsion was higher than the threshold for hazardous levels of shock (8.5 m/s^1.75). However, this value is defined by an exposure period of eight hours. Based on the community propulsion data, the MWUs only propelled for an average of around two hours a day. For a vibration exposure time of two hours, ISO 2631, the HGCZ is defined as RMS: 1.0 – 1.6 m/s^2 and VDV: 19.8 – 31.6 $m/s^{1.75}$ [11, 12]. This implies that the WBV is well under the threshold for hazardous exposure to MWUs when taking into account the propulsion time.

The average propulsion RMS value of 0.29 m/s^2 and no propulsion RMS value of 0.13 m/s^2 indicate that the RMS vibration levels are both below the hazardous threshold of 0.43 m/s^2 for 8 hours of exposure and 1.0 m/s^2 for 2 hours of exposure. This suggests that the vibrations experienced during both propulsion and no propulsion phases of wheelchair use are generally within safe limits based on RMS measurements. However, the average propulsion VDV value of 8.67 m/s^1.75 and no propulsion VDV value of 9.92 m/s^1.75 are slightly higher than the lower end of the VDV threshold of 8.5 m/s^1.75 for 8 hours of exposure (but do not exceed the high end of the range of 14 m/s^2) and well below the limit for 2 hours (19.8 m/s^1.75).

Table 4: Average daily mobility characteristics of the MWUs. Distance, time in propulsion, RMS and VDV values are shown.

		Average	Std Dev
Average Days Monitored	Days	10.14	3.40
Average Daily Distance	meters	1208.11	969.47
Average Daily Propulsion Time	minutes	114.18	80.45
Average Propulsion RMS	m/s^2	0.29	0.16
Average Propulsion VDV	m/s^1.75	8.67	4.83
Average No Propulsion RMS	m/s^2	0.13	0.07
Average No Propulsion VDV	m/s^1.75	9.92	3.69

Spearman's correlation (Table 5) and linear regression (Table 6) results showed that most demographics were not significantly related to mobility characteristics, with the exception of BMI and area of residence. MWUs with a higher BMI were significantly more likely to have lower daily distance traveled compared to those with a lower BMI (p-value = 0.04). Area of residence also showed significant relationships, with MWUs in small towns significantly more likely to have a higher daily displacement than their counterparts who live in cities (p-value = 0.05). Similarly, MWUs in small cities and rural areas are significantly more likely to experience larger VDV shocks when not propelling than those residing in big cities (p-value = 0.04 and 0.03 respectively).

Table 5: Results from the Spearman's Correlation test for demographcis and community WBV. Correlation coefficient is shown. No variables were determiend to be significnat.

Mobility Variable	Avg Daily Distance	Avg Daily Time	Avg Propolusion RMS	Avg Propulsion VDV	Avg No Propulsion RMS	Avg No Propulsion VDV
Age (years):	-0.04	0.11	0.34	0.26	0.26	0.27
bmi	-0.31	-0.13	-0.22	-0.24	-0.33	-0.21
Time Using WC Months	0.17	0.08	0.28	0.27	0.26	0.29

Table 6: Linear regression results on demographics and commuity WBV. Beta coefficients are shown, with significant differences denoted by * and ** (p < 0.05 and 0.01 respectively.)

Mobility Variable	Avg Daily Displacement	Avg Daily Time	Avg Propolusion RMS	Avg Propulsion VDV	Avg No Propulsion RMS	Avg No Propulsion VDV
Linear Regression Output	Beta	Beta	Beta	Beta	Beta	Beta
(Constant)	4183.05 *	226.17	0.49	15.54	0.28	13.67 *
Age (years):	-9.27	1.00	0.00	0.06	0.00	0.04

Gender=Male	Ref	Ref	Ref	Ref	Ref	Ref
Gender=Female	-1180.42	-50.46	-0.17	-5.32	-0.11	-4.69
bmi	-98.87 *	-5.23	-0.01	-0.39	-0.01	-0.29
ASIA=A	Ref	Ref	Ref	Ref	Ref	Ref
ASIA=B	143.37	-19.31	0.03	1.02	0.04	1.35
ASIA=C	615.32	35.09	0.06	3.34	0.00	1.65
ASIA=D	110.53	-26.49	0.12	2.61	-0.01	1.11
Time Using WC Months	-0.17	-0.07	0.00	0.00	0.00	0.00
AreaofResidence =A big city	Ref	Ref	Ref	Ref	Ref	Ref
AreaofResidence =A town/small city	25.64	-2.12	0.01	2.93	0.04	4.12 *
AreaofResidence =A small town/village	1192.29 *	94.30	0.02	3.26	0.06	3.29
AreaofResidence =Rural area	1524.53	83.60	0.19	10.20	0.13	11.03 *

Results from the Shapiro-Wilks test for normality conducted on the difference between baseline and post intervention for the pain and fatigue surveys indicated that many of the measures were not normally distributed. Table 7 displays the median, mean, and standard deviation of each of the pain and fatigue survey outcomes from baseline and post intervention. In general, all average and median scores from the baseline period are below cutoffs for detrimental health implications. Baseline measures from the Fatiguability Index imply that there is some fatigue present (1.74± 0.70 out of 3. Fatigue Severity Scale scores, while not above the threshold of 36, are close to suggesting the presence of fatigue (30.08±14.7 out of 63). All other surveys reported fairly low pain and fatigue scores. Baseline encounters and avoidance showed no differences pre-post intervention.

Table 7: Baseline and post intervention pain and fatigue survey scores. Minimum, median, maximum, and interquartile range for all outcomes are shown. Ranges and maximum values for each survey and scale are indicated where appropriate.

		Baseline					Post				
		Min	25%	Median	75%	Max	Min	25%	Median	75%	Max
SCI Pain Basic	Pain Interference (max 10)	0.0	1.4	3.0	6.0	9.0	0.0	0.0	0.5	4.2	7.0
	Num Pain Problems (up to 5)	0.0	1.3	3.0	4.0	5.0	0.0	0.0	1.5	3.0	5.0
Brief Pain	Brief Pain Intensity (max 10)	0.0	0.4	2.8	5.1	8.3	0.0	1.3	2.8	4.4	8.3
	Brief Pain Interference (max 10)	0.0	1.0	2.4	4.8	8.9	0.0	1.1	3.1	5.7	8.6
Daily Pain Fatigue	Fatiguability Index (max 3)	0.0	1.3	1.8	2.3	3.0	0.8	1.0	1.5	1.8	2.5
	NRS Average (max 10)	0.0	0.0	2.2	3.9	8.0	0.0	0.0	0.8	2.3	7.0
	NRS U. Back (max 10)	0.0	0.0	1.5	3.8	8.0	0.0	0.0	0.0	4.8	7.0
	NRS L. Back (max 10)	0.0	0.0	0.0	4.8	10.0	0.0	0.0	0.0	2.8	7.0
	NRS Neck (max 10)	0.0	0.0	1.5	4.8	8.0	0.0	0.0	0.0	2.8	7.0
Fatige Severity Scale	FSS [9-63] (>36)	9.0	15.8	27.5	42.8	56.0	9.0	15.8	33.0	41.5	54.0
Iowa Fatigue	Total Fatigue [30-39],[40-55]	11.0	18.3	24.5	35.5	45.0	15.0	20.0	26.5	32.8	44.0
	Cognitive Fatigue (max 20)	4.0	5.0	8.0	11.0	18.0	4.0	5.0	8.0	10.8	14.0
	Fatigue (max 10)	1.0	2.0	4.0	6.0	7.0	2.0	3.3	5.0	5.8	10.0
	Energy (max 15)	3.0	7.0	8.0	12.3	15.0	5.0	8.0	9.0	11.0	15.0
	Productivity (max 10)	2.0	3.0	4.5	6.0	9.0	2.0	3.3	4.5	6.5	8.0
Neck Disability	NDI (>40%)	0.0	4.0	22.0	39.5	54.0	0.0	4.0	23.0	38.0	52.0
WUSPI	WUSPI (max 150)	0.0	0.5	8.0	17.2	59.4	0.0	0.0	5.7	22.1	63.0
EAMQ	Encounters [21-105]	47.0	60.8	66.5	75.5	88.0	46.0	53.3	64.0	74.8	90.0
	Avoidance [15-75]	19.0	26.3	35.5	43.8	61.0	19.0	27.5	36.0	47.0	56.0
	A/E	0.3	0.4	0.5	0.7	1.3	0.2	0.4	0.6	0.7	1.2

Since normality testing revealed that many of the surveys were not normally distributed, a Wilcoxon Rank Sign test was performed on all surveys along with a Student's T-test. Since the results of the T-test did not differ ($p < 0.05$ for significant surveys), only results from the non-parametric Wilcoxon Rank Sign test are shown. Table 8 displays the median, mean, and standard deviation of the difference between baseline and post values. A negative difference indicates that the measure decreased from baseline to post, whereas a positive difference indicates an increase in the measure. Z statistic and p-values were used to determine direction of significance, since all tests were performed on a two-tail assessment. Notable significant differences were found in two surveys: SCI Pain Basic Dataset and the Daily Questionnaire for Pain and Fatigue. Specific survey outcomes include: pain interference over a week, number of pain problems, Fatiguability Index and the NRS score for the neck area. Al four outcomes showed significant decreases in pain and fatigue measures. SCI Pain Basic Dataset survey showed a decrease of median pain scores of 2.50 out of a maximum of 10 in pain interference (around 25% median decrease) over the week and a median decrease of 1.50 pain problems after three months of using the suspension wheels. Fatigability Index median decreased by 0.25 out of a total score of 3 for fatigue (around 8% median decrease). Median pain intensity in the neck area showed to decrease significantly by 1.50 out of 10 after using the suspension wheels (around 15% decrease). All other survey outcomes showed no significant differences between baseline and post intervention.

Table 8: Results from the Wilcoxon Rank Sign test. Pairwise median, mean, and standard deviation of the differenecs in surey scores (post – baseline) are shown. Cohen's D, Z statistic and two-tail p-value denote estimated effect size and significant differences with * (p < 0.05).

		Paired Differences				Wilcoxon Rank Sign		Effect Size
		Median	Percentage Difference (% max score)	Mean	(Std. Deviation)	Z	p-value (2-tail)	Cohen's D
SCI Pain Basic	Pain Interference (max 10)	-2.50	-25	-1.51	(2.87)	-2.42	0.02*	0.49
	Num Pain Problems (up to 5)	-1.50		-1.04	(1.71)	-2.69	0.01*	0.55
Brief Pain	Brief Pain Intensity (max 10)	0.00	0	-0.29	(2.13)	-0.85	0.40	0.17
	Brief Pain Interference (max 10)	0.79	8	0.23	(2.41)	-1.29	0.20	0.26
Daily Pain Fatigue	Fatiguability Index (max 3)	-0.25	-8	-0.30	(0.59)	-2.74	0.01*	0.56
	NRS Average (max 10)	-1.33	-13	-0.69	(2.60)	-1.76	0.08	0.36
	NRS U. Back (max 10)	-1.50	-15	-0.50	(1.44)	-1.75	0.08	0.36
	NRS L. Back (max 10)	0.00	0	-0.42	(3.12)	-0.63	0.53	0.13
	NRS Neck (max 10)	-1.50	-15	-1.46	(2.38)	-2.60	0.01*	0.53
Fatigue Severity Scale	FSS [9-63] (>36)	5.50	9	0.25	(11.38)	-0.32	0.75	0.07
Iowa Fatigue	Total Fatigue [30-39],[40-55]	2.00	4	1.13	(5.08)	-1.08	0.28	0.22
	Cognitive Fatigue (max 20)	0.00	0	-0.42	(1.82)	-0.66	0.51	0.13
	Fatigue (max 10)	1.00	10	0.67	(1.95)	-1.53	0.13	0.31
	Energy (max 15)	1.00	7	0.63	(2.04)	-1.58	0.11	0.32
	Productivity (max 10)	0.00	0	0.25	(1.73)	-0.64	0.52	0.13
Neck Disability	NDI (>40%)	1.00	10	0.33	(11.18)	-0.53	0.60	0.11
WUSPI	WUSPI (max 150)	-2.36	-2	0.40	(14.91)	-0.61	0.54	0.12
EAMQ	Encounters [21-105]	-2.50	-2	-2.13	(7.99)	-1.45	0.15	0.30
	Avoidance [15-75]	0.50	1	1.08	(9.92)	-0.18	0.85	0.04
	A/E	0.05	5	0.04	(0.18)	-0.18	0.29	0.04

PGIC and Exit survey results are shown in Table 9. After the 12-week suspension wheel intervention, participants reported overall improvement in quality of life (average 2.83 ± 1.09 on a scale from 1,better to 7,worse) and a positive degree of change (average 3.81±1.51 on a scale from 0,improved to 10, worse) on the PGIC survey. The maximum score for each outcome measure was 5 and 6 respectively, which are both below the maximum range of the scale (7 and 10 respectively). Histograms of these two outcomes shows most people rated Quality of Life (Figure 12) change at a 3 (N=8) and Degree of Change (Figure 13) at a 4 (N=6). Exit survey results show overall positive outcomes, with nearly no change in spasticity symptoms (0.42±0.88 on a scale from -3,problematic to 3,helpful), no feelings of being limited (1.54±0.98 on a scale from 1,not at all to 5,extremely), and a high recommendation of LoopWheels (3.25±1.33 on a scale from 1,not at all to 5,extremely).

Table 9: PGIC and Exit Survey results. Mean, standard deviation, max, and min for each question are shown. Scale for questions are also noted.

Survey	Outcome	Median	Mode	Max	Min
PGIC	Quality of Life Change (1 - better, 7 - worse)	3.0	3.0	5.0	1.0
	Degrees of Change (0 - improved, 10 - worse)	4.0	3.0	6.0	0.0
Exit	Affect Spacticity (-3 - problemtic, 3 - helpful)	0.0	0.0	3.0	0.0
	Feeling Limited (1 - not at al, 5 - extremely)	1.0	1.0	4.0	1.0
	Recommend Suspension Wheels (1 - not at al, 5 - extremely)	3.0	3.0	5.0	1.0

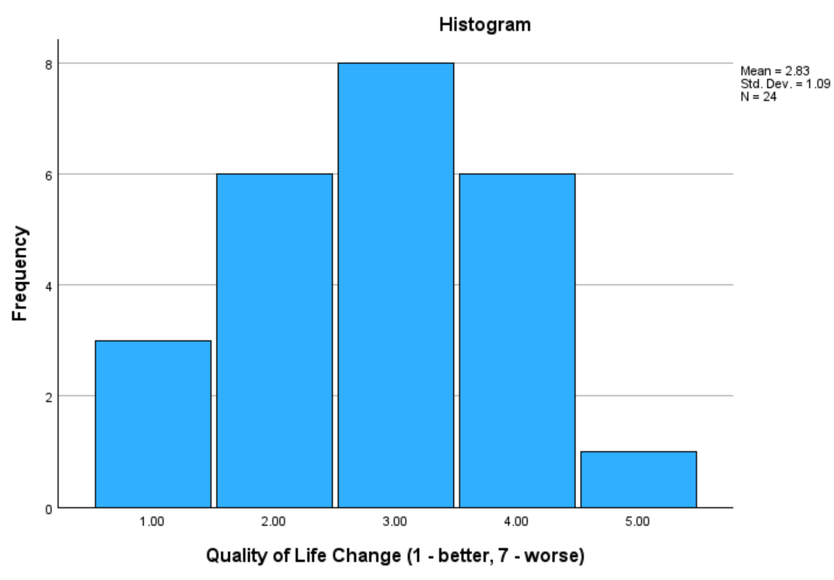


Figure 12: Histogram for Quality of Life change.

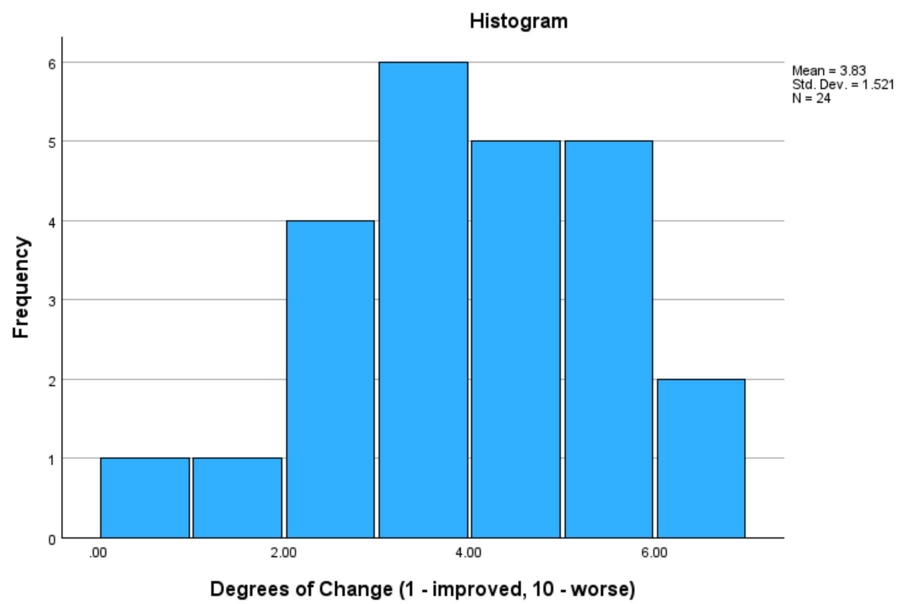


Figure 13: Histogram for overall degrees of change.

Spearman's correlations between PGIC, Exit survey, and demographics showed no significant correlations. Furthermore, ordinal regression on raw scores and logistic regression on whether the outcome improved or not with the demographic and baseline survey scores showed no significant relationships. For the 20 participants with mobility characteristic data, another Spearman's correlation was performed on PGIC, Exit survey, and community mobility characteristics to determine potential relationships (Table 10). MWUs who traveled short distances and for less time were significantly more likely to report higher degrees of change (p-value 0.04 and 0.02 for distance and time respectively) and were significantly more likely to recommend the suspension wheels to other MWUs (0.03 for distance). Similarly, MWUs who experience less community levels of WBV during both propulsion and no propulsion were significantly more likely to recommend the suspension wheels (all p-values < 0.02).

Table 10: Spearman's Correlation test for PGIC, Exit survey, and mobility characteristics. Correlation coefficients are shown with significant differences denoted by * (p < 0.05).

Mobility Characterisitics	QoL	Degree of Change	Spacticity	Limited	Recommend
'Average Daily Distance	-0.208	-.462*	0.061	0.17	-.478*
'Average Daily Time'	-0.114	-.532*	-0.109	0.284	-0.265
Average Propulsion RMS	-0.341	-0.34	-0.376	-0.095	-.531*
Average Propulsion VDV	-0.378	-0.288	-0.231	-0.076	-.513*
Average No Propulsion RMS	-.511*	-0.427	-0.352	-0.095	-.619**
Average No Propulsion VDV	-0.36	-0.218	-0.158	-0.132	-.513*

3.5 Discussion:

In this study, we aimed to characterize the impact of in-wheel suspension systems on community levels of WBV exposure, pain, fatigue, participation and overall quality of life. Previous studies on in suspension systems have been limited to the laboratory environment and do not represent the views of experienced wheelchair users [6, 10, 14, 20-22]. The results from this study revealed some interesting findings when compared to other literature. The mobility characteristics show an average of around two hours of propulsion from the community sensor data. This time is higher than from other community based studies which found that MWUs only propel for around one hour on average [2, 7, 33]. The MWUs traveled for approximately 1208.11± 969.47 meters, which was lower than values reported previously of around 2500 ± 1000 meters per day [2, 7, 33]. These results indicate that the MWUs in this study exhibit different activity profiles from the ones measured in the previous studies. Participants of two of the prior studies were para-athletes who participated in the National Veterans Wheelchair Games, so it is expected that their average mobility each day would be higher [2, 7]. Another factor that could explain the decreased mobility may be the time the data was collected. While from a similar demographic population, results from Sonenblum et al. found that MWUs travel for around 1953 ± 1525 meters, which is still larger then what was found in this study [33]. However, data collected by Sonenblum et al. was before the COVID-19 pandemic [33]. After the pandemic, there was a notable shift in society towards more work from home and less outdoor time in general. It’s possible these trends may be even more pronounced among a wheelchair user population given the many barriers associated with transportation and the increase in home-based services that are available now (e.g. food delivery).

When propelling, the participants were exposed to less community vibration than previously found which may be explained in part by the less distance covered per day. Garcia-Mendez et al. found that in the community MWUs are exposed to around 0.83 m/s^2 RMS vibration and 17.26 m/s^1.75 VDV shock; suspension systems were found to have no significant effect on levels of vibration [7]. The same study found that when compared to the HGCZ from ISO 2631, MWU are always either within or exceed the thresholds for hazardous exposure to vibration and shock [7]. In this study, we found that MWUs who use in-wheel suspension are exposed to on average 0.29 ± 0.14 m/s^2 RMS vibration and 8.67 ± 4.07 m/s^1.75, around a 35% reduction in vibration and a 50% reduction in shock. However the Garcia-Mendez et al. study used the combined vertical and anterior-posterior accelerations to measure WBV at the seat panel which are known to inflate WBV values compared to using just the vertical direction [7]. We found the in-wheel suspension showed vibration exposure to be below the HGCZ (0.29 << 0.43 m/s2) and the level of shock exposure to be just within the zone (8.67 > 8.5 m/S^1.75) for an 8 hour exposure time [11, 12]. When compared to Garcia-Mendez et al. which also tested suspension frame systems, it seems in-wheel suspension could offer a better solution to decreasing the amount of WBV exposure to below hazardous levels.

The comparison of vibration levels measured in this study against established thresholds provides valuable insights into the potential risks associated with wheelchair use. It's important to contextualize these findings within the framework of exposure time, as the specified thresholds are typically based on longer exposure durations, typically ranging from 4 to 8 hours. In this study, participants were assessed over a period of around 2 hours, which represents a shorter exposure time compared to the standard 4-8 hour window used for establishing vibration thresholds. It's worth noting that vibration exposure thresholds tend to increase with shorter exposure times, reflecting the body's reduced capacity to tolerate vibrations over shorter durations. Despite the shorter exposure time in this study, the observed vibration levels, particularly the RMS values, remained below the specified hazardous thresholds. This suggests that even during a relatively brief period of wheelchair use, the RMS vibrations experienced by participants were within safe limits. However, the propulsion and no propulsion VDV values remained within the range that indicates hazardous levels of shock for 8 hours of exposure. It should also be noted that the VDV values are well below the upper threshold for the HGCZ (14 m/s^1.75), suggesting the high potential for LoopWheels to attenuate shocks in the community. Given the shorter exposure duration in this study, the findings suggest that wheelchair users may be exposed to acceptable levels of vibration during typical daily activities. From the two hour propulsion time found in this study, the levels for hazardous WBV increase to 1.0 – 1.6 m/s^2 for RMS and 19.8 – 31.6 $m/s^{1.75}$ for VDV [11, 12], indicating that the community WBV is well below the threshold defined by HGCZ. Furthermore, a community RMS of 0.29 m/s^2 indicates that MWUs should be able to propel for up to 14-16 hours over community surfaces before being exposed to harmful WBV.

However, caution should be exercised, especially considering the potential for longer exposure times in certain scenarios (e.g. hiking, traveling, recreation, unavoidable terrain). Prolonged or repeated exposure to vibrations, even at levels below established thresholds, could still pose risks to musculoskeletal health and comfort over time. Therefore, while the results provide reassurance regarding vibration exposure during short-term wheelchair use, continued monitoring and adherence to ergonomic guidelines are recommended to minimize the risk of musculoskeletal discomfort and injury among wheelchair users over the long term. Additionally, future research with extended time of use and monitoring would provide further insights into the cumulative effects of vibration on musculoskeletal health and well-being.

Considering these findings, it's essential to interpret the results cautiously. Although the RMS vibrations are below the hazardous threshold, the slightly elevated VDV values during both propulsion and no propulsion phases warrant attention. While these values may not pose an immediate risk, prolonged or repeated exposure to shocks at these levels could potentially contribute to musculoskeletal discomfort or injury among wheelchair users. Furthermore, the variability in vibration levels among study participants, as indicated by the standard deviations, highlights the importance of individualized assessments and interventions to mitigate the potential risks associated with vibration exposure during wheelchair use.

In summary, while the RMS vibrations remain within safe limits, the slightly elevated VDV values highlight the need for continued monitoring and intervention to ensure the musculoskeletal health and well-being of wheelchair users, particularly during prolonged or frequent wheelchair use. These findings provide insights into the daily activities and associated vibration exposure experienced by wheelchair users. Furthermore, they contribute to our understanding of the potential health risks associated with wheelchair propulsion and stationary periods, particularly in relation to whole-body vibration exposure. Further analysis comparing these results to established health guidance caution zones, such as those outlined in ISO 2631, can help assess the potential risk of adverse health effects among wheelchair users.

The large standard deviations observed in the mean differences between baseline and post-survey results indicate substantial variability among participants in their responses to the interventions. This variability suggests that each participant had a unique experience with the intervention, leading to diverse outcomes in terms of pain, fatigue, and other measures. For instance, in the Fatigability Index, the median difference in average fatigue felt per day was 0.25 out of a maximum of 3 with a standard deviation of 0.59. Median NRS pain scores for the neck region decreased by 1.5, but had a higher standard deviation at 2.38 out of 10. Similarly, in the SCI Pain Basic survey, the median difference in weekly pain interference was 2.5 out of 10 with a standard deviation of 2.87 and the median decrease in number of pain problems was 1.5 with a standard deviation of 1.7. However, none of these differences were comparable to the minimal detectable change described by each of the surveys. This suggests that while, on average, there was a significant reduction in fatigue and pain interference post-intervention, the individual experiences varied widely. Some participants may have experienced a substantial decrease in pain and fatigue, while others may have experienced minimal change or even an increase in levels. The high variability in individual responses suggests that the effects of LoopWheels on pain and fatigue are highly situational and dependent on individual factors such as baseline health status, level of disability, activity levels, and personal preferences. Therefore, while the overall findings may show statistically significant differences, it is important to consider the individual experiences and tailor interventions accordingly. Cohen's D estimates for surveys with significant differences indicate that there is a medium effect based on the results from this study (Cohen's D > 0.5).

The lack of statistically significant findings in certain surveys, despite observing notable mean differences in others, exemplifies the importance of considering individual variability and preferences in assessing the efficacy of LoopWheels in alleviating pain and fatigue among wheelchair users. For surveys such as the Brief Pain Inventory (BPI), Neck Disability Index (NDI), and Fatigue Severity Scale, where the mean differences between baseline and post-survey results did not reach statistical significance, the individualistic approach may offer insights into the lack of significant findings. It's conceivable that certain participants responded differently to the intervention based on their unique physiological and psychological profiles. For example, individuals with higher baseline pain or fatigue levels may have a more noticeable difference in pain and/or fatigue if the wheels were helpful, whereas an individual with lower baseline scores would be less likely to notice any affect the in-wheel suspension has. Furthermore, MWUs may require more time or additional interventions to experience significant improvements. Similarly, differences in wheelchair propulsion techniques, environmental factors, and daily activities could have contributed to variations in outcomes among participants. Furthermore, the subjective nature of pain and fatigue assessment tools introduces inherent variability in responses. Participants' perceptions of pain and fatigue may be influenced by factors beyond the physical effects of the intervention, such as mood, stress levels, and expectations. Further research may be needed to explore the factors influencing individual responses to LoopWheels and their effects on pain and fatigue outcomes. Additionally, the short amount of testing time in this study (~12 weeks) may not have been enough time for full acclimation to new technology. Therefore, a longer duration of using the in-wheel suspension may be necessary for the MWU to fully feel the effect.

The lack of significant findings in certain surveys does not necessarily negate the potential benefits of suspension wheels but rather highlights the need for a nuanced understanding of individual experiences and responses to the intervention. With the community mobility characteristics, linear regression revealed that individuals with a higher BMI were more likely to travel less distance each day. MWUs who travel less distance each day were also more likely to report a positive degree of change (from the PGIC) and were much more likely to recommend the suspension wheels (Exit survey). MWUs who had less community WBV were also more likely to recommend the suspension wheels. The coupled results suggest that heavier, less active MWUs are reporting more benefit from the in-wheel suspension. One reason for this trend could be due the increased propulsion cost for the in-wheel suspension found by Misch et al. and the increased weight of the wheels [20]. It’s possible that MWUs who are traveling more per day on average need to expend greater amounts of energy to go the same distance, thus decreasing the overall reported benefits of the wheels. On the other hand, MWUs who travel less will notice a much smaller change in energy cost, thereby increasing the perceived benefits of the in-wheel suspension. Another factor for why individuals with higher BMI report more perceived benefit could be that the LoopWheels are more responsive at higher loads. As a spring-dampener based system with a set spring constant based on a max load of 120 kg, it's possible that LoopWheels has better responsiveness when impacting an obstacle with heavier individuals in the wheelchair. This increased responsiveness may be creating a more comfortable and smoother ride for heavier MWUs, whereas someone with a lower BMI still feels a more bumpy ride. Tuning the spring-dampening elements to the weight of the individual may be something to consider for future versions of the wheel. Future research should continue to explore subgroups of participants who may benefit most from LoopWheels and investigate additional factors that may influence treatment outcomes. Adopting a personalized approach to intervention delivery, tailored to individual needs and preferences, may enhance the effectiveness of in-wheel suspension in improving pain and fatigue management among wheelchair users.

These findings of below HGCZ limits of WBV exposure also support the results from the pain and fatigue surveys. After a three-month intervention period of using the in-wheel suspension, MWUs reported around an 8% decrease in neck pain and around 8-25% decrease in perceived fatigue. As there have not been previous studies looking at pain and fatigue outcomes after other suspension methods, it is difficult to say if in-wheel suspension offers a more optimal solution to reducing pain and fatigue in MWUs. Additionally, we did not obtain community levels of WBV with the participants before the LoopWheels intervention, so it is difficult to fully associate the decreased pain and fatigue with the in-wheel suspension. However, findings also support results of the PGIC and the qualitative feedback (verbal and written) from the participants at the end of the study. Participants reported an overall improvement in quality of life and were likely to recommend the in-wheel suspension to other MWUs after using the wheels for 12 weeks. We found that this was significantly positively correlated with increased BMI (driven mostly by increased weight), indicating that heavier individuals are gaining more benefit from the wheels. Since the LoopWheels come in different stiffnesses based on the MWU's weight, this is an important factor to investigate. On a closer analysis, we found that BMI was not significantly correlated with baseline pain and fatigue levels, but it was significantly positively correlated with a decrease in both pain and fatigue (p-values < 0.05 and Pearson's coefficient > 0.44). This is further evidence that the in-wheel suspension wheels may be more effective for heavier individuals. LoopWheels recommends the maximum load for the Regular Spring wheels to be 120 kg, with a softer and a stiffer spring for lighter and heavier individuals respectively. This result may imply that the wheels should be manufactured for more discrete weight classes, as the current range of weights covered by the standard stiffness may be too wide.

Results indicate that individuals who live in more urban areas were more likely to recommend the wheel, meaning that lifestyle and environment play a large role in experience with in-wheel suspension. MWUs in more urban areas are more likely to travel over sidewalks, curb-cuts, gravel, etc. than those that live in more suburban or rural environments. Interestingly, this goes against the correlations we see in the wheelchair mobility characteristics of the participants. We found that MWUs who travel more (both in distance and time per day) were significantly less likely to find an improvement in quality of life and were less likely to recommend the wheels. However, this correlation is also highly dependent on many other factors such as weather, time of year, and other life events. The times of intervention for the participants ranged over all parts of the year and the mobility characteristics were calculated based on a limited number of days in the community (around 10.14 ± 3.4 days on average). These factors could influence the level of activity each participant had.

Lastly, when we look at the qualitive feedback from the participants after using the LoopWheels, we can see some common themes forming between the likes and dislikes. Similar to what was reported in other studies, MWUs found that the in-wheel suspension offered significant improvement to the number and impact of bumps and vibrations experienced in the community [29]. Specific comments included "Seemed to roll over rocks and bumps more smoothly" and "Providing impact absorption on going down curbs". Further benefits found were in improvements in pain: "My shoulder and neck hurt less I was less squirmy in my chair" and "They made my shoulder pain noticeably less frequent". Furthermore, participants particularly enjoyed the look and style of the LoopWheels. The most commonly used word used by the MWUs for the dislikes was "heavy". Individuals noted that the increased weight of the wheels made it significantly more difficult to push long distances and up inclines. One user noted that they were having trouble using the accessible ramp for their van, something they need to do very regularly. Further complaints included: increased difficulty of transfers and in width of the wheelchair. The wheels have a wider profile than traditional wheels, and some users reported increased collisions and that it was harder to enter certain narrow door frames. These results indicate that the suspension systems do make a noticeable difference in bumps and uneven terrain, but the physical weight and dimensions introduce additional barriers to overcome. This, along with the health benefits found for certain demographics, signify that the potential for in-wheel suspension to improve quality of life and health exists, but further refinements and modifications are necessary to overcome these new additional challenges.

3.6 Limitations and Future Directions:

There were a number of challenges faced during this study involving equipment, hardware and software. The first set of LoopWheels that were sent out to participants experienced a manufacturing defect; the bolts meant to hold the carbon fiber springs were not fitted properly and become loose in several instances. While this did not lead to any injury among the participants of this study, the effect on the overall experience and quality of the wheels was impacted. Additionally, it took some time to repair the wheels for the participants, so for some of the participants the intervention period was not continuous. In the future, considerations should be made to ensure manufacturing quality and maintenance policies should be in place when failures occur. Secondly, the Verisense sensors used for monitoring the community levels of WBV exposure and mobility characteristics faced some challenges. The typical use of these sensors is for able-bodied individual activity tracking and were repurposed to collect daily mobility and vibrations. The process of setting up the sensors for long term collection did not allow for our team to assess the presence and quality of the data being collected when in use. Future studies should investigate different sensors or use custom made data-loggers to ensure accurate long-term collection.

3.7 Conclusions:

In this study we aimed to investigate and characterize the impact of in-wheel suspension systems on daily neck/back pain, fatigue, and community levels of WBV exposure. We found that MWUs who use LoopWheels Urban suspension wheels for around 12 weeks report a significant improvement in daily pain and fatigue experienced. Participants specifically reported that the wheels improved performance over bumpy surfaces and absorbed shock better than traditional wheels. Furthermore, it was found that WBV levels were below the hazardous levels defined by ISO 2631. Specifically, MWUs with higher BMI and lower daily mobility characteristics were more likely to note positive degrees of change and were more likely to recommend the wheels. The increased weight and width of the wheels introduced additional barriers to quality of life for MWUs. Therefore, further research is necessary to determine more optimal methods to implement in-wheel suspension.

4.0 Characterizing the Rolling Resistance and Impact of Deformation on In-Wheel Suspension Systems in Manual Wheelchairs.

4.1 Chapter Summary:

Manual wheelchair users frequently endure discomfort and pain due to whole body vibration (WBV) caused by navigating uneven terrain. In-wheel suspension systems offer a potential remedy, yet their impact on rolling resistance (RR) and propulsion mechanics remains poorly understood. This study compares an in-wheel suspension system, LoopWheels Urban to a Spinergy CLX and standard spoked wheel to elucidate their comparative effects. Component-level analysis using a drum-based testing apparatus revealed that LoopWheels had significantly higher RR compared to standard wheels, demonstrating 118% and 44% increases on linoleum and carpet surfaces, respectively. Conversely, the CLX wheel displayed 53% higher RR on linoleum and 13% lower RR on carpet relative to standard wheels. LoopWheels demonstrated greater deformation (0.27 inches) compared to both standard and CLX wheels under static loading conditions. LoopWheels demonstrated greater fore-aft amplitude (64% larger) and vertical amplitude (225% larger) for oscillations compared to standard wheels. Moreover, it showed a subtle phase shift in acceleration profiles suggests a more dynamic behavior of the LoopWheels' wheel hub relative to the rim during propulsion compared to the other two wheels. The increased oscillations during propulsion may help explain the perception among manual wheelchair users that LoopWheels impedes their propulsion efficiency, despite their smoother ride and reduction of vibrations and shocks. By shedding light on the complex interplay between in-wheel suspension, RR, propulsion

mechanics, and user perception, this study contributes insight into the impact of this technology on meeting the overall mobility needs of manual wheelchair users.

4.2 Introduction:

Around 66% manual wheelchair users (MWUs) report experiencing neck and back pain and discomfort that can often impact their ability to lead independent lives after beginning to use a wheelchair [2, 13, 18, 52]. This pain and discomfort is linked to whole body vibration (WBV) which is caused by the rough terrain and uneven surfaces in the MWU's environment [4, 6, 7]. Methods to reduce WBV include various suspension elements such as shocks embedded within the wheelchair frame, suspension castors, and in-wheel suspension systems; however, the former two options have shown to have non-significant and, in some cases, detrimental effects [16, 20-22]. In-wheel suspension systems offer an adaptable solution for MWUs as they are able to retrofit onto any existing model frame with quick-release axle wheels. In the first study of this dissertation, in-wheel suspension wheels were compared against a standard spoked wheel and a Spinergy CLX wheel (Chapter 2.0). LoopWheels Urban (LoopWheels, Jelly Products Ltd, Nottinghamshire, United Kingdom), is composed of an aluminum rim connected to a central hub with three C-shaped carbon fiber springs [23]. These springs are designed to absorb and deflect the axle of the wheelchair in the event of large vibrations or shocks and to provide a smoother more comfortable ride over rough and uneven terrain. The LoopWheels in the prior study reduced WBV and shocks at the backrest and footrest by 7-10% compared to standard wheels (Section 2.5.3). The CLX, on the other hand, showed no significant advantage in reducing WBV values. Furthermore, using in-wheel suspension was found to improve user comfort for novice MWUs, providing an overall smoother ride and fewer bumps transmitted to the user [29]. Experienced MWUs, however, reported no significant difference in comfort, stability, or maneuverability (Chapter 2.0).

User feedback from 24 experienced MWUs using LoopWheels for 12 weeks indicated that while the Loopwheels reduce the amount of vibration and shock felt, the wheels were heavy and often absorbed the energy of a push, preventing the MWU from traveling as far with each push as normal (Chapter 3.0). This suggests that the suspension wheels have lower propulsion efficiency compared to standard wheels. This result is further supported by research with a robotic wheelchair that found LoopWheels increased propulsion cost by 12-16% over different surfaces [20].

However, previous studies on propulsion efficiency in experienced MWUs have found several key human related factors that have a significant effect on efficiency that the robotic wheelchair study fails to account for. These include: joint accelerations, joint range of motion, stroke patterns, technique, experience, baseline strength, and pushrim geometries [53-56]. Research shows that two of the most impactful factors on efficiency are propulsion pattern and rolling resistance [57-60]. Using a semi-circular stroke pattern as opposed to a pumping pattern, decreases the number of abrupt changes in motion, decreases joint accelerations, and increases time spent during the propulsion phase; all of which contribute to increased biomechanical efficiency [56]. Rolling resistance is the amount of force opposing the motion of the wheel as it rolls across various surfaces [57, 60-63]. Many factors can affect the amount of RR including contact surface area, material composition, laden weight, tire pressure, and the inelastic deformation of the tire and wheel during propulsion [57, 60, 64]. Additionally, increased load has been shown to increase the rolling resistance significantly [56, 62].The greater the RR the more force that is needed to move the wheelchair [57, 60].

The effect of in-wheel suspension systems on rolling resistance is largely unknown [62]. Furthermore, it is unclear how in-wheel suspension affects propulsion efficiency in human users. In-wheel suspension systems operate by deforming to absorb shock and vibration during periods of high intensity vibrations or shocks transmitted from the environment, but it is unknown how they operate during propulsion. If they are also deforming in a method that absorbs or inhibits the transfer of propulsion torque when the MWU propels, then they could theoretically increase the propulsion cost and rolling resistance of the system by altering factors such as contact surface area. Some in-wheel suspension systems, such as the LoopWheels, are also heavier than standard wheels and have considerably different geometries and inertial properties, which could further affect the rolling resistance and propulsion cost. Lastly, if the suspension wheels are deforming under just the load of the MWU, then it would change factors that impact propulsion mechanics, such as the position of the shoulders relative to the pushrim. By sitting lower, the MWU is in a less biomechanically efficient position to propel [55, 56, 59]. Other effects due to the deformation such as dynamic axle movement and changes in front-rear weight distribution may also impact overall experience. To this end, the objectives of this study were to compare the rolling resistance values between in-wheel suspension systems and standard wheels and compare the propulsion mechanics of in-wheel suspension systems compared to standard wheels. The specific aims for this study are to:

1. Determine and compare the component level rolling resistance between standard spoke, Spinergy CLX, and LoopWheels Urban wheels. We hypothesized that the rolling resistance modulus for the LoopWheel would be highest over all loading conditions, followed by the CLX, then the standard spoke wheel.

2. Measure and compare the static axle position deformation of each wheel type under static loading conditions
    a. We hypothesized that the vertical deformation for the LoopWheel will be statistically significantly larger than that of the standard spoke wheel and CLX.
3. Investigate the harmonic oscillatory effects of in-wheel suspension during propulsion
    a. We hypothesized that the fore-aft amplitude, vertical amplitude, and oscillation period of the in-wheel suspension wheel will be significantly larger than standard spoked wheels and CLX.

4.3 Materials and Methods:

4.3.1 Component Level RR Testing

One validated method of evaluating rolling resistance is a component level analysis using an independent wheel (no wheelchair frame) on an active rolling drum and force sensors to determine the rolling resistance force when a constant speed and load are applied [61, 62]. This component level analysis also allows for testing on different surfaces. This methodology has been used to compare the rolling resistance between several different wheel and surface types. Studies have found that toe angle, tire pressure, load, and tire type have the largest impact on rolling resistance in manual wheelchairs [61, 62]. The component level rolling resistance analysis was conducted on a drum-based machine developed by Ott et al. (Figure 14) [61]. Using the drum tester, three wheel types were tested and compared in this study: the LoopWheels Urban, Spinergy CLX, and a standard spoked Quickie (Quickie, SouthwestMedical LLC, Phoenix, Arizona) wheels

(Figure 3). All wheels were outfitted with standard ½” (1.27 cm) diameter anodized push-rims and Marathon Plus tires filled to the recommended 125 psi.

Figure 14: Drum-based rolling resistance tester developed by Ott et al.[61]

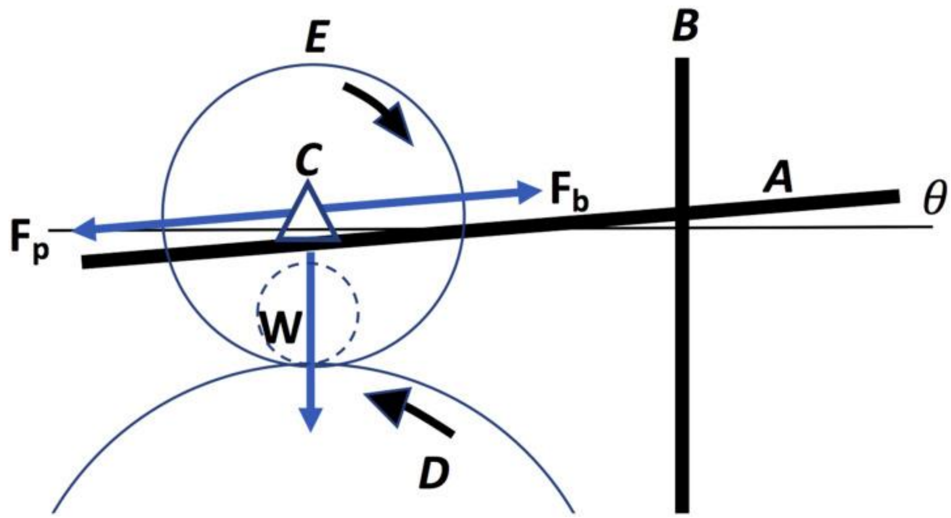


Figure 15: Free body diagram of component-level RR test where F_p = primary load cell force, F_b = backload load cell force, θ = angle caused by arm deflection from varying applied test weight, A = arm supporting air bearings, B = vertical support for arm, pivot point, C = axle sled (frictionless contact because of air bearings), D = rotating drum, E = propulsion wheel (caster wheel shown with dotted line), W = user and wheelchair weight applied to that wheel. Diagram and description was obtained with permission from [61, 65].

This device utilized an active drum to apply a constant tangential force to the wheel, air bearings providing frictionless movement, load cells measuring the amount of force resisting the direction of motion and the (1 lb.) backload. A machinist digital level measures angle of the air bearings on the arm assembly, which is used to calculate the small residual load (due to backload). These calculations are based off of the free body diagram in Figure 15. Exact specifications and details on the device are described elsewhere [61, 62, 65]. Wheels were tested on two surfaces, Forbo linoleum and low-pile carpet with ¼" felt carpet pad, at a tangential speed of 1 meter per second with three trials (30 seconds) for each of the five different loads (35-115 lbs., increments of 20 lb.). Factors and ranges were chosen to compare findings of suspension wheels to a previous study on non-suspension wheels [62].

4.3.2 Static Wheel Deformation

Wheel deformation was determined from a static position and displacement of the axle under the same loading conditions from the drum-based testing. A level sensor provided the angle of displacement (Figure 3 theta) in the anterior-posterior direction and the length of the arm (24 inch) (Figure 13 distance from B to C) multiplied by the sine of the angle provided the amount of deformation. Dynamic wheel deformation while rolling at steady state (1 m/s) was determined by recording the range of the angle of displacement (Theta min and Theta max) and then using the right angle relationship to obtain the variability in vertical axle displacements. An example setup with the LoopWheels is shown in Figure 16.

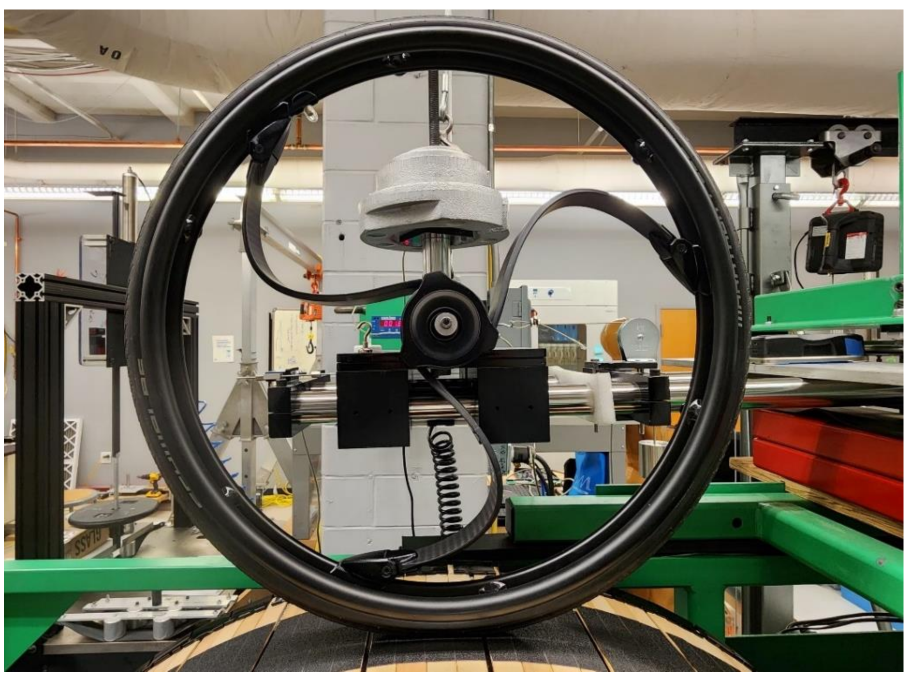


Figure 16: Example set up of the RR drum tester. The configuration shown is with the LoopWheels loaded at 115 lbs.

4.3.3 Propulsion Mechanics Testing

Prior studies with in-wheel suspension systems have suggested that the repetitive loading and unloading of the springs and dynamic changes in axle position create a oscillatory and/or pitch perturbation that could be responsible for the decreased propulsion efficiency [20, 66] In an attempt to capture the oscillatory effects on the wheelchair, propulsion mechanics was investigated using a system level approach. A Panthera X ultralight wheelchair was fitted with a Shimmer3 accelerometer attached to the middle of the axle tube, directly under where the MWU's spine would be (Figure 17). The +Z axis on the sensor was oriented to point in the direction of gravity (straight down) and the +X axis was oriented anteriorly. A male able-bodied surrogate MWU (weight: 67.13 kgs (148 lb), height: 170.18 cm (67 in)) propelled in a straight line using a semi-circular propulsion pattern over a smooth linoleum surface 22 meters (72 feet) in length (Figure 18) using the standard spoked, CLX, and LoopWheels. An auditory metronome set at 60 beats per minute was used to ensure repeatability and frequency of pushing for all trials. This pace and surface length resulted in six steady-state propulsion cycles per trial. A trial was repeated if more or less than six propulsion cycles occurred. Data collection from the accelerometer was initiated from start up and continued until the user passed the finish (taped) line. Post-processing removed start up and finish conditions to portray only steady state propulsion. Data from sensors was collected at 100 Hz. This ensured collected data would represent frequencies related with propulsion.

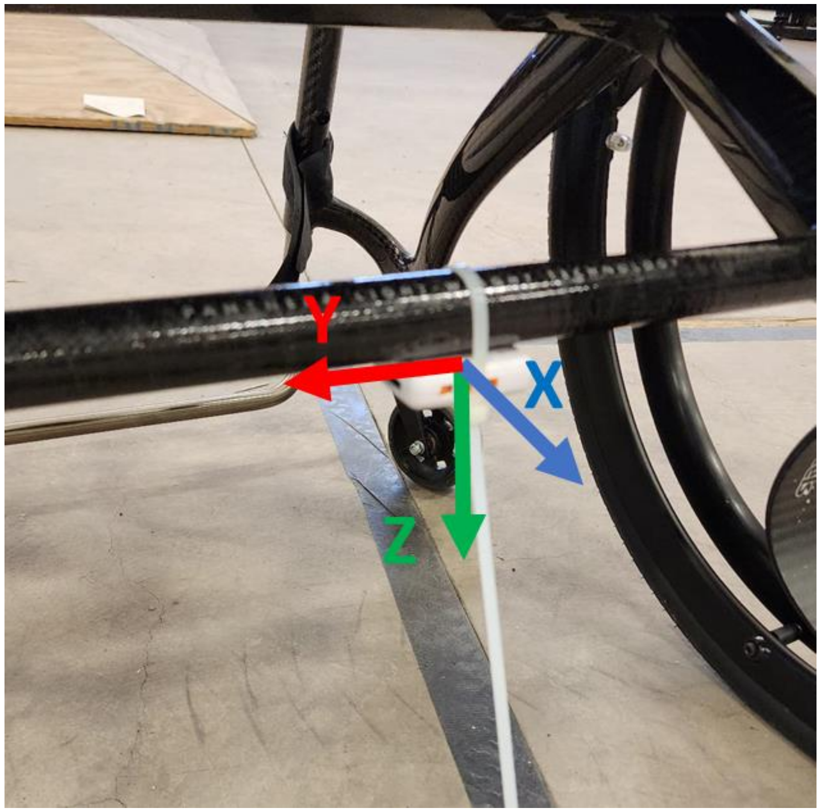


Figure 17: Shimmer3 placement for propulsion testing. Sensor coordinate system is also shown.

Figure 18: Linoleum surface used for propulsion testing (72 feet).

4.3.4 Data Analysis

Rolling resistance values from during the trials and vertical axle displacement values from right before the three trials (derived from the angle arm displacement) at each of the five loading conditions were averaged and compared in separate two-way ANOVAs comparing the main effects and interaction effects of load and suspension type on rolling resistance and vertical displacement for each surface tested. Significant differences reported in the overall F-test prompted individual pairwise comparisons using Bonferroni's adjustment.

Accelerometer data was analyzed separately for the vertical (Z axis) and fore-aft (X axis) directions. A 4th order lowpass Butterworth filter with a cut off frequency of 5 Hz was applied to remove high frequency noise and retain propulsion characteristics (each cycle was approximately 1 Hz). Vertical accelerations were post-processed to remove the effects of gravity by removing the DC component of the signal. Acceleration data in the forward direction was used to approximate propulsion phase and recovery phase. Local minimum X axis acceleration was used to estimate the start of propulsion (where force is applied to the pushrim and the wheelchair accelerates) and a local maximum was used to estimate the end of the propulsion phase (e.g. when the hand is released from the pushrim, the wheelchair will begin to experience negative accelerations due to the coast down and rolling resistance forces). Acceleration data in both X and Z directions were split into propulsion cycles (start of propulsion phase to start of next propulsion phase) with the transition to the recovery phase noted. Time spent within each cycle was normalized to percent of propulsion cycle (0-100%) and all six cycles were averaged. Total amplitude (maximum minus minimum signal) in fore-aft and vertical directions, as well as the period between oscillations in the vertical direction were calculated and averaged over all pushes for each wheel type. A repeated measures ANOVA investigating the effects of wheel type on each of the oscillation variables described above from the six pushes (n=6), conducted at the 95% confidence level, was used to determine significant differences between wheel types for oscillation variables. Characteristic differences between propulsion profiles were also qualitatively analyzed. All statistical tests were performed in IBM SPSS 29 (SPSS, IBM, Armonk, New York, USA).

4.4 Results:

4.4.1 Rolling Resistance

Results from the RR testing are shown in Figure 19-21. The LoopWheels had increased RR for both linoleum and carpet surfaces compared to the CLX and standard spoke. The relationship between CLX and standard spoke flipped between surface types, with the CLX having a higher RR on linoleum and the standard spoke having higher RR on the carpet (Figure 17). Carpet surface produced higher RR than linoleum, which was expected. All wheels showed similar levels of RR at the lowest load (around 0.1 and 0.5 lb for linoleum and carpet respectively) and the CLX and standard spoke show a very similar linear relationship as load increases. The LoopWheels has a much steeper relationship compared to the CLX and standard spoke, meaning that increasing the weight had a larger marginal effect on the RR for the LoopWheels compared to the others. When averaged over all loads, LoopWheels produced 118% more RR than standard spoke on the linoleum and 44% more RR on carpet. Over all wheel types, carpet produced 143% more RR than linoleum (around 0.5 lbs).

Results of the two-way ANOVA and Bonferroni's pairwise comparison indicate that increased load increases RR ($p < 0.01$), which is supported by findings from other studies [67]. Significant differences were also found between wheel types for RR over both surfaces ($p < 0.01$), revealing that LoopWheels had significantly higher RR than standard spoke on linoleum and carpet (p-value < 0.05) and higher RR than CLX on carpet (p-value < 0.01) (Figure 17). No differences were found between CLX and standard spoke on either surface.

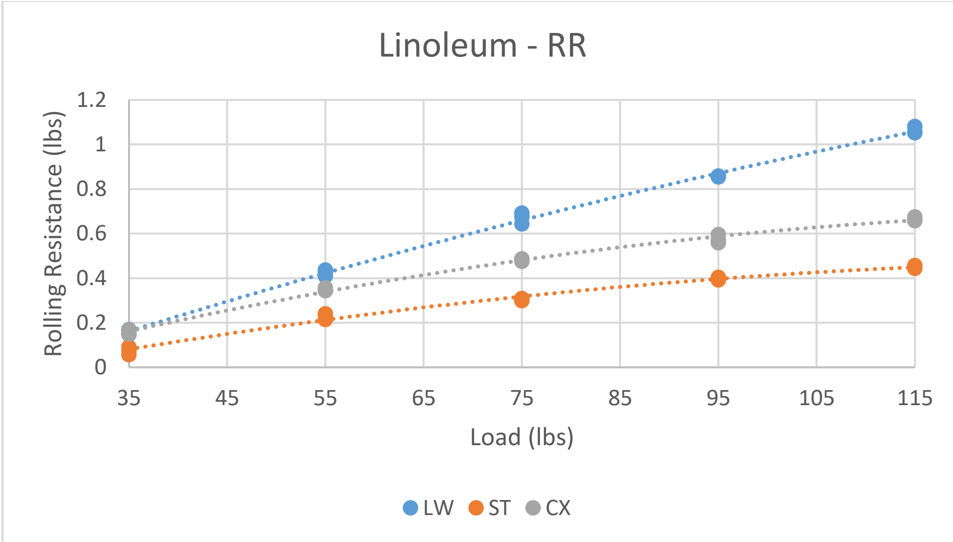


Figure 19: Rolling resistance versus load for linoleum surface.

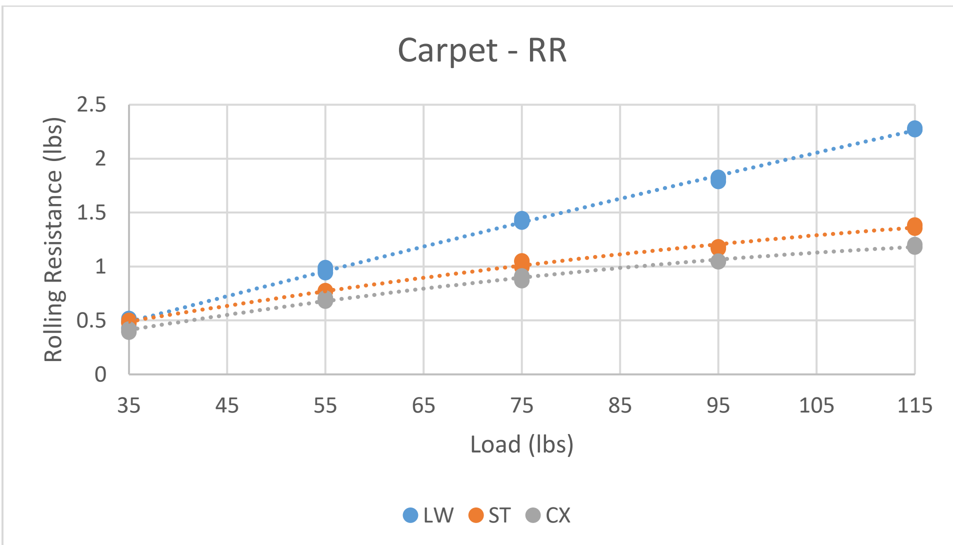


Figure 20: Rolling resistance versus load for carpet surface.

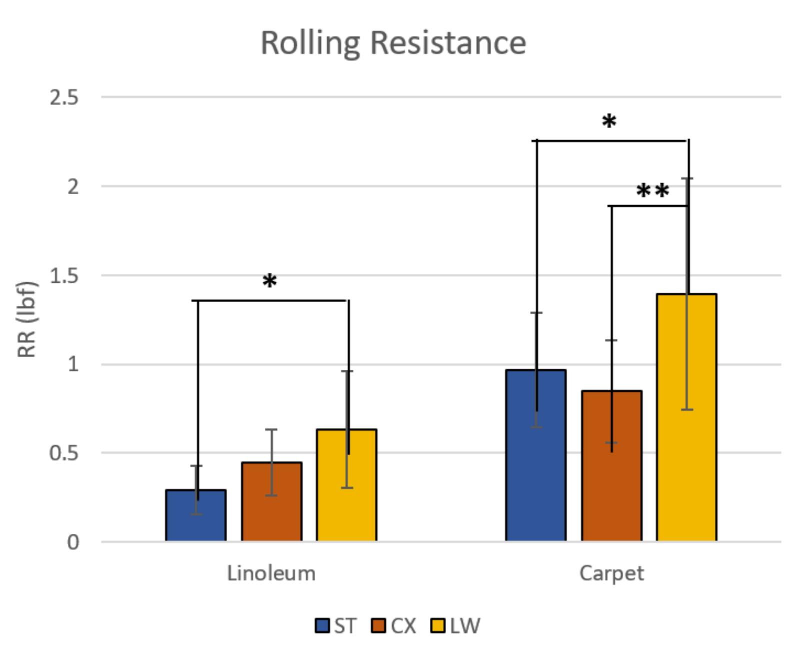


Figure 21: Bar plot of averaeg RR values between wheel type over linoleum and carpet. Significant differences aer denoted withg * and ** (p < 0.05 and 0.01 respectively).

4.4.2 Static Axle Vertical Displacement

Results from the static axle displacement calculations (Figure 22-24) show a similar relationship to the RR values for all wheels and loads (Figure 22. LoopWheels exhibited the largest amount of displacement at all loading conditions and had a steeper slope than CLX and standard spoke, indicating that there was a larger increase in static displacement for LoopWheels when the weight was increased compared to the others. CLX showed higher static displacement on the linoleum and less on the carpet compared to the standard spoke, which is the same trend observed in the RR values. Overall, LoopWheels moved around 0.7 cm (0.27 inches) statically on average over all loading conditions. The values from the dynamic displacement analysis showed that the CLX and standard spoke do not vary in height of the axle during steady state, but the LoopWheels had on average 0.05 cm (0.02 inches) of dynamic movement during steady state at all loading conditions. Two way ANOVA and Bonferroni pairwise analysis between wheel types showed that LoopWheels deforms significantly more compared to standard spoke and CLX on both linoleum and carpet surfaces (p-value < 0.01) (Figure 18). No significant differences were found between standard spoke and CLX wheels on either surface.

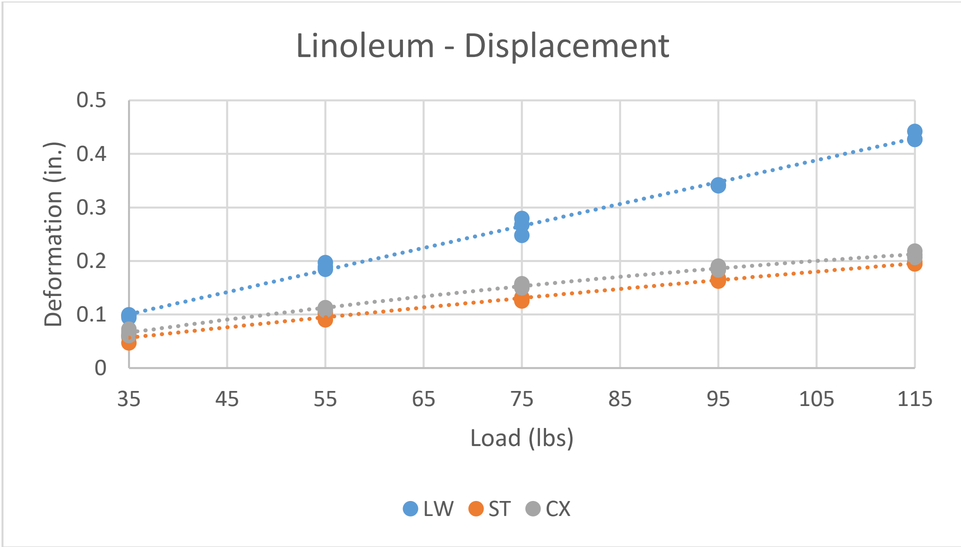


Figure 22: Static vertical displacement versus load for linoleum surface.

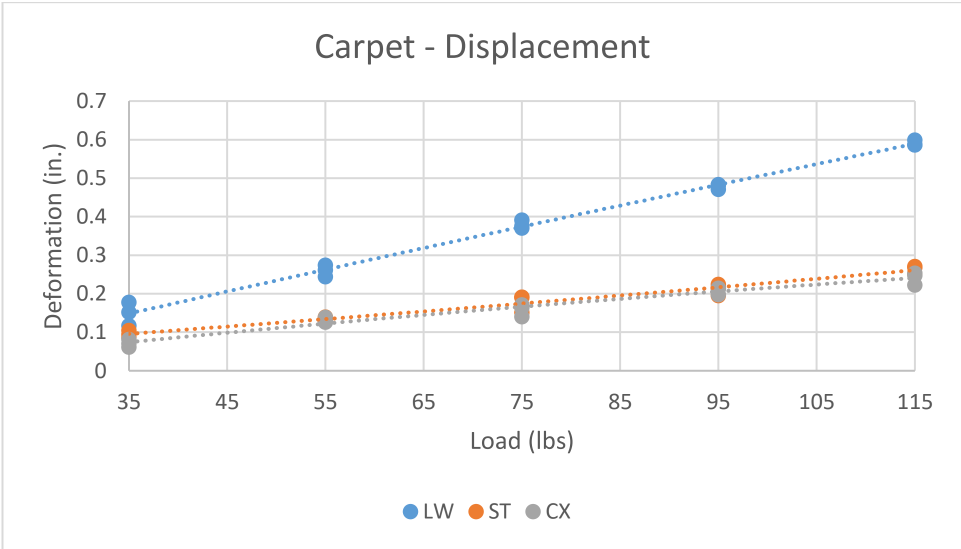


Figure 23: Static vertical displacement versus load for linoleum surface.

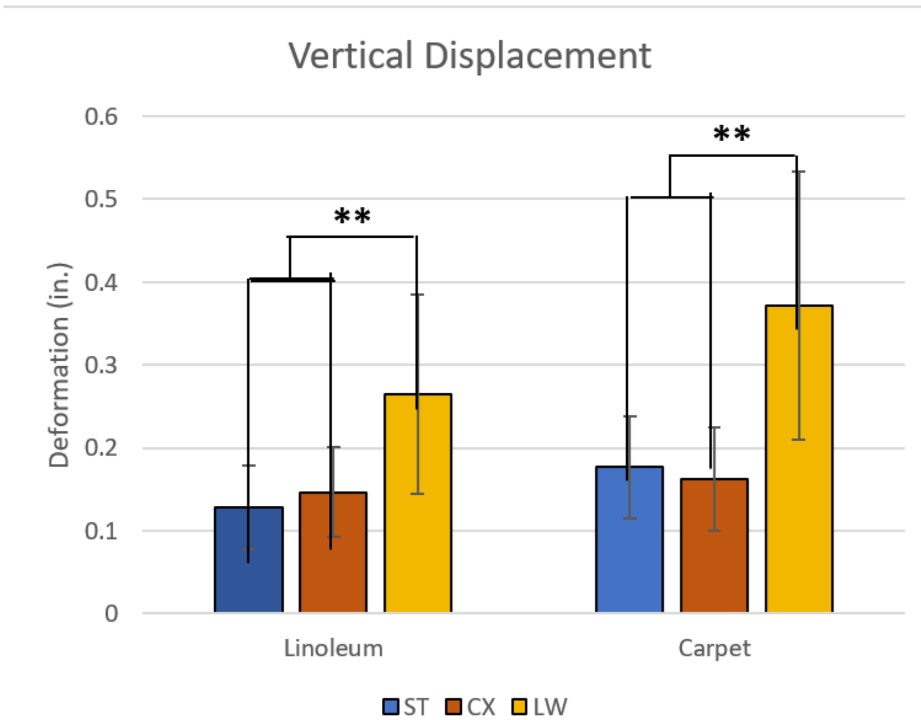


Figure 24: Bar plot of average static vertical displacement values between wheel type over linoleum and carpet. Significant differences aer denoted withg ** (p < 0.01).

4.4.3 Propulsion Mechanics

Analysis of the acceleration data during propulsion trials (Figure 25 and 26) revealed notable differences between the wheel types in each variable. LoopWheels exhibited the highest mean AP amplitude at 2.67± 0.30 m/s^2, followed by CLX at 1.78± 0.21 m/s^2, and standard spoke at 1.62± 0.23 m/s^2, indicating that LoopWheels provided significantly greater forward-backward frame acceleration amplitudes compared to the other wheel types (p-value < 0.01). In terms of vertical amplitude, LoopWheels again demonstrated significantly increased values (p-values < 0.01) than the other two types with a mean of 1.02± 0.07 m/s^2, while CLX had a mean of 0.31± 0.07 m/s^2, and standard had a mean of 0.35± 0.07 m/s^2. Although the differences were not significant, LoopWheels showed a slightly longer vertical period (0.29± 0.02 seconds) compared to CLX (0.26± 0.02 m/s^2) and standard (0.26± 0.02 m/s^2), indicating a slightly slower oscillation frequency. Overall, the results indicate that LoopWheels demonstrate increased harmonic oscillations in terms of AP amplitude and vertical amplitude compared to CLX and standard spoke wheels, with comparable oscillation frequencies across all three wheel types.

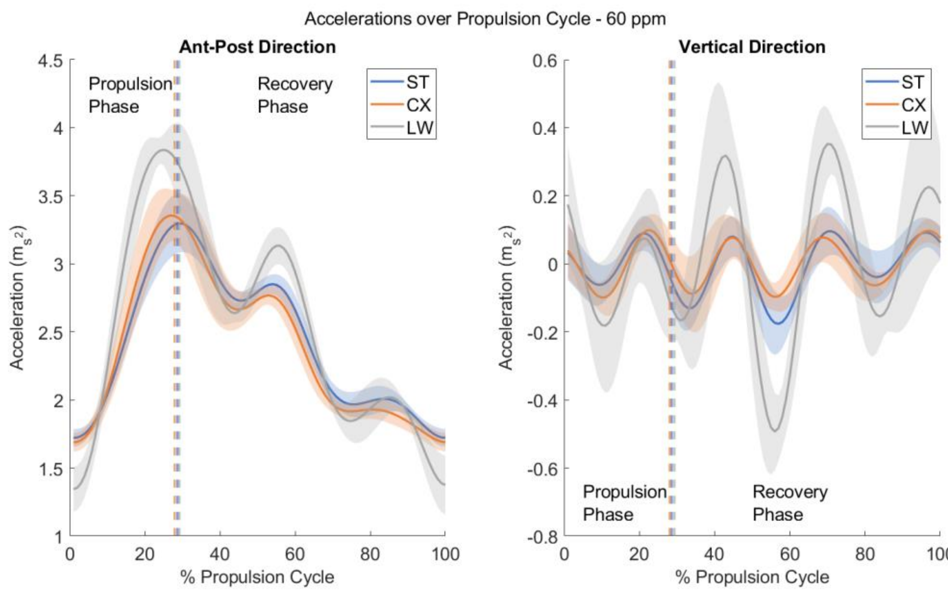


Figure 25: Average acceleration profiles for fore-aft and vertical direction during propulsion cycles. Propulsion and recovery phases are denoted and standard deviation is shown in shaded region.

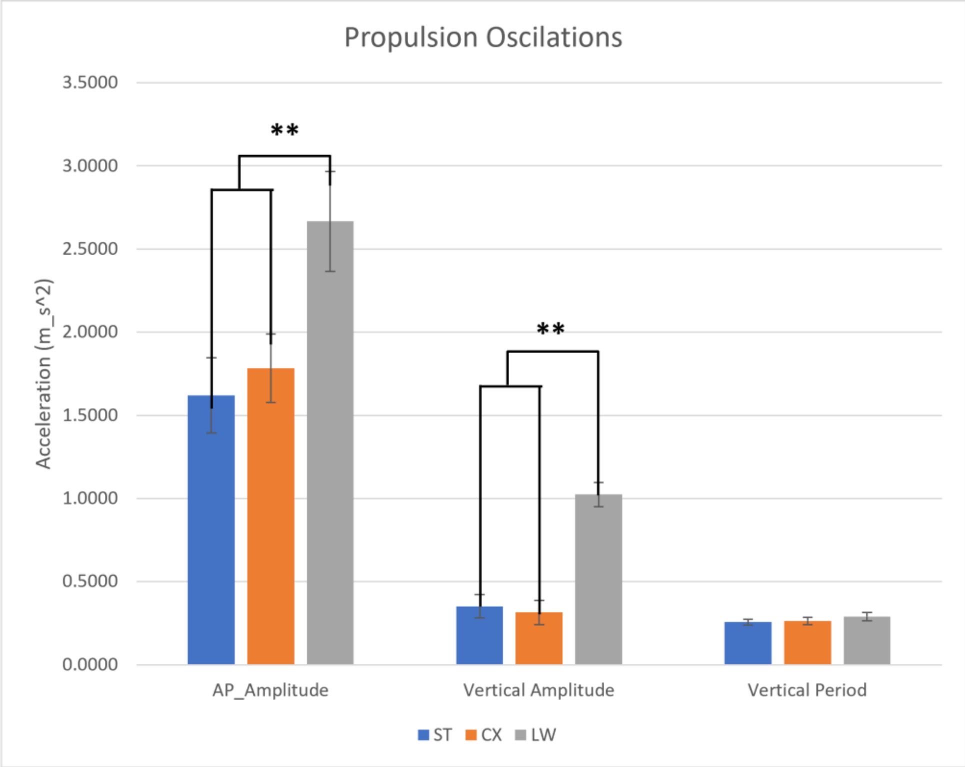


Figure 26: Bar plot of averaeg propulsion oscillation values between wheel type over linoleum and carpet. Significant differences aer denoted withg * and ** (p < 0.05 and 0.01 respectively).

The analysis of forward and backward amplitude (Figure 18) revealed notable differences between the standard spoke, CLX, and LoopWheels. LoopWheels exhibited a significantly higher fore-aft amplitude compared to both standard spoke and CLX conditions, with 1.04 m/s^2 (65%) ($p < 0.01$) and 0.88 m/s^2 (50%) ($p < 0.01$) increased fore-aft amplitude respectively. Similarly, the analysis of vertical amplitude indicated significant differences among the three conditions. LoopWheels demonstrated a substantial increase in vertical amplitude compared to both standard spoke and CLX conditions, with 225% ($p < 0.01$) and 192% ($p < 0.01$) more amplitude respectively. Regarding vertical period, although not statistically significant, there were trends suggesting differences among the conditions. LoopWheels showed a slight increase in vertical period compared to standard spoke and CLX conditions, with around 13% ($p = 0.06$) and 10% ($p = 0.176$) larger period respectively.

4.5 Discussion:

In this study, the effect of in-wheel suspension on rolling resistance and propulsion cost was characterized through various component and system level tests. This investigation was prompted from MWU feedback regarding use of the LoopWheels during a three month community-based field trial study. Comments from end users included that while the wheels absorbed shock and vibration when going over bumps and uneven terrain they also felt it was difficult to push long distances. Some users also commented that the wheels seemed to absorb some amount of force typically going towards forward propulsion as they were not able to travel as far with each push. Two key factors, rolling resistance and suspension deformation, were assessed to understand the impact of using LoopWheels on propulsion when compared to a standard spoked and CLX wheel. Component level testing of each wheel under different loading conditions and surfaces revealed that LoopWheels Urban produced significantly higher levels of RR compared to standard spoked wheels on linoleum and carpet (p-value < 0.05) and higher levels than the CLX on the carpet (p-value < 0.01). LoopWheels also had significantly higher static vertical axle displacement compared to standard and CLX wheels (p-value < 0.01). Results from this study are comparable to previous studies with the drum tester where pneumatic tires were shown to have RR values ranging from around 0.2 to 0.6 lbf [62]. LoopWheels has increased RR values compared to the other wheels tested previously, with the exception of airless insert tires which had a much large RR, even compared to LoopWheels (max 1.6 lbf compared to 1 lbf respectively) [62].

Findings from the static vertical displacement and dynamic movement of the axle during steady state indicate that there is a change in seat position with the LoopWheels at higher loads. This is supported by feedback from the longitudinal study (Chapter 3.0), where MWUs found it harder to transfer in and out of their wheelchair. Participants specifically indicated that they felt they had to lift themselves higher to transfer. While the static vertical displacement measured in this study was only 0.7 cm, this small change may be enough to drastically impact perception. This makes sense when considering that MWUs are used to performing transfers from their own equipment, which likely does not have the same amount of vertical displacement. Another factor that was not investigated in this study is the fore-aft displacement due to the static loading conditions. Findings from a bicycle study with in-wheel suspension found that the axle did not only move in the vertical, but in the fore-aft and rotational directions as well [66]. Their findings indicate that the axle position changes the contact surface area for the wheel, further causing energy losses in the system.

Previous studies investigating wheelchair and propulsion biomechanics have investigated the role of seat position on energy cost [58, 68, 69]. Smaller distances between the shoulder and wheel hub requires significantly less force and number of pushes to go the same distance when compared to larger distances. However, in this study LoopWheels had significantly more vertical displacement than the other two wheels, but yet had the greatest decrease in perceived propulsion efficiency. Therefore, the static vertical deformation alone is not enough to explain the increase in RR or estimated propulsion cost.

One other factor that might explain the increased perceived cost could be the additional weight of the wheels. While a relatively small amount compared to the weight of the MWU, studies have shown that small weights (<10 kgs) added onto a wheelchair significantly increase propulsion cost over a number of surfaces [67, 69]. A singular LoopWheel weighs about 0.80 kgs more than a standard spoked wheel, so this total 1.60 kg additional weight may account for increased RR and propulsion cost, however, this is a very small amount compared to the weight of the MWU and the rest of the wheelchair [67, 69].

These factors discussed so far, weight and axle position, affect all wheel types during propulsion. A more in-wheel suspension specific factor that has recently emerged in the literature is harmonic oscillation that occurs during movement [66]. A study conducted by Corno et al. investigated the effect of in-wheel suspension on bicycle mechanics; specifically, the effect of SoftWheels on vertical, horizontal, and pitch oscillations at varying speeds [66]. While not the same wheel type as the one investigated in this study, the general geometry and function between the SoftWheels and LoopWheels are similar. Namely, both consist of three equally spaced springs in place of traditional spokes on the wheel. Researchers used a model of the SoftWheels to simulate movement of the axle over different terrains and speeds [66]. Each spring-spoke was modeled as a spring and dampener in series, to account for the compression and decompression factors of the wheels [ref]. The springs undergo deformation three times during a period of revolution and this consistent compression and decompression accounts for a 7-15% absorption of power compared to standard wheels [66]. While the spring style used between SoftWheels and LoopWheels differs, it is likely that they share this same power dissipation property.

In this study, the fore-aft accelerations ranged from 1.4 to 3.9 m/s^2 for the LoopWheels and 1.8 to 3.3 m/s^2 for the other wheels. In the vertical direction the acceleration ranged from -0.5 to 0.4 m/s^2 for the LoopWheels, where as the other two only showed a range from -0.2 to 0.1 m/s^2. The acceleration patterns for standard spoke, CLX, and LoopWheels were visually different during both the propulsion phase and recovery phase of the cycle. This could be due to the differences in the wheel's suspension system and the harmonic oscillations of the wheel. For instance, the LoopWheels showed a different pattern of acceleration compared to the standard and

CLX wheels during the propulsion phase, with around 64% larger amplitude in the fore-aft direction compared to standard wheels. The LoopWheels started from a lower acceleration and increased to a greater acceleration at the end of the phase compared to the other wheels. This could be because the LoopWheels built-in suspension system is absorbing the initial input of force from the push rim but then “springs” it back towards the end of propulsion. This is symbolic of the feelings of the wheels absorbing the push of the MWU. In the recovery phase the LoopWheels have a much larger amplitude during the oscillation through the recovery phase. This increased acceleration could imply that more energy is going towards moving the wheel hub relative to the wheel rim than is going towards forward propulsion. The larger amounts of oscillation and magnitude in accelerations found in the recovery phase could also be representative of the loading of the springs during the propulsion phase and the unloading from the force input in the recovery phase. In addition, the energy dissipated during the cyclic compression and decompression of the springs during both phases causes MWUs to feel the need to exert more energy to maintain speed. Furthermore, the increased RR present in the LoopWheels could be decreasing coast down time, causing users to put more effort (acceleration) in to achieve the same amount of distance as with their standard wheels. Previously, researchers were able to determine the amount of forces and moments induced on the axle during propulsion using an instrumented pushrim [70]. Results on vertical and fore-aft forces profiles on the axle during propulsion are very similar to the acceleration profiles found in this study; however, the presence of the oscillation patterns was not seen in this previous study [70]. Indicating that there may be notable differences in force profiles as well between the in-wheel suspension and standard conditions.

Other notable characteristics of the acceleration profiles are the small changes in fore-aft accelerations and the increased accelerations in the vertical during the recovery phase. The first bump in the fore-aft acceleration data in the recovery phase may be indicative of the MWU shifting their weight from the front to the back of the wheelchair after releasing the hand rim. While this bump exists in all three wheel types, the LoopWheels displays a much larger amplitude difference. This could mean that the LoopWheels causes deformation and oscillatory effects during mundane actions, such as repositioning, and is highly sensitive to movement. The same is true in the vertical accelerations, where the LoopWheels showed a 225% larger amplitude than the standard wheels. Furthermore, a slight phase shift can be identified in the LoopWheels profile, where there is a slight delay in the oscillations whereas the CLX and standard behave similarly and with a similar period of oscillation. Matching up the time point for the shifting of weight in the recovery phase, the LoopWheels presents not only significantly larger vertical acceleration amplitudes compared to the other wheels but also compared to the propulsion phase. The spring-dampener behavior of the in-wheel suspension is likely causing this increased frame acceleration. This suggests that LoopWheels behave differently during the propulsion and recovery phase, with more energy being absorbed and used for deformation rather than forward propulsion. This dynamic response is most likely what accounts for the MWUs' perception that the LoopWheels is absorbing their push. While not conclusive, these trends suggest potential alterations in movement patterns with the use of LoopWheels, characterized by slightly longer intervals between vertical oscillations. These differences in the LoopWheels may contribute to smoother rides and improved comfort for wheelchair users, but may also dissipate and absorb the energy required for forward propulsion. This suggests that while LoopWheels might offer superior shock absorption and wheel deflection capabilities during periods of uneven terrain, they also cause energy absorption over smooth surfaces. Furthermore, the increased frame accelerations indicate that greater amounts of energy are dissipated in the system during propulsion over a smooth surface. This effect may be undesirable, as many MWUs propel over indoor smooth surfaces for employment, recreation, around their home, and clinics. However, since there was only one individual who did all the pushes, the statistical significance is likely inflated. Thus, further testing with multiple individuals and other factors on propulsion mechanics should be investigated.

4.6 Limitations and Future Directions:

Static vertical displacement values cannot be validated for accuracy as measurements and calculations were based on an assumption of a right triangle between the horizontal, the arm holding the wheel, and the vertical deformation. However, results from previous studies on in-wheel suspension in bicycles reveals that there is anterior-posterior deformation and pitch deformation as well as vertical. The fore-aft and angular deformation could not be accurately measured in this study, so the true translation and rotation of the wheel hub is unknown. Lastly, we were unable to measure the actual propulsion cost or force required from the MWU. An instrumented pushrim or physiological analysis would enable a more accurate assessment of propulsion cost for the in-wheel suspension. Future investigations on the dynamic behavior of the wheel are necessary to fully characterize mechanics behind MWU propulsion.

Future investigation is necessary for a more full characterization of harmonic oscillation differences between LoopWheels and the others. Testing with different applied loads and over different surface inclines could further elucidate the propulsion mechanics and harmonic oscillation effects that the in-wheel suspension introduces. Findings from this study indicate that the LoopWheels having a significantly higher RR and vertical deformation compared to standard and CLX wheels may lead to significantly higher levels of estimated propulsion costs. This effect was seen over smooth surfaces which are commonly traversed by MWUs.

4.7 Conclusions:

In conclusion, this study provides valuable insights into the impact of in-wheel suspension systems on rolling resistance (RR), deformation, and propulsion mechanics in manual wheelchairs (MWUs). Our findings indicate that LoopWheels Urban, equipped with in-wheel suspension, exhibited significantly higher RR compared to standard spoked wheels on both linoleum and carpet surfaces, as well as higher levels than the CLX on carpet surfaces. Specifically, LoopWheels demonstrated 118% more RR than standard spoked wheels on linoleum and 44% more on carpet. Moreover, LoopWheels exhibited significantly higher deformation compared to standard and CLX wheels, with an average deformation of around 0.27 inches over all loading conditions. Analysis of frame accelerations revealed distinct differences among the wheel types, with LoopWheels showing significantly greater fore-aft and vertical amplitude compared to standard spoked and CLX wheels. These findings suggest that LoopWheels facilitate smoother movement dynamics characterized by increased oscillations during propulsion, potentially leading to improved comfort for MWUs. However, these effects may be largely undesirable as this increased oscillation may also contribute to increased energy loss by the wheel during propulsion, as noted in user feedback. Furthermore, dynamic analysis revealed a slight phase shift in the acceleration profile of LoopWheels during both propulsion and recovery phases, indicating a more dynamic behavior of the wheel hub relative to the rim caused by the cyclic loading and unloading of the spring dampener system.

In summary, our findings highlight the complex interplay between in-wheel suspension, rolling resistance, deformation, and propulsion mechanics in manual wheelchairs. By elucidating these relationships, we contribute to the ongoing efforts to optimize wheelchair design for enhanced comfort, efficiency, and overall quality of life for manual wheelchair users. Other options to be developed in the future could include more technical approaches, such as a switch to turn on or off the suspension system.

5.0 Impact Of In-Wheel Suspension On Mwus Summary

The study set out to assess the efficacy of LoopWheels Urban, an in-wheel suspension system, in mitigating harmful vibrations and shocks experienced by manual wheelchair users (MWUs) across different terrains. By directly involving experienced MWUs in their own wheelchairs, the study ventured beyond previous investigations, incorporating real-world scenarios. Results indicate that LoopWheels Urban holds promise in reducing whole-body vibration (WBV) levels at the backrest and footrest, compared to standard spoked wheels and CLX wheels, though no significant differences were noted at the seat or in perceived comfort. Furthermore, LoopWheels was assessed on community levels of WBV exposure, pain, fatigue, and overall quality of life among MWUs. Unlike prior studies confined to laboratory environments, this study embraced the lived experiences of seasoned wheelchair users, providing insightful data on the practical benefits of in-wheel suspension systems in enhancing daily comfort and mobility for MWUs. Finally, the effects of in-wheel suspension systems on MWUs from multiple perspectives were examined. The implications of such systems on rolling resistance (RR), deformation, and propulsion mechanics were investigated, while also considering their influence on pain, fatigue, and overall quality of life among MWUs. This holistic approach sheds light on the multifaceted impacts of in-wheel suspension systems, offering valuable insights into their potential benefits and limitations in real-world scenarios.

5.1 Vibration Analysis:

The study found that LoopWheels significantly reduced RMS vibrations at the backrest and footrest by 7-10% compared to CLX and standard wheels and significantly reduced shocks at the footrest by 7% compared to the CLX. However, all wheel types produced RMS and VDV values either within or exceeding the HGCZ for 4-8 hours of exposure, indicating potential health concerns. This suggests that efforts to reduce WBV in MWUs need further investigation. Furthermore, no significant differences in perceived comfort were observed across wheel types, but this could be due to the short duration of exposure to each of the obstacles and the limited number and type of obstacles tested. Natural and built environment often introduce unexpected and unique terrain, so it is important to test WBV on MWUs under real world living conditions to understand the true levels of exposure.

5.2 Quality of Life and User Experience:

The study further explored the impact of in-wheel suspension on pain, fatigue, and overall quality of life among MWUs. After using LoopWheels for around 12 weeks, MWUs reported a significant improvement in daily pain (15% less median neck pain and 25 less median pain interference per day) and fatigue (8% less median fatigue) experienced, attributing it to the wheels' ability to absorb shocks and vibrations better than traditional wheels. Participants reported an overall improvement in quality of life and were likely to recommend the in-wheel suspension to other MWUs. However, some users noted drawbacks such as increased weight and width, which made propulsion more challenging and navigating certain environments more cumbersome. Other notable changes included increased difficulty of transferring due to possible changes in seat position and feelings that the in-wheel suspension was absorbing their energy, thereby decreasing overall propulsion efficiency perceived by the MWU.

5.3 Community Vibration Exposure:

Participants using LoopWheels experienced a significant reduction in both RMS vibration (around 35%) and VDV shock (around 50%) compared to standard wheels when juxtaposed with other community based vibration studies, bringing the vibration exposure levels below hazardous thresholds defined by ISO 2631. Despite shorter exposure durations in this study, the observed vibration levels remained within safe limits, suggesting potential benefits for MWUs in daily activities. A lack of control period using their standard equipment makes it difficult to know the true effects that the suspension had on reducing the overall daily WBV.

5.4 Individual Variability and Preferences:

The study highlighted substantial variability among participants in their responses to the intervention, indicating that individual experiences with in-wheel suspension systems vary widely. While the overall findings show statistically significant improvements in pain and fatigue, individual responses may differ based on factors such as baseline health status, level of disability, and personal preferences. Findings indicate that specific demographics of MWUs will benefit the most from the in-wheel suspension; specifically individuals with a higher BMI and who are less active are more likely to find a perceived benefit from the in-wheel suspension. A personalized approach to intervention delivery may enhance the effectiveness of in-wheel suspension systems in improving pain and fatigue management for MWUs. Furthermore, a more detailed clustering analysis of the user characteristics in relation to their health outcomes and perceptions of the LoopWheels may be possible with a larger dataset which could further guide the recommendations for MWUs.

5.5 Propulsion Efficiency and Trade-offs:

While LoopWheels demonstrated effectiveness in reducing vibration, previous studies noted an increase in propulsion cost compared to standard wheels. The study suggests a potential trade-off between vibration reduction and increased energy cost, as seen in other suspension systems in bicycles and wheelchairs. Despite the benefits of vibration reduction, the additional weight of LoopWheels and the effect of the suspension elements may pose challenges in propulsion and other aspects of wheelchair use.

5.6 RR and Propulsion Cost Analysis:

Dynamic and static testing were conducted to assess the effect of in-wheel suspension on RR and propulsion cost. LoopWheels exhibited significantly higher RR compared to standard spoked wheels on linoleum and carpet surfaces (118% and 44% more, respectively), while CLX reported 53% more RR on linoleum and 13% less on carpet. Additionally, LoopWheels demonstrated significantly higher deformation compared to standard and CLX wheels under static loading conditions, indicating a potentially smoother movement dynamic but also potentially contributing to greater energy absorption by the wheel due to the movement of the springs, as noted in user feedback.

5.7 Propulsion Mechanics Analysis:

The analysis of propulsion mechanics revealed distinct differences among the wheel types. LoopWheels exhibited greater fore-aft (50-64%) and vertical amplitude (190-225%) compared to standard spoked and CLX wheels during both propulsion and recovery phases. This suggests a more dynamic behavior of the wheel hub relative to the rim, potentially leading to improved comfort for MWUs but also energy losses in the system affecting propulsion efficiency.

5.8 Limitations and Future Directions:

One limiting factor that occurred while conducting this study was in the RR drum testing machine. The machine is highly sensitive to levels of moisture in the air bearings and load cells, and there were periods of time between testing of the wheels where levels of moisture exceeded recommended levels. In this case, testing was halted and a new dehumidifier was installed into the system. These changes to the equipment and hardware should not have had a large effect on testing procedures, but as these issues were not seen in the prior validation testing for the machine, it is not possible to say it has no effect.

The study faced several other limitations, notably challenges in accurately measuring vibration transmitted to MWUs and the use of standard pushrims that may have interfered with the users propulsion ability and comfort during the in lab study. To address this, future studies should prioritize direct measurement of vibrations on MWUs and assess propulsion efficiency with actual users and the same type of pushrim that they use on a regular basis. Moreover, factors related to long-term use, such as impacts on transfers, transportability, durability, as well as pain and fatigue, warrant further investigation. Other notable surface and maneuvers that were not tested in the in lab portion of the study, such as curb ascents, may pose different WBV profiles. Further testing on several different obstacles is necessary to fully characterize the impact of in-wheel suspension. Additionally, the study encountered various challenges, including manufacturing defects in LoopWheels, issues with data collection using Verisense sensors, and falling short of the original number of subjects (collected 26 instead of the proposed 30). This delayed start is largely due to vendor supply issues and the COVID-19 pandemic. Furthermore, the study acknowledged limitations such as moisture sensitivity in the RR testing machine and difficulties in accurately measuring vertical deformation. The estimated propulsion cost initially relied on static and dynamic analysis of the axle deflections during drum operation, suggesting the necessity for additional dynamic testing to fully understand the mechanics behind MWU propulsion. Dynamic testing was limited to the instrumentation available at the time of the study (e.g. frame mounted accelerometers). Future investigations are required, preferably using motion capture to more accurately characterize the axle deflections and spring-dampener motions during propulsion. Instrumented wheels and pushrims could also be utilized to further understand the effect of in-wheel suspension on propulsion costs. Despite these limitations, the study provides compelling evidence for the potential benefits of in-wheel suspension systems in enhancing quality of life and reducing WBV exposure for MWUs. Moving forward, future investigations should prioritize examining dynamic behavior under varying loads and surface conditions to gain deeper insights into the effects on propulsion efficiency and user experience. Particularly, possible considerations could be made into a toggle for suspension or one that is more capable of adapting to different terrain and that can change the effect of the suspension based on the environment. One method of implementing this could be an additional stiff beam from the axle hub to the wheel rim. When secured, the system would behave similar to a standard wheel with the spokes, but the beams can then be disengaged to allow the C-shaped springs to absorb impact from the environment.

Another interesting avenue for exploration could be the specific obstacles encountered by MWUs in the community. In lab testing on different surfaces or action, such as curb drops, performed by MWUs could enable the creation of a database of acceleration profiles for common surfaces. Principle component analysis on the acceleration profiles along with classification models would allow more accurate tracking on MWU's encounters of obstacles and what type of obstacles are commonly propelled over. This future investigation could reveal that MWUs are only going over curb drops once every week, whereas they propel over tile surfaces multiple times a day. These individual lifestyles could be instructive on whether or not an individual will benefit from using the in-wheel suspension system. This future investigation could also reveal underlying education disparities in the MWUs. If not many individuals are performing actions such as wheelies or curb ascents, then further efforts into educating the MWUs need to be made.

Lastly, there were several confounder and covariates in the study that were not analyzed. These include any accessories on the wheelchair, the type of car the MWUs transfers to (if they drive), and use of power assist devices such as SmartDrive. Many of these factors have the potential to interact with the in-wheel suspension and affect the perception by the MWU.

5.9 Conclusion:

In-wheel suspension systems offer a potential solution to mitigate WBV for MWUs, with LoopWheels displaying promising results in vibration reduction. Nevertheless, challenges persist in attaining optimal vibration reduction without compromising propulsion efficiency and other aspects of user experience. Further research is warranted to address these challenges and refine wheelchair design to enhance comfort and mobility for MWUs.

The study significantly contributes to our understanding of the impact of in-wheel suspension systems on pain, fatigue, and WBV exposure among MWUs. LoopWheels demonstrated effectiveness in reducing vibration exposure and enhancing daily comfort for participants. However, challenges such as the increased weight and width of the wheels underscore the necessity for ongoing research to optimize in-wheel suspension systems for MWUs. Overall, the findings indicate that in-wheel suspension systems hold promise as a viable solution for improving quality of life and mobility for MWUs, but further refinements and modifications are required to overcome existing challenges.

In conclusion, the study offers valuable insights into the intricate relationship between in-wheel suspension, rolling resistance (RR), deformation, propulsion mechanics, and the effect on pain and fatigue in manual wheelchairs. By elucidating these connections, continuous efforts can be made to refine wheelchair design, leading to enhanced comfort, efficiency, and overall quality of life for MWUs. This advancement in assistive technology has the potential to significantly improve mobility options for individuals with mobility impairments.

Appendix A VAS Comfort Survey

Wheelchair Ride Comfort and Ergonomics Questionnaire

< User Survey >

Enter Device ID for this Evaluation ____________

Please answer the following questions by marking your answers with a cross on the corresponding line, where the ends of the lines equal the strength of your response. Mark only one answer per question unless directed.

For the following questions, please use the following scale: 0 = extremely uncomfortable; 10 = extremely comfortable

1. How comfortable were you when pushing your wheelchair over the following surfaces?

Surface 1?

0 10

Surface 2?

0 10

Surface 3?

0 10

Surface 4?

0 10

Surface 5?

0 10

Surface 6?

0 10

Surface 7?

0 10

Surface 8?

0 10

Surface 9?

0 10

2. For the following questions please select your response based on your overall experience when pushing over these surfaces

How **stable or secure** did you feel when pushing over these surfaces?

______Not at All

______Fairly

______Moderately

______Very

______Extremely

Overall, how **easy was it to maneuver** over these surfaces?

______Not at All

______Fairly

______Moderately

______Very

______Extremely

How **comfortable were your hands on the pushrim** when pushing over these surfaces?

______Not at All

______Fairly

______Moderately

______Very

______Extremely

How **comfortable overall** did your body feel when pushing over these surfaces?

______Not at All

______Fairly

______Moderately

______Very

______Extremely

How **efficient** did you feel when pushing over these surfaces?

______Not at All

______Fairly

______Moderately

______Very

______Extremely

Are there any other comments you would like to provide? ____
____
____
____
____

Appendix B In-Lab Vibration Analysis Data Post Processing

Appendix B.1.1 Gravity Rotation:

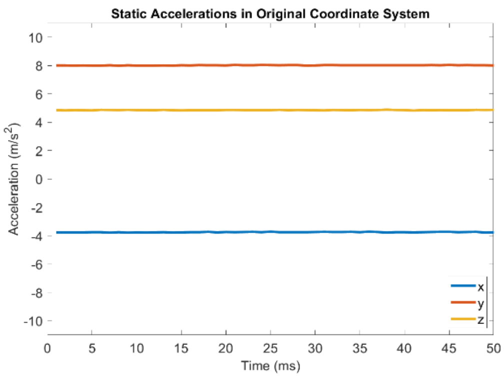


Figure 27: The plot above show the raw acceleration data from five seconds of a static position.

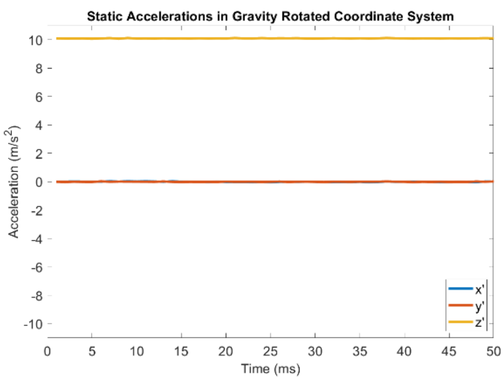


Figure 28: The plots above show the rotated acceleration data from five seconds of a static position.

Appendix B.1.2 Vibration Filtering:

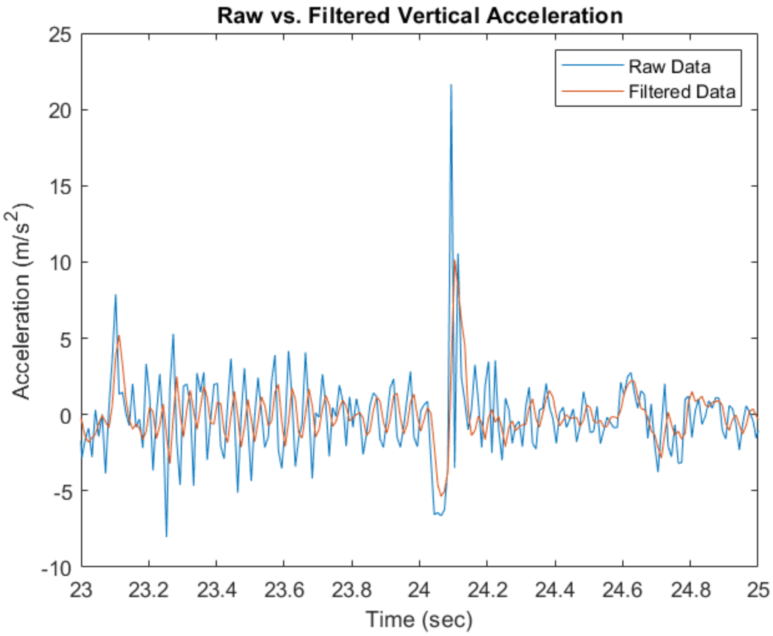


Figure 29: The plot above shows the raw and filtered vertical acceleration from two seconds of a trial.

Appendix B.1.3 Intensity Region Separation:

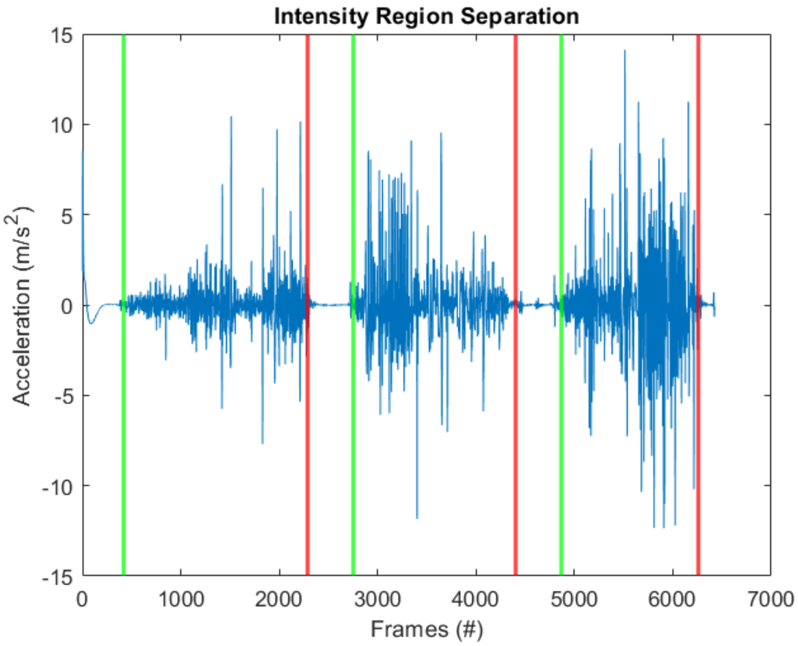


Figure 30: The plot above shows the acceleration measured by a sensor for the whole obstacle course. The intensity regions are defined by their start (green line) and their end (red line). Only data within each region was used to calculate WBV values.

Intensity region intervals were identified manually by visual analysis of the raw acceleration data. Since the participants are asked to start from a rest and end with a complete stop, the boundaries were chosen to include the region with steady-state motion.

Appendix C Subject Wise Percent Difference Histogram

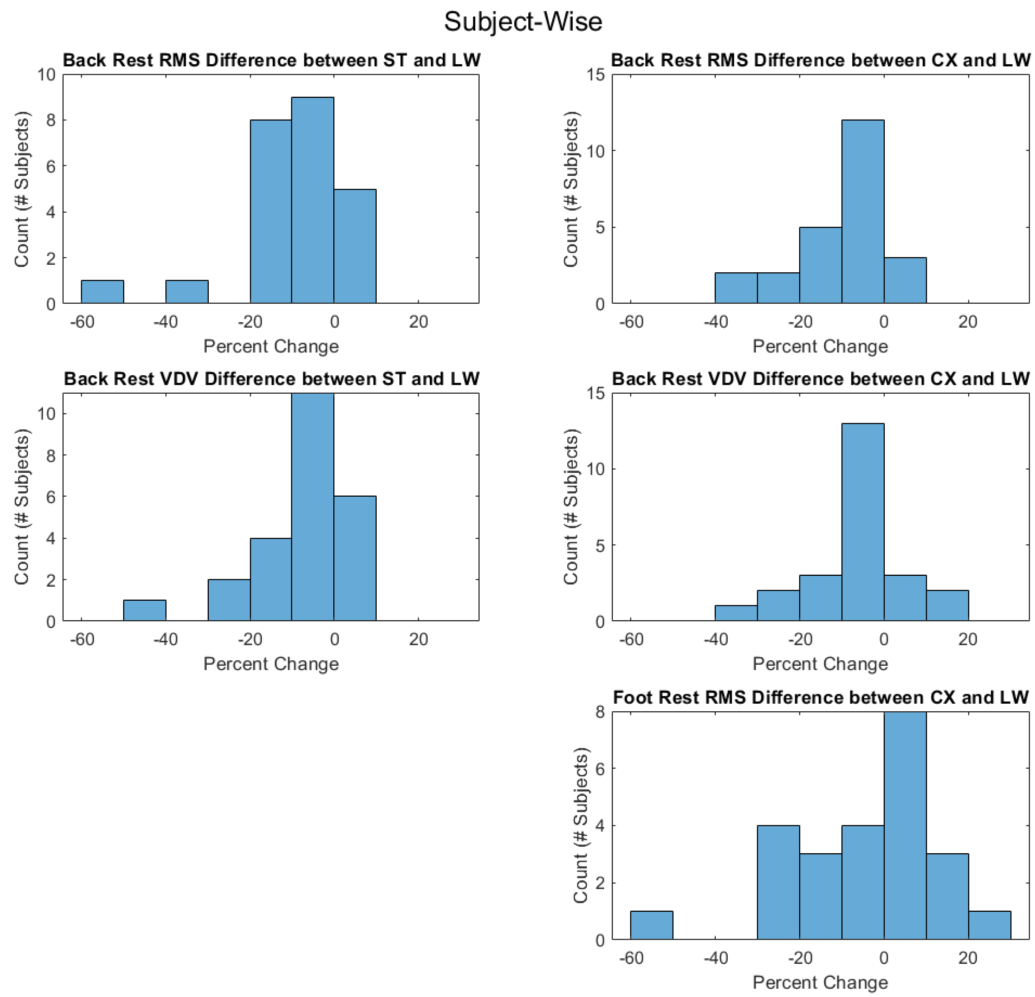


Figure 31: Subject-wise histogram of percent differences in WBV. Positive percent change indicates LoopWheels increases WBV and negative percent change indicates LoopWheels decreases WBV.

Appendix D Brief Pain Inventory

1903

Date: ___ / ___ / ___
(month) (day) (year)

Subject's Initials : ____________

Study Subject #: ____

PLEASE USE BLACK INK PEN

Study Name: ____________

Protocol #: ____________

PI: ____________

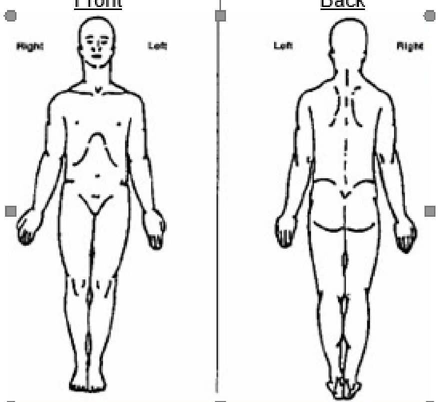
Revision: 07/01/05

Brief Pain Inventory (Short Form)

1. Throughout our lives, most of us have had pain from time to time (such as minor headaches, sprains, and toothaches). Have you had pain other than these everyday kinds of pain today?

☐ Yes ☐ No

2. On the diagram, shade in the areas where you feel pain. Put an X on the area that hurts the most.



3. Please rate your pain by marking the box beside the number that best describes your pain at its worst in the last 24 hours.

☐ 0 ☐ 1 ☐ 2 ☐ 3 ☐ 4 ☐ 5 ☐ 6 ☐ 7 ☐ 8 ☐ 9 ☐ 10

No Pain — Pain As Bad As You Can Imagine

4. Please rate your pain by marking the box beside the number that best describes your pain at its least in the last 24 hours.

☐ 0 ☐ 1 ☐ 2 ☐ 3 ☐ 4 ☐ 5 ☐ 6 ☐ 7 ☐ 8 ☐ 9 ☐ 10

No Pain — Pain As Bad As You Can Imagine

5. Please rate your pain by marking the box beside the number that best describes your pain on t average.

☐ 0 ☐ 1 ☐ 2 ☐ 3 ☐ 4 ☐ 5 ☐ 6 ☐ 7 ☐ 8 ☐ 9 ☐ 10

No Pain — Pain As Bad As You Can Imagine

6. Please rate your pain by marking the box beside the number that tells how much pain you have right now.

☐ 0 ☐ 1 ☐ 2 ☐ 3 ☐ 4 ☐ 5 ☐ 6 ☐ 7 ☐ 8 ☐ 9 ☐ 10

No Pain — Pain As Bad As You Can Imagine

Page 1 of 2

1903

PLEASE USE BLACK INK PEN

Date: ___ / ___ / ___ (month) (day) (year)

Subject's Initials : ________

Study Subject #: | | | | |

Study Name: ____________

Protocol #: ____________

PI: ____________

Revision: 07/01/05

7. What treatments or medications are you receiving for your pain?

8. In the last 24 hours, how much relief have pain treatments or medications provided? Please mark the box below the percentage that most shows how much relief you have received.

0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
☐ No Relief	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐ Complete Relief

9. Mark the box beside the number that describes how, during the past 24 hours, pain has interfered with your:

A. General Activity

☐ 0 Does Not Interfere ☐ 1 ☐ 2 ☐ 3 ☐ 4 ☐ 5 ☐ 6 ☐ 7 ☐ 8 ☐ 9 ☐ 10 Completely Interferes

B. Mood

☐ 0 Does Not Interfere ☐ 1 ☐ 2 ☐ 3 ☐ 4 ☐ 5 ☐ 6 ☐ 7 ☐ 8 ☐ 9 ☐ 10 Completely Interferes

C. Mobility

☐ 0 Does Not Interfere ☐ 1 ☐ 2 ☐ 3 ☐ 4 ☐ 5 ☐ 6 ☐ 7 ☐ 8 ☐ 9 ☐ 10 Completely Interferes

D. Norm al W ork (includes both work outside the hom e and housework)

☐ 0 Does Not Interfere ☐ 1 ☐ 2 ☐ 3 ☐ 4 ☐ 5 ☐ 6 ☐ 7 ☐ 8 ☐ 9 ☐ 10 Completely Interferes

E. Relations with other people

☐ 0 Does Not Interfere ☐ 1 ☐ 2 ☐ 3 ☐ 4 ☐ 5 ☐ 6 ☐ 7 ☐ 8 ☐ 9 ☐ 10 Completely Interferes

F. Sleep

☐ 0 Does Not Interfere ☐ 1 ☐ 2 ☐ 3 ☐ 4 ☐ 5 ☐ 6 ☐ 7 ☐ 8 ☐ 9 ☐ 10 Completely Interferes

G. Enjoym ent of life

☐ 0 Does Not Interfere ☐ 1 ☐ 2 ☐ 3 ☐ 4 ☐ 5 ☐ 6 ☐ 7 ☐ 8 ☐ 9 ☐ 10 Completely Interferes

Appendix E Daily Questionnaire for Pain and Fatigue

Daily Questionnaire for Pain and Fatigue

Please answer the following questions based on how you were feeling over the past 24 hours

1. How much physical fatigue was caused when using your wheelchair around your home?

    _______ no fatigue
    _______ mild fatigue
    _______ moderate fatigue
    _______ extreme fatigue
    _______did not use wheelchair in the home today

2. How much physical fatigue was caused when using your wheelchair outside of the home?

    _______ no fatigue
    _______ mild fatigue
    _______ moderate fatigue
    _______ extreme fatigue
    _______did not use wheelchair outside of the home today

3. How much physical fatigue was caused when using your wheelchair over smooth surfaces?

    _______ no fatigue
    _______ mild fatigue
    _______ moderate fatigue
    _______ extreme fatigue
    _______did not use wheelchair today

4. How much physical fatigue was caused when using your wheelchair over uneven surfaces?

    _______ no fatigue
    _______ mild fatigue
    _______ moderate fatigue
    _______ extreme fatigue
    _______did not use wheelchair on uneven surfaces today

5. Please rate your pain on average over the past 24 hours in the following areas

Left wrist

0- ________ **No Pain**
1- ________
2- ________
3- ________
4- ________
5- ________
6- ________
7- ________
8- ________
9- ________
10- ________ **Pain as bad as you can imagine**

Right wrist

0- ________ **No Pain**
1- ________
2- ________
3- ________
4- ________
5- ________
6- ________
7- ________
8- ________
9- ________
10- ________ **Pain as bad as you can imagine**

Left elbow

0- ________ **No Pain**
1- ________
2- ________
3- ________
4- ________
5- ________
6- ________
7- ________
8- ________
9- ________
10- ________ **Pain as bad as you can imagine**

Right elbow

0- ________ **No Pain**
1- ________
2- ________
3- ________
4- ________
5- ________
6- ________
7- ________
8- ________

9- ________
10- ________ Pain as bad as you can imagine

Left shoulder

0- ________ No Pain
1- ________
2- ________
3- ________
4- ________
5- ________
6- ________
7- ________
8- ________
9- ________
10- ________ Pain as bad as you can imagine

Right Shoulder

0- ________ No Pain
1- ________
2- ________
3- ________
4- ________
5- ________
6- ________
7- ________
8- ________
9- ________
10- ________ Pain as bad as you can imagine

Upper back

0- ________ No Pain
1- ________
2- ________
3- ________
4- ________
5- ________
6- ________
7- ________
8- ________
9- ________
10- ________ Pain as bad as you can imagine

Neck

0- ________ No Pain
1- ________

2- _______
3- _______
4- _______
5- _______
6- _______
7- _______
8- _______
9- _______
10- _______ **Pain as bad as you can imagine**

6. Did you need to take any medication, patches or creams for the pain that you experienced over the last 24 hours? YES / NO
   If Yes, please provide the following information:

   What type of pain did you take this medication for? _____________________

   Indicate type of medication:
   ___Pill
   ___Patch
   ___Cream

   If Pill,
   Medication Name: ______________
   Dosage or strength (mg): ___________

   If Patch,
   Name of Patch:
   ____Fentanyl
   ____Buprenorphine
   ____Other (Name): _______________

   If Cream,
   Name of Cream:_________________

7. In general, how much did pain interfere with your daily activities over the last 24 hours?

0- _______ **No Interference**
1- _______
2- _______
3- _______
4- _______
5- _______
6- _______
7- _______

8- _______
9- _______
10- _______ Extreme Interference

8. Have you taken on any new activities (e.g. work, exercises, therapies, etc.) that may have altered your pain levels? YES / NO If YES Please describe ______________________________

Appendix F Environmental Aspects of Mobility Questionnaire

Environmental Aspects of Mobility Questionnaire (EAMQ)

Please answer the below questions by circling the number that best fits your response.

Never=1, Rarely=2, Sometimes=3, Often=4, Always=5

	Never → Always				
1. How often do you push yourself long distances (>10 blocks)?	1	2	3	4	5
2. How often do you avoid pushing yourself long distances (>10 blocks)?	1	2	3	4	5
3. How often do you push across a street with a traffic light?	1	2	3	4	5
4. How often do you avoid pushing across a street with a traffic light?	1	2	3	4	5
5. How often do you push across a busy street?	1	2	3	4	5
6. How often do you avoid pushing across a busy street?	1	2	3	4	5
7. How often do you go out when it is dark?	1	2	3	4	5
8. How often do you go out when it is snowing?	1	2	3	4	5
9. How often do you go out when it is very hot?	1	2	3	4	5
10. How often do you go out when it is very cold?	1	2	3	4	5

11. How often do you go out when it is icy?	1	2	3	4	5
12. How often do you go out when it is wet?	1	2	3	4	5
13. How often do you walk up or down a single flight of stairs?	1	2	3	4	5
14. How often do you avoid walking up or down a single flight of stairs?	1	2	3	4	5
15. How often do you go up or down escalators in your manual wheelchair?	1	2	3	4	5
16. How often do you avoid going up or down escalators in your manual wheelchair?	1	2	3	4	5
17. How often do you go up or down curbs in your manual wheelchair (i.e. you are not going up/down the curb cuts, but off the side of the sidewalk)?	1	2	3	4	5
18. How often do you avoid going up or down curbs in your manual wheelchair (i.e. you are not going up/down the curb cuts, but off the side of the sidewalk)?	1	2	3	4	5
19. How often do you push yourself over uneven surfaces?	1	2	3	4	5
20. How often do you avoid pushing yourself over uneven surfaces?	1	2	3	4	5
21. How often do you carry ≥ 2 heavy items while pushing your manual wheelchair?	1	2	3	4	5
22. How often do you avoid carrying ≥ 2 heavy items while pushing your manual wheelchair?	1	2	3	4	5
23. How often do you encounter heavy manual doors?	1	2	3	4	5
24. How often do you avoid heavy manual doors?	1	2	3	4	5

25. How often do you reach above shoulder height?	1	2	3	4	5
26. How often do you avoid reaching above shoulder height?	1	2	3	4	5
27. How often do you reach below knee height?	1	2	3	4	5
28. How often do you avoid reaching below knee height?	1	2	3	4	5
29. How often do you travel alone?	1	2	3	4	5
30. How often do you avoid travelling alone?	1	2	3	4	5
31. How often do you travel to noisy or busy places?	1	2	3	4	5
32. How often do you avoid travelling to noisy or busy places?	1	2	3	4	5
33. How often do you go to unfamiliar places?	1	2	3	4	5
34. How often do you avoid going to unfamiliar places?	1	2	3	4	5
35. How often do you go to crowded places where people might bump into you?	1	2	3	4	5
36. How often do you avoid going to crowded places where people might bump into you?	1	2	3	4	5

Appendix G International SCI Pain Basic Dataset

INTERNATIONAL SPINAL CORD INJURY PAIN BASIC DATA SET

DATA COLLECTION FORM – Version 2.0

Date of data collection: YYYY/MM/DD

Have you had any pain during the last seven days including today:
☐ No ☐ Yes

If yes:

Please note that the time period during the last week applies to all pain interference questions.

In general, how much has pain interfered with your day-to-day activities in the last week?
No interference ☐ 0 - ☐ 1 - ☐ 2 - ☐ 3 - ☐ 4 - ☐ 5 - ☐ 6 - ☐ 7 - ☐ 8 - ☐ 9 - ☐ 10 Extreme interference

In general, how much has pain interfered with your overall mood in the last week?
No interference ☐ 0 - ☐ 1 - ☐ 2 - ☐ 3 - ☐ 4 - ☐ 5 - ☐ 6 - ☐ 7 - ☐ 8 - ☐ 9 - ☐ 10 Extreme interference

In general, how much has pain interfered with your ability to get a good night's sleep?
No interference ☐ 0 - ☐ 1 - ☐ 2 - ☐ 3 - ☐ 4 - ☐ 5 - ☐ 6 - ☐ 7 - ☐ 8 - ☐ 9 - ☐ 10 Extreme interference

How many different pain problems do you have?
☐ 1; ☐ 2; ☐ 3; ☐ 4; ☐ ≥5

Please describe your three worst pain problems:

Worst pain problem:

Pain locations /sites (can be more than one, so check all that apply): right (R), midline (M), or left (L)	**R**	**M**	**L**
Head			
Neck/shoulders			
throat			
neck			
shoulder			
Arms/hands			
upper arm			
elbow			
forearm			
wrist			
hand/fingers			
Frontal torso/genitals			
chest			
abdomen			
pelvis/genitalia			
Back			
upper back			
lower back			
Buttocks/hips			
buttocks			
hip			
anus			
Upper leg/thigh			
Lower legs/feet			
knee			
shin			
calf			
ankle			
foot/toes			

Type of pain
Intensity and **duration of pain**
Treatment of pain

Type of pain (check one):

Nociceptive
☐ Musculoskeletal
☐ Visceral
☐ Other

Neuropathic
☐ At-level SCI
☐ Below-level SCI
☐ Other

☐ **Other**

☐ **Unknown**

Intensity and **duration of pain:**
Average pain intensity in the last week:
0 = no pain; 10 = pain as bad as you can imagine
☐ 0; ☐ 1; ☐ 2; ☐ 3; ☐ 4; ☐ 5;
☐ 6; ☐ 7; ☐ 8; ☐ 9; ☐ 10

Date of onset: YYYY/MM/DD

Are you using or receiving any treatment for your pain problem:
☐ No ☐ Yes

Appendix H Neck Disability Index

Neck Disability Index

Office Use Only
Name ______________________________
Date ______________________________

This questionnaire has been designed to give us information as to how your neck pain has affected your ability to manage in everyday life. Please answer every section and **mark in each section only the one box that applies to you**. We realise you may consider that two or more statements in any one section relate to you, but please just mark the box that most closely describes your problem.

Section 1: Pain Intensity

☐ I have no pain at the moment
☐ The pain is very mild at the moment
☐ The pain is moderate at the moment
☐ The pain is fairly severe at the moment
☐ The pain is very severe at the moment
☐ The pain is the worst imaginable at the moment

Section 2: Personal Care (Washing, Dressing, etc.)

☐ I can look after myself normally without causing extra pain
☐ I can look after myself normally but it causes extra pain
☐ It is painful to look after myself and I am slow and careful
☐ I need some help but can manage most of my personal care
☐ I need help every day in most aspects of self care
☐ I do not get dressed, I wash with difficulty and stay in bed

Section 3: Lifting

☐ I can lift heavy weights without extra pain
☐ I can lift heavy weights but it gives extra pain
☐ Pain prevents me lifting heavy weights off the floor, but I can manage if they are conveniently placed, for example on a table
☐ Pain prevents me from lifting heavy weights but I can manage light to medium weights if they are conveniently positioned
☐ I can only lift very light weights
☐ I cannot lift or carry anything

Section 4: Reading

☐ I can read as much as I want to with no pain in my neck
☐ I can read as much as I want to with slight pain in my neck
☐ I can read as much as I want with moderate pain in my neck
☐ I can't read as much as I want because of moderate pain in my neck
☐ I can hardly read at all because of severe pain in my neck
☐ I cannot read at all

Section 5: Headaches

☐ I have no headaches at all
☐ I have slight headaches, which come infrequently
☐ I have moderate headaches, which come infrequently
☐ I have moderate headaches, which come frequently
☐ I have severe headaches, which come frequently
☐ I have headaches almost all the time

Section 6: Concentration

☐ I can concentrate fully when I want to with no difficulty
☐ I can concentrate fully when I want to with slight difficulty
☐ I have a fair degree of difficulty in concentrating when I want to
☐ I have a lot of difficulty in concentrating when I want to
☐ I have a great deal of difficulty in concentrating when I want to
☐ I cannot concentrate at all

Section 7: Work

☐ I can do as much work as I want to
☐ I can only do my usual work, but no more
☐ I can do most of my usual work, but no more
☐ I cannot do my usual work
☐ I can hardly do any work at all
☐ I can't do any work at all

Section 8: Driving

☐ I can drive my car without any neck pain
☐ I can drive my car as long as I want with slight pain in my neck
☐ I can drive my car as long as I want with moderate pain in my neck
☐ I can't drive my car as long as I want because of moderate pain in my neck
☐ I can hardly drive at all because of severe pain in my neck
☐ I can't drive my car at all

Section 9: Sleeping

☐ I have no trouble sleeping
☐ My sleep is slightly disturbed (less than 1 hr sleepless)
☐ My sleep is mildly disturbed (1-2 hrs sleepless)
☐ My sleep is moderately disturbed (2-3 hrs sleepless)
☐ My sleep is greatly disturbed (3-5 hrs sleepless)
☐ My sleep is completely disturbed (5-7 hrs sleepless)

Section 10: Recreation

☐ I am able to engage in all my recreation activities with no neck pain at all
☐ I am able to engage in all my recreation activities, with some pain in my neck
☐ I am able to engage in most, but not all of my usual recreation activities because of pain in my neck
☐ I am able to engage in a few of my usual recreation activities because of pain in my neck
☐ I can hardly do any recreation activities because of pain in my neck
☐ I can't do any recreation activities at all

Score: ___/50 Transform to percentage score x 100 = %points

Scoring: For each section the total possible score is 5: if the first statement is marked the section score = 0, if the last statement is marked it = 5. If all ten sections are completed the score is calculated as follows: Example: 16 (total scored)
50 (total possible score) x 100 = 32%

If one section is missed or not applicable the score is calculated: 16 (total scored)
45 (total possible score) x 100 = 35.5%

Minimum Detectable Change (90% confidence): 5 points or 10 %points

---

NDI developed by: Vernon, H. & Mior, S. (1991). The Neck Disability Index: A study of reliability and validity. Journal of Manipulative and Physiological Therapeutics. 14, 409-415

Appendix I Wheelchair User Shoulder Pain Index

Participant ID ____________

WHEELCHAIR USERS SHOULDER PAIN INDEX

Place an "**X**" on the scale to estimate your level of pain with the following activities. Check box at right if the activity was not performed **in the past week.**
Based on your experiences in the past week, how much shoulder pain do you experience when:

		not performed
1. transferring from a bed to a wheelchair?	No Pain [ ] ________ Worst Pain Ever Experienced	[ ]
2. transferring from a wheelchair to a car?	No Pain [ ] ________ Worst Pain Ever Experienced	[ ]
3. transferring from a wheelchair to the tub or shower?	No Pain [ ] ________ Worst Pain Ever Experienced	[ ]
4. loading your wheelchair into a car?	No Pain [ ] ________ Worst Pain Ever Experienced	[ ]
5. pushing your chair for 10 minutes or more?	No Pain [ ] ________ Worst Pain Ever Experienced	[ ]
6. pushing up ramps or inclines outdoors?	No Pain [ ] ________ Worst Pain Ever Experienced	[ ]
7. lifting objects down from an overhead shelf?	No Pain [ ] ________ Worst Pain Ever Experienced	[ ]
8. putting on pants?	No Pain [ ] ________ Worst Pain Ever Experienced	[ ]
9. putting on a t-shirt or pullover?	No Pain [ ] ________ Worst Pain Ever Experienced	[ ]
10. putting on a button down shirt?	No Pain [ ] ________ Worst Pain Ever Experienced	[ ]
11. washing your back?	No Pain [ ] ________ Worst Pain Ever Experienced	[ ]
12. usual daily activities at work or school?	No Pain [ ] ________ Worst Pain Ever Experienced	[ ]
13. driving?	No Pain [ ] ________ Worst Pain Ever Experienced	[ ]
14. performing household chores?	No Pain [ ] ________ Worst Pain Ever Experienced	[ ]
15. sleeping?	No Pain [ ] ________ Worst Pain Ever Experienced	[ ]

Appendix J Fatigue Severity Scale

FATIGUE SEVERITY SCALE (FSS)

Date ______________________________ Name __________________________________

Please circle the number between 1 and 7 which you feel best fits the following statements. This refers to your usual way of life within the last week. 1 indicates "strongly disagree" and 7 indicates "strongly agree."

Read and circle a number.	Strongly Disagree → Strongly Agree						
1. My motivation is lower when I am fatigued.	1	2	3	4	5	6	7
2. Exercise brings on my fatigue.	1	2	3	4	5	6	7
3. I am easily fatigued.	1	2	3	4	5	6	7
4. Fatigue interferes with my physical functioning.	1	2	3	4	5	6	7
5. Fatigue causes frequent problems for me.	1	2	3	4	5	6	7
6. My fatigue prevents sustained physical functioning.	1	2	3	4	5	6	7
7. Fatigue interferes with carrying out certain duties and responsibilities.	1	2	3	4	5	6	7
8. Fatigue is among my most disabling symptoms.	1	2	3	4	5	6	7
9. Fatigue interferes with my work, family, or social life.	1	2	3	4	5	6	7

Appendix K Iowa Fatigue Scale

Iowa Fatigue Scale (IFS)

Please circle the number of the response that best indicates how you have felt in the *past month*.

	Not at all	A Little	Moderately	Quite a bit	Extremely
1. I feel worn out	1	2	3	4	5
2. I feel energetic	1	2	3	4	5
3. I feel slowed down in my thinking	1	2	3	4	5
4. I do quite a lot within a day	1	2	3	4	5
5. I have trouble concentrating	1	2	3	4	5
6. I feel drowsy	1	2	3	4	5
7. Physically I feel in good shape	1	2	3	4	5
8. I have low output	1	2	3	4	5
9. I have trouble with my memory	1	2	3	4	5
10. I feel rested	1	2	3	4	5
11. I can concentrate well.	1	2	3	4	5

Scoring
Total = Q1 + (6-Q2) + Q3 + (6-Q4) + Q5 + Q6 + (6-Q7) + Q8 + Q9 + (6-Q10) + (6-Q11)

Fatigue Cut-offs for Total Score
Fatigue = 30 – 39
Severe fatigue = 40 - 55

Subscales
Cognitive = Q3 + Q5 + Q9 + (6-Q11)
Fatigue = Q1 + Q6
Energy = (6-Q2) + (6-Q7) + (6-Q10)
Productivity = (6-Q4) + Q8

Appendix L Patient Global Impressions of Change

Participant ID _________

Patient's Global Impression of Change (PGIC) Scale

Date:

Since beginning to use the suspension wheels how would you describe the change (if any) in Activity Limitations, Symptoms, Emotions, and Overall Quality of Life? (check one box)

- 1: Very much improved
- 2: Much improved
- 3: Minimally improved
- 4: No change
- 5: Minimally worse
- 6: Much worse
- 7: Very much worse

In a similar way, please circle the number below, that matches your degrees of change since beginning to use the suspension wheels:

Much Better — No Change — Much Worse

0 1 2 3 4 5 6 7 8 9 10

Appendix M Matlab Code for Chapter 2.0 WBV

```
clear; clc; close all; warning off;
% set(0,'DefaultFigureWindowStyle','docked')
disp('Starting RMS and VDV Analysis')

% load file path directories
abs_path = "E:\";
data_path = abs_path + "IWSData\Subject Data\";

% choose subjects to analyze
% need to include subject 1 for processing, but can ignore in all analysis
subjectlist = 1:26;

% load in timepoint for special cases
specialtimepoints = load('full_exact_timepoints.mat');
timepoints = specialtimepoints.time0;

disp('Analyzing for subjects: '); disp(subjectlist);

% generate blank output tables
vertrms = zeros(3,3,3,length(subjectlist),3);
vertvdv = zeros(3,3,3,length(subjectlist),3);

% write header line for output table
rowtowrite = {'Wheel Type','Intensity','Sensor','Subject','Trial
Number','RMS','VDV'};
count = 2;

% loop through subjects
for sub = subjectlist
    disp(strcat('Analysis for subject #',string(sub)))
    if sub > 9
        subject = strcat('IWS_',num2str(sub),'\');
    else
        subject = strcat('IWS_0',num2str(sub),'\');
    end
    subject_path = data_path + subject;

    vib_path = subject_path + "Shimmer\";

    % sensor names correspond with sheet names in excel file
    sensor_list = {'Back Rest','Seat Panel','Foot Rest'};

    fs_vib = 100; %Hz

    % obtain all trial names from directory
    filenames = dir(vib_path); filenames = {filenames.name};

    disp('Starting individual trail analysis')
```

```
%     loop through filesnames (individual trials)
    for f = 3:length(filenames)
            % loop through each sensor position (sheet name)
            for s = 1:length(sensor_list)
                % determine wheel type, trial number, and sensor position
                trialname = filenames{f}(end-8:end-5);
                if strcmp(trialname(1:2),'ST')
                    wheeltype = 1;
                elseif strcmp(trialname(1:2),'CX')
                    wheeltype = 2;
                elseif strcmp(trialname(1:2),'LW')
                    wheeltype = 3;
                end
                trialnumber = str2double(trialname(end));
                sensor = sensor_list{s};



                % load in data for specific file
                data =
readtable(strcat(vib_path,filenames{f}),"Sheet",sensor,'VariableNamingRule','modify')
;
                % convert timestamp to seconds
                data.Time = (data.TimeStamp - data.TimeStamp(1))./1000; %milliseconds
to seconds

                % generate original coordinate system
                origx = [1 0 0];
                origy = [0 1 0];
                origz = [0 0 1];

                origmat = [origx; origy; origz];

                % obtain first 0.5 seconds of stationary data
                caldata = data(1:fs_vib*0.5,:);

                calaccdataraw = [(caldata.Acc_X) (caldata.Acc_Y) (caldata.Acc_Z)];

                % calculate average for each axis
                % this is the gravity vector
                calaccdata = [mean(caldata.Acc_X); mean(caldata.Acc_Y);
mean(caldata.Acc_Z)];

                % generate new coordiante system with gravity in the z axis
                % this is done with cross products
                newz = (calaccdata./(norm(calaccdata)))';
                newx = cross(origz,newz);
                newx = newx./norm(newx);
                newy = cross(newz,newx);
                newy = newy./norm(newy);

                newmat = [newx; newy; newz];

                % calculate rotation matrix from origial to gravity rotated
                rot = newmat/(origmat);
```

```
                % extract raw acceleration data
                rawaccdata = data{:,2:4};

                % rotate acceleration data to gravity
                %             rotaccdata = NaN(flip(size(rawaccdata)));
                %             for i = 1:length(rawaccdata)
                %                 rotaccdata (:,i) = rot*rawaccdata(i,:)';
                %             end

                rotaccdata = [rot*rawaccdata'];

                % extract the z axis (Vertical axis)
                vertaccdata = rotaccdata(3,:);

                %generate filter transfer function
                % includes bandpass (0.4 - 100 Hz)
                % acceleration-velocity transition
                % upward step
                [b, a] = weighting_filter();

                % filter extracted vertial acceleration data
                filtvertaccdata = filter(b,a,vertaccdata);
                %%
                % obtain index of no motion point

% figure(1)
% plot(data.Time,(vertaccdata-mean(vertaccdata)),'DisplayName','Raw Data')
% hold on
% plot(data.Time,filtvertaccdata,'DisplayName','Filtered Data')
% title('Raw vs. Filtered Vertical Acceleration')
% xlabel('Time (sec)')
% ylabel('Acceleration (m/{s^{2}})')
% legend()
% xlim([23 25])
% hold off
% pause
                idx_start = round(squeeze(timepoints(sub,wheeltype,s,trialnumber,[1 3
5])));
                idx_end = round(squeeze(timepoints(sub,wheeltype,s,trialnumber,[2 4
6])));

                if idx_end(3) > length(filtvertaccdata)
                    disp('clicked too far')
                    idx_end(3) = length(filtvertaccdata)
                end

%                 figure(2)
%                 plot(filtvertaccdata)
%                 hold on
%                 title('Intensity Region Separation')
%                 xlabel('Frames (#)')
%                 ylabel('Acceleration (m/{s^{2}})')
%                 xline(idx_start,'Color','green','LineWidth',2)
%                 xline(idx_end,'Color','red','LineWidth',2)
```

```
%                   hold off
% pause
                  % loop through intensity regions
                  for int = 1:3
                      % exytract intensity region acceleration and time

                          intvertacc = filtvertaccdata(idx_start(int):idx_end(int));
                          intverttime = data.Time(idx_start(int):idx_end(int));

                      % calculate rms and vdv

                      vertrms(wheeltype,int,s,sub,trialnumber) =
sqrt((trapz(intverttime,intvertacc.^2)/data.Time(end)));
                      vertvdv(wheeltype,int,s,sub,trialnumber) =
(trapz(intverttime,intvertacc.^4)).^(0.25);

                      % write to table row
                      rowtowrite(count,:) =
{trialname(1:2),int,sensor,sub,trialnumber,vertrms(wheeltype,int,s,sub,trialnumber),v
ertvdv(wheeltype,int,s,sub,trialnumber)};
                      count = count + 1;
                  end
              end
              disp(strcat('Done with trial ',trialname))
%            end
    end
        disp(strcat('Done with subject #',string(sub)))
    disp('Finished all subject analysis')
end
    %%
    % calculate trial averages
    trialavgrms = mean(vertrms,5);
    trialavgvdv = mean(vertvdv,5);

    % new header row for table
    newrowtowrite = {'Wheel
Type','Intensity','Sensor','Subject','AVG_RMS','AVG_VDV'};
    count = 2;

    disp('Calculating averages over trials')
    % loop through wheeltype, intensity, sensor position, and subject
    for w = 1:3
        if w == 1
            wheel = 'ST';
        elseif w == 2
            wheel = 'CX';
        elseif w == 3
            wheel = 'LW';
        end
        for i = 1:3
            for sen = 1:3
                sensor = sensor_list{sen};
                for sub = subjectlist
                    % write data to new file
```

```
                    newrowtowrite(count,:) =
{wheel,i,sensor,sub,trialavgrms(w,i,sen,sub),trialavgvdv(w,i,sen,sub)};
                    count = count + 1;
                end
            end
        end
    end
    disp('Done calculating averages')

    %%

    newnewrowtowrite = {'Subject','Sensor','Sensor ID','Trial
Number','RMS_ST_I1','RMS_ST_I2','RMS_ST_I3','RMS_CX_I1','RMS_CX_I2','RMS_CX_I3','RMS_
LW_I1','RMS_LW_I2','RMS_LW_I3','VDV_ST_I1','VDV_ST_I2','VDV_ST_I3','VDV_CX_I1','VDV_C
X_I2','VDV_CX_I3','VDV_LW_I1','VDV_LW_I2','VDV_LW_I3'};
    backrowtowrite = {'Subject','Trial
Number','RMS_ST_I1','RMS_ST_I2','RMS_ST_I3','RMS_CX_I1','RMS_CX_I2','RMS_CX_I3','RMS_
LW_I1','RMS_LW_I2','RMS_LW_I3','VDV_ST_I1','VDV_ST_I2','VDV_ST_I3','VDV_CX_I1','VDV_C
X_I2','VDV_CX_I3','VDV_LW_I1','VDV_LW_I2','VDV_LW_I3'};
    seatrowtowrite = {'Subject','Trial
Number','RMS_ST_I1','RMS_ST_I2','RMS_ST_I3','RMS_CX_I1','RMS_CX_I2','RMS_CX_I3','RMS_
LW_I1','RMS_LW_I2','RMS_LW_I3','VDV_ST_I1','VDV_ST_I2','VDV_ST_I3','VDV_CX_I1','VDV_C
X_I2','VDV_CX_I3','VDV_LW_I1','VDV_LW_I2','VDV_LW_I3'};
    footrowtowrite = {'Subject','Trial
Number','RMS_ST_I1','RMS_ST_I2','RMS_ST_I3','RMS_CX_I1','RMS_CX_I2','RMS_CX_I3','RMS_
LW_I1','RMS_LW_I2','RMS_LW_I3','VDV_ST_I1','VDV_ST_I2','VDV_ST_I3','VDV_CX_I1','VDV_C
X_I2','VDV_CX_I3','VDV_LW_I1','VDV_LW_I2','VDV_LW_I3'};

    senrowtowrite = {backrowtowrite seatrowtowrite footrowtowrite};

    count = 2;
    disp('Writing ANOVA data file')

    for sub = subjectlist
        for sen = 1:3
            sencount = 2;
            sensor = sensor_list{sen};
            for t = 1:3
                newnewrowtowrite(count,:) =
{sub,sensor,sen,t,vertrms(1,1,sen,sub,t),vertrms(1,2,sen,sub,t),vertrms(1,3,sen,sub,t
),vertrms(2,1,sen,sub,t),vertrms(2,2,sen,sub,t),vertrms(2,3,sen,sub,t),vertrms(3,1,se
n,sub,t),vertrms(3,2,sen,sub,t),vertrms(3,3,sen,sub,t),vertvdv(1,1,sen,sub,t),vertvdv
(1,2,sen,sub,t),vertvdv(1,3,sen,sub,t),vertvdv(2,1,sen,sub,t),vertvdv(2,2,sen,sub,t),
vertvdv(2,3,sen,sub,t),vertvdv(3,1,sen,sub,t),vertvdv(3,2,sen,sub,t),vertvdv(3,3,sen,
sub,t)};
                count = count + 1;
                sencount = sencount + 1;
            end
        end
    end

    for sen = 1:3
        sennewrowtowrite{sen}(1,:) = {'Subject','Trial
Number','RMS_ST_I1','RMS_ST_I2','RMS_ST_I3','RMS_CX_I1','RMS_CX_I2','RMS_CX_I3','RMS_
```

```
LW_I1','RMS_LW_I2','RMS_LW_I3','VDV_ST_I1','VDV_ST_I2','VDV_ST_I3','VDV_CX_I1','VDV_C
X_I2','VDV_CX_I3','VDV_LW_I1','VDV_LW_I2','VDV_LW_I3'};
        sencount = 2;
        for sub = subjectlist
            sensor = sensor_list{sen};
            for t = 1:3
                sennewrowtowrite{sen}(sencount,:) =
{sub,t,vertrms(1,1,sen,sub,t),vertrms(1,2,sen,sub,t),vertrms(1,3,sen,sub,t),vertrms(2
,1,sen,sub,t),vertrms(2,2,sen,sub,t),vertrms(2,3,sen,sub,t),vertrms(3,1,sen,sub,t),ve
rtrms(3,2,sen,sub,t),vertrms(3,3,sen,sub,t),vertvdv(1,1,sen,sub,t),vertvdv(1,2,sen,su
b,t),vertvdv(1,3,sen,sub,t),vertvdv(2,1,sen,sub,t),vertvdv(2,2,sen,sub,t),vertvdv(2,3
,sen,sub,t),vertvdv(3,1,sen,sub,t),vertvdv(3,2,sen,sub,t),vertvdv(3,3,sen,sub,t)};
                count = count + 1;
                sencount = sencount + 1;
            end
        end
    end

    for sen = 1:3
        avgsennewrowtowrite{sen}(1,:) =
{'Subject','RMS_ST_I1','RMS_ST_I2','RMS_ST_I3','RMS_CX_I1','RMS_CX_I2','RMS_CX_I3','R
MS_LW_I1','RMS_LW_I2','RMS_LW_I3','VDV_ST_I1','VDV_ST_I2','VDV_ST_I3','VDV_CX_I1','VD
V_CX_I2','VDV_CX_I3','VDV_LW_I1','VDV_LW_I2','VDV_LW_I3'};
        sencount = 2;
        for sub = subjectlist
            avgsennewrowtowrite{sen}(sencount,:) =
{sub,mean(vertrms(1,1,sen,sub,:),5),mean(vertrms(1,2,sen,sub,:),5),mean(vertrms(1,3,s
en,sub,:),5),mean(vertrms(2,1,sen,sub,:),5),mean(vertrms(2,2,sen,sub,:),5),mean(vertr
ms(2,3,sen,sub,:),5),mean(vertrms(3,1,sen,sub,:),5),mean(vertrms(3,2,sen,sub,:),5),me
an(vertrms(3,3,sen,sub,:),5),mean(vertvdv(1,1,sen,sub,:),5),mean(vertvdv(1,2,sen,sub,
:),5),mean(vertvdv(1,3,sen,sub,:),5),mean(vertvdv(2,1,sen,sub,:),5),mean(vertvdv(2,2,
sen,sub,:),5),mean(vertvdv(2,3,sen,sub,:),5),mean(vertvdv(3,1,sen,sub,:),5),mean(vert
vdv(3,2,sen,sub,:),5),mean(vertvdv(3,3,sen,sub,:),5)};
            sencount = sencount + 1;
        end
    end
    %%
    %save variables and output tabless
    disp('Saving variables and database')
    save('vertdata.mat','vertrms','vertvdv')
    writecell(rowtowrite,'RMSVDVAnalysis.csv','WriteMode','overwrite')
    writecell(newrowtowrite,'RMSVDVAnalysisTrialAvg.csv','WriteMode','overwrite')
    writecell(newnewrowtowrite,'RMSVDV_ANOVA.csv','WriteMode','overwrite')

writecell(avgsennewrowtowrite{1},'RMSVDV_AVG_BackANOVA.csv','WriteMode','overwrite')

writecell(avgsennewrowtowrite[32],'RMSVDV_AVG_SeatANOVA.csv','WriteMode','overwrite')

writecell(avgsennewrowtowrite{3},'RMSVDV_AVG_FootANOVA.csv','WriteMode','overwrite')
    writecell(sennewrowtowrite{1},'RMSVDV_BackANOVA.csv','WriteMode','overwrite')
    writecell(sennewrowtowrite[32],'RMSVDV_SeatANOVA.csv','WriteMode','overwrite')
    writecell(sennewrowtowrite{3},'RMSVDV_FootANOVA.csv','WriteMode','overwrite')
    disp('Operation Complete')
```

Appendix N Matlab Code for Chapter 3.0 Community WBV

```
clear;
clc;
close all;

abs_path = "E:\";
data_path = abs_path + "IWSData\";

%load data, variables
fs = 52;
onemin = 60*fs;
dt = 1/fs;
duration = 60*onemin; %window
%long term data

% subjectlist = [1:13 15:16 19];
subjectlist = [1:16 19:21 23 25];
subjectlist = [26];


for (sub = subjectlist)
    data_to_write = cell(30,1);
    data_to_write(1:5,1) = {'Year','Month','Day','DoW','Hour'};
    data_to_write(6:end-1,1) = num2cell([1:24]);
    data_to_write2 = data_to_write;

    disp(strcat('Analysis for subject #',string(sub)))
    if sub > 9
        subject = strcat('IWS_',num2str(sub),'\');
    else
        subject = strcat('IWS_0',num2str(sub),'\');
    end
    subject_path = data_path + subject;

    vib_path = subject_path + "Verisense\";

    gyro_path = vib_path + 'Gyro\';
    accel_path = vib_path + 'Accel\';

    % obtain all trial names from directory
    filenames = dir(gyro_path); filenames = {filenames.name}; filenames =
filenames(3:end);
    filenames_acc = dir(accel_path); filenames_acc = {filenames_acc.name};
filenames_acc = filenames_acc(3:end);
    filenames_accdate = cell(1,length(filenames_acc));
    for f = 1:length(filenames_acc)
        filenames_accdate{f} = filenames_acc{f}(1:13);
    end
```

```
    firstday = filenames{1}; firstday = firstday(1:6);
    day_num = 2;

    for f = 1:length(filenames)
        data_file = filenames{f};
        have_accel = 0;

        if ~strcmp(data_file(1:6),firstday)
            day_num = day_num +1;
            firstday = data_file(1:6);
        end

        data = readtable(strcat(gyro_path,data_file));

        year_num = data_file(1:2);
        year = str2double(strcat('20',year_num));
        month = str2double(data_file(3:4));
        day = str2double(data_file(5:6));

        hour = str2double(data_file(8:9));
        minute = str2double(data_file(10:11));
        second = str2double(data_file(12:13));

        daystring = strcat(year,'-',month,'-',day);

        hour_num = hour;

    %times of sensor collection -- INPUT
        starttime = posixtime(datetime(year,month,day,hour,minute,second));

        %length of trial
        time = ([0:height(data)-1]'*dt) + starttime;


    %% making bins, math etc

    %one hour bins
        binsize = 60*onemin; %one hour
        bins = [1:binsize:length(time)];
        %separate data into hours
        for i = 2:length(bins)+1
%             clear angdisp movement usedata usetime
            if i == length(bins)+1
                usedata = abs(data.Gyro_X(bins(i-1):end));
                usetime = time((bins(i-1):end));
            elseif i ~= length(bins)+1
                usedata = abs(data.Gyro_X(bins(i-1):bins(i)));
                usetime = time((bins(i-1):bins(i)));
            end

            movement = [abs(usedata) > 1.5];

            usedata([~movement]) = 0;
            %                      plot(usedata)
            % hold on
```

```
            %gyroscope --> angular displacement

            angdisp = cumtrapz(usetime,usedata);


            finaldisp =  deg2rad(angdisp(end))*0.3;

            totaltime = (sum(movement)/52)/60; %total time per in prop in minutes per
hour

            data_to_write{1,day_num} = year_num;
            data_to_write{2,day_num} = month;
            data_to_write{3,day_num} = day;
            [~,data_to_write{4,day_num}] = weekday(strcat(string(year),'-
',string(month),'-',string(day)));
            data_to_write{hour_num+6,day_num} = finaldisp;

            data_to_write2{1,day_num} = year_num;
            data_to_write2{2,day_num} = month;
            data_to_write2{3,day_num} = day;
            [~,data_to_write2{4,day_num}] = weekday(strcat(string(year),'-
',string(month),'-',string(day)));
            data_to_write2{hour_num+6,day_num} = totaltime;

            hour_num = hour_num+1;

        end
        % data_to_write{30,day_num} =
(sum([[data_to_write{6:29,day_num}]>0])/length([[data_to_write{6:29,day_num}]])).*100
;
    end


writecell(data_to_write,strcat(subject_path,'IWS_',num2str(sub),'_longitudinal_analys
is.csv'))

writecell(data_to_write2,strcat(subject_path,'IWS_',num2str(sub),'_time_longitudinal_
analysis.csv'))
end
```

Appendix O Matlab Code for Chapter 4.0 Propulsion Mechanics

```
clc; clear; close all; warning off;

c = {[68/255 114/255 196/255] [237/255 125/255 49/255] [165/255 165/255 165/255]};
wheel = {"ST" "CX" "LW"};

fc = 5;
fs = 100;

[b,a] = butter(4,fc/(fs/2),"low");

filenames = dir("C:\Users\netia\OneDrive - University of
Pittsburgh\Beyond_2023\Research\In-Wheel Suspension\Wheel Propel Testing"); filenames
= {filenames.name};
for f = 3:length(filenames)
    data{f-2} = readtable(filenames{f},"Sheet","Seat Panel");
end

for f = 1:3
    datat = data{f};
    data_z{f} = filtfilt(b,a,-(datat.Acc_Z - mean(datat.Acc_Z)));
    data_x{f} = filtfilt(b,a,(datat.Acc_X));
end

for f = 1:3
    % figure()
    datax = data_x{f};
    datax = filtfilt(b,a,datax);
    dataz = data_z{f};
    dataz = filtfilt(b,a,dataz);
    % findpeaks(datax,"MinPeakDistance",80)
    % findpeaks(dataz)
    [pks, rlocs] = findpeaks(datax,"MinPeakDistance",80);
    [pks, plocs] = findpeaks(-datax,"MinPeakDistance",80);
    % plot(datax)
    % hold on
    % xline(plocs,'Color','g')
    % xline(rlocs,'Color','r')
    % hold off
    avg_push_length(f) = mean(diff(plocs(2:end)))/100;
    push_count(f) = length(plocs)-1;
    if rlocs(1) < plocs(1)
        rlocs = rlocs(2:end);
    end
    for i = 3:length(plocs)-1
        push_range = plocs(i-1):plocs(i);

        [pks, olocs] = findpeaks(dataz(push_range));
```

```
        prop_split{f,i-2} = 100*(rlocs(i-1)-plocs(i-1))/(plocs(i)-plocs(i-1));
        antpostamp{f,i-2} = max(datax(push_range)) - min(datax(push_range));
        vertamp{f,i-2} = max(dataz(push_range)) - min(dataz(push_range));
        vertosc{f,i-2} = mean(diff(olocs(2:end)))/100;
        t = linspace(plocs(i-1),plocs(i),100);
        datax_split{f,i-2} = interp1(push_range,datax(push_range),t);
        dataz_split{f,i-2} = interp1(push_range,dataz(push_range),t);
    end
end

%%
clf;
f = figure(1);
sgtitle("Accelerations over Propulsion Cycle - 60 ppm")
speed = 1:3;
for w = speed
    pushesx = datax_split(w,:);
    pushesz = dataz_split(w,:);
    propt = prop_split(w,:);
    for f = 1:6
        pushx(w,f,1:100) = pushesx{f};
        pushz(w,f,1:100) = pushesz{f};
        prop(w,f) = propt{f};
    end
end

subplot(1,2,1)
title('Ant-Post Direction')
xlabel('% Propulsion Cycle')
ylabel('Acceleration (m_{s^{2}})')
hold on
push_avg = squeeze(mean(pushx,2));
push_std = squeeze(std(pushx,0,2));
x = 1:size(push_avg,2);
for w = speed
    leg(w) = plot(push_avg(w,:),"DisplayName",wheel{w},"Color",c{w},'LineWidth',2);
    xline(prop(w),'Color',c{w},'LineWidth',2,'LineStyle',"--","DisplayName","")
    fill([x, flip(x)], [push_avg(w,:)+push_std(w,:), flip(push_avg(w,:)-
push_std(w,:))],c{w},"FaceAlpha",0.25,"DisplayName","",'LineStyle',"none")
end
text(prop(w)-25,4.25,{"Propulsion","Phase"})
text(prop(w)+5,4.25,{"Recovery","Phase"})
legend(leg)
hold off

subplot(1,2,2)
title('Vertical Direction')
xlabel('% Propulsion Cycle')
ylabel('Acceleration (m_{s^{2}})')
hold on
push_avg = squeeze(mean(pushz,2));
push_std = squeeze(std(pushz,0,2));
x = 1:size(push_avg,2);
for w = speed
    leg(w) = plot(push_avg(w,:),"DisplayName",wheel{w},"Color",c{w},'LineWidth',2);
```

```
    xline(prop(w),'Color',c{w},'LineWidth',2,'LineStyle',"--","DisplayName","")
    fill([x, flip(x)], [push_avg(w,:)+push_std(w,:), flip(push_avg(w,:)-
push_std(w,:))],c{w},"FaceAlpha",0.25,"DisplayName","",'LineStyle',"none")
end
legend(leg)
text(prop(w)-25,-.7,{"Propulsion","Phase"})
text(prop(w)+5,-.7,{"Recovery","Phase"})
hold off
fontsize(16,'points')
```

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