INVESTIGATION OF TERRAIN EFFECTS ON WHEELCHAIR PROPULSION AND VALIDITY OF A WHEELCHAIR PROPULSION MONITOR

by

JUI-TE LIN

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This thesis was presented

by

JUI-TE LIN

It was defended on

July 19, 2011

and approved by

Rory Cooper, PhD, Rehabilitation Science and Technology

Dan Ding, PhD, Rehabilitation Science and Technology

Alicia M. Koontz, PhD, Rehabilitation Science and Technology

Thesis Advisor: Dan Ding, PhD, Rehabilitation Science and Technology

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Jui-Te Lin, M.S.

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This thesis is composed of two studies related to wheelchair propulsion biomechanics. The first study investigated the impact of cross-slope and surface roughness on wheelchair propulsion. Fifteen manual wheelchair users propelled across a five-meter platform which were set to level, 1°, or 2° cross slope, and attached with one of three surfaces including Teflon (slippery), wood (normal), and blind guide (rough). The study found main effects of both cross slope and surface roughness on stroke number and sum of work, and a main effect of cross slope on velocity. Subjects travelled slower, used more strokes, and expended more work with increasing cross slope. Subjects also used more strokes when propelling on the slippery and rough surfaces than on the level surface. They expended more work when propelling on the rough surface than on the level surface. When looking into bilateral propulsion parameters, we found that peak resultant force, peak wheel torque, and sum of work became significantly asymmetrical with the increase of cross slopes. Exposure to biomechanics loading can be reduced by avoiding slippery, rough, and cross slopes when possible. The second study consisted of a preliminary analysis on the validity of a wheelchair propulsion monitor (WPM) in estimating wheelchair propulsion biomechanics. The WPM integrates three devices including a wheel rotation datalogger, and an accelerometry-based device on the upper arm and underneath the wheelchair seat, respectively. Five wheelchair users were asked to push their own wheelchairs fitted with a SMART^{Wheel} over level and sloped surfaces on two separate visits. The estimated stroke number and cadence by the WPM were consistent with the criterion measures by the SMART^{Wheel} (ICC= 0.99 for stroke number, ICC=0.97 for cadence) with less than 5% absolute percentage errors for stroke number and 9% for cadence. The peak resultant force and wheel torque could be predicted to some extent by acceleration features on an individual subject basis. The study demonstrated the potential of the WPM in tracking wheelchair propulsion characteristics in the natural environment of wheelchair users.

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PREFACE

I am sincerely appreciated that I had an opportunity to enrich my mind and expand my professional knowledge at Human Engineering Research Laboratories. I would like to express my gratitude to the faculty and staffs at University of Pittsburgh who have made this study a success. I would especially like to thank my academic advisor, Dr. Ding, for her support, and encouragement throughout the length of the study. Her door is always open to help and answer different type of questions. I would also like to extend my appreciation for my committee members Dr. Cooper and Dr. Koontz for providing me with their assistance and time while serving on my thesis committee. I would also like to express my sincere gratitude to John Coltellaro for his guidance throughout my internship at CAT. In addition, I would like to thank all the students and staffs at HERL for their support to this study. Their support and encouragement make me feel warm and not lonely. At last, I would like to thank my family for guiding and supporting me in every step of my life. They have always been my greatest inspiration and the source of energy.

1.0 INTRODUCTION

The US census in 2002 estimated more than 2.7 million community-dwelling individuals in the United States have a disability requiring the use of a wheelchair (Erika, 2002). A majority of these individuals use manual wheelchair users as their primary means of independent mobility including individuals with spinal cord injury, spina bifida, lower-limb amputation, stroke, multiple sclerosis, rheumatoid arthritis, as well as other diagnoses (Finley, 2004). Long-term use of upper limbs for performing daily activities in manual wheelchair users has been associated with the prevalence of musculoskeletal injuries and reports of pain. Between 49 % and 73% of manual wheelchair users have experienced carpal tunnel syndrome (Aljure, 1985; Burnham R.S., 1994; Sie, 1992), and between 30% and 73% of them have experienced rotator cuff tendinopathy or shoulder pain (Ballinger, 2000; Gellman, 1988; W. E. Pentland, Twomey, L. T., 1991). Previous research has identified specific biomechanical parameters of wheelchair propulsion such as high cadence and forces associated with risk of injury to the upper limbs (Andersen, 2002; M. L. Boninger, Koontz, A. M., Sisto, S. A., Dyson-Hudson, T. A., Chang, M., Price, R., Cooper, R. A., 2005; Frost, 2002; Mercer, 2006; Roquelaure, 1997).

This thesis consisted of two studies related to wheelchair propulsion biomechanics. The first study investigated the impact of cross slope and surface roughness on wheelchair propulsion biomechanics. The second study was a preliminary evaluation of the validity of a wheelchair propulsion monitor in estimating key biomechanical parameters of wheelchair propulsion. The thesis will provide insights into the environmental impact on wheelchair propulsion and contribute to a potential tool that can track propulsion characteristics in the natural environment of wheelchair users.

2.0 INVESTIGATION OF THE TERRAIN EFFECTS ON WHEELCHAIR PROPULSION

2.1 ABSTRACT

Certain surface characteristics may act as barriers to wheelchair propulsion. To what extent these surfaces impact stresses on the upper extremities of manual wheelchair users (MWUs) is unclear. The purpose of this study is to examine the impact of cross-slope and surface roughness on wheelchair propulsion. Fifteen MWUs propelled over a five-meter platform which were set to level, 1°, or 2° cross slope, and attached with one of three surfaces including Teflon (slippery), wood (normal), blind guide (rough). The study found that subjects travelled slower, used more strokes, and had greater amount of work with increasing cross slope. Subjects also used more strokes when propelling on the slippery and rough surfaces than on the level surface. In addition, we found that resultant force, wheel torque, and sum of work became significantly asymmetrical with increasing cross slope. The study indicates that small cross slopes ($\leq$ 2°), slippery, and rough surfaces could result in increased repetitiveness of upper-extremity motion and the amount of total work, as well as unbalanced effort between two upper extremities. Long-term exposure to such terrains should be minimized when possible to reduce the risk of injury.

KEY WORDS

Manual Wheelchair, Propulsion, Biomechanics, Cross-Slope, Terrain, Surface roughness.

2.2 INTRODUCTION

Manual wheelchairs are widely used for people with mobility impairments to participate in community (R. E. Cowan, Boninger, M. L., Sawatzky, B. J., Mazoyer, B. D., Cooper, R. A., 2008; Kilkens, 2005). Manual wheelchair users likely traverse a variety of surfaces such as sloped, cross-sloped, slippery, and rough surfaces. Surface characteristics may facilitate or hinder the ability of manual wheelchair users in propelling their wheelchairs and participating in community (A. M. Koontz, Roche, B. M., Collinger, J. L., Cooper, R. A., Boninger, M. L., 2009; Richter, 2007). For example, flat and smooth surfaces usually allow for greater ease of propulsion, while sloped and uneven surfaces may create potential barriers to manual wheelchair propulsion (R. A. Cooper, Teodorski, E. E., Sporner, M. L., Collins, D. M., 2011; Kilkens, 2005; Meyers, 2002). In this study, we looked into two surface characteristics including cross slope and surface roughness, which are frequently found throughout our community and regarded as being more difficult to traverse than a regular level surface (R. E. Cowan, Nash, M. S., Collinger, J. L., Koontz, A. M., Boninger, M. L., 2009; Hurd, 2008a, 2008b; A. M. Koontz, Cooper, R. A., Boninger, M. L., Yang, Y., Impink, B. G., Van der Woude, L. H., 2005; Richter, 2007).

Cross-slope is a transversal slope with respect to the horizon (Kockelman, 2001; Richter, 2007). It is a common design feature in roads and sidewalks for promoting water drainage in daily environment. According to the specifications in the Americans with Disability

Accessibility (ADA) Act (ADA, 2009), the accessible routes to a building, including sidewalks, ramps, and parking spaces should have cross-slopes no greater than 1:50 (i.e., 1.15°). A recent literature review on manual wheelchair propulsion over cross-sloped surfaces conducted by Cooper et al. only found six studies relevant to propulsion over cross-slopes (R. A. Cooper, Teodorski, E. E., Sporner, M. L., Collins, D. M., 2011) including three experimental design (Brubaker, 1986; Chesney, 1996; Richter, 2007) and three survey studies (Kara K., 2002; Kockelman, 2001; Longmuir, 2003). Richter et al. examined 26 manual wheelchair users as they propelled their wheelchairs on a treadmill set to level, 3°, and 6° cross slopes. Using an instrumented test wheel, they found that force, moment, and power were linearly related to the degree of cross-slope. Neither the push angle nor the push frequency was affected by traversing cross-slopes (Richter, 2007). The net distance traveled per push was found to be significantly decreased on cross-sloped surfaces, requiring individuals to push harder and to increase their number of pushes to cover the same distance (Richter, 2007). Brubaker et al. conducted an experiment with a single subject and found that the total drag force was roughly doubled on a 2° cross-slope treadmill due to the downward turning moment of the wheelchair, and the net oxygen consumption of propulsion on the cross-slope was 30% greater than on a level surface (Brubaker, 1986). Chesney et al. examined one able-body subject to traverse surfaces of varying firmness in different configurations ranging from a 1°-14° ramp and a 1°-11° cross-slope. The results indicated that both running slope and cross-slope were significantly correlated with forces applied to the pushrim. However, running slope had a stronger effect than cross-slope (Chesney, 1996). Hurd et al. recruited 12 manual wheelchair users to evaluate upper-extremity symmetry during wheelchair propulsion across multiple terrain surfaces including a 2° outdoor cross slope. The result demonstrated that wheelchair propulsion asymmetry was significantly greater in outdoor community than during laboratory conditions and there was a significant asymmetry pattern on forces, moments, and work while propelling on a 2° cross slope (Hurd, 2008a). Most studies are in agreement that traversing a cross-slope in a manual wheelchair is more difficult than propelling on a level surface (Brubaker, 1986; Chesney, 1996; Kara K., 2002; Kockelman, 2001; Longmuir, 2003; Richter, 2007). However, agreement has not been reached as to the percentage increase in effort or the optimal degree of cross-slope that should be used as a maximum acceptable standard (R. A. Cooper, Teodorski, E. E., Sporner, M. L., Collins, D. M., 2011). This study is designed to evaluate the impact of small cross slopes (i.e., 1° and 2°) that are close to the ADA standard (i.e., 1.15°) and will provide insights into the impact of the ADA standard on wheelchair propulsion mechanics.

In addition to cross-sloped surfaces, manual wheelchair users may also encounter different surface conditions, such as grass, gravel, mud, and those that are wet or snowed covered are considered obstacles for manual wheelchair propulsion (Meyers, 2002). However, the impact of surface roughness in terms of slippery or uneven surfaces on propulsion mechanics is not well understood. Hurd et al. asked 14 manual wheelchair users to propel their wheelchairs at a self-selected speed on a variety of surfaces. The study found that propelling across aggregate concrete had 37%-50% greater kinetic values (i.e. propulsion frequency, forces and moments) than across tiled floor surface, and 20%-25% greater than the smooth concrete and carpeted surfaces (Hurd, 2008b). Koontz et al. conducted a kinetic analysis of wheelchair propulsion during start-up on a series of indoor and outdoor surfaces with 11 manual wheelchair users. They reported that running slope, grass, and interlocking pavers required greater forces and wheel torques than indoor tile, wood, smooth level concrete, and high- and low-pile carpet (A. M. Koontz, Cooper, R. A., Boninger, M. L., Yang, Y., Impink, B. G., Van der Woude, L. H., 2005).

Cowan et al. asked 53 elders to propel a wheelchair with different added weights and axle positions over different surfaces. They found that participants decreased self-selected speed and increased propulsion forces as rolling resistance of the surface increased. The ramped condition was traversed at the slowest velocity using the highest forces, lowest push frequency, and shortest stroke length compared with tile, low and high carpet (R. E. Cowan, Nash, M. S., Collinger, J. L., Koontz, A. M., Boninger, M. L., 2009). All these studies pointed out the importance of evaluating wheelchair propulsion over a range of surfaces, yet little information is available on propulsion mechanics under slippery or uneven surfaces.

Even with the growing number of studies on manual wheelchair propulsion biomechanics, only a few studies have recently emerged that have investigated propulsion in the natural environment, and the different surfaces examined remain quite limited (R. A. Cooper, Teodorski, E. E., Sporner, M. L., Collins, D. M., 2011). The purpose of this study is to examine the impact of cross slope and surface roughness on kinetic characteristics and bilateral demands of over ground wheelchair propulsion. The findings of this study are expected to provide evidence for defining or refining pathway accessibility, and contribute to the knowledge base of environmental impact on upper extremity loading among manual wheelchair users.

2.3 METHODS

The study was conducted during the 2009 National Veteran Wheelchair Games (NVWG) in Spokane, WA. Any veteran who used a wheelchair for independent mobility was eligible to participate. The events available for participants to compete in at the NVWG range from low to high intensity. The study was approved by the Department of Veterans Affairs (VA) National

Special Events Committee, the local Pittsburgh VA Research and Development Committee, the VA Human Studies Subcommittee, and the University of Pittsburgh's Institutional Review Board.

2.3.1 Study Participants

A convenience sample of 15 manual wheelchair users participated in the study. Subject recruitment was conducted by study personnel at the NVWG-sponsored exposition and different event venues. Subjects were included if they were between 18 and 70 years old and used a manual wheelchair as a primary means of mobility. To be eligible for participation in the NVWG, all participants underwent a medical examination and obtained clearance from a physician. All participants provided a written informed consent prior to participating in the study.

2.3.2 Experimental Protocol

Subjects were asked to complete a brief demographic questionnaire. They will then be asked to participate in the study using their own wheelchair. The wheels of their own wheelchair were replaced with two SMART^{Wheel} s (Three River Holdings, Mesa, Arizona) which collect propulsion kinetics in 6 degrees of freedom (R. A. Cooper, Robertson, R. N., VanSickle, D. P., Boninger, M. L., Shimada, S. D., 1997). The use of SMART^{Wheel} s did not change the camber, axle position, and diameter of the subject's rims. Each SMART^{Wheel} had a solid treaded tire and weighted around 4.98 kg (Figure 1). A five-meter wood platform with fixtures to set cross slopes at 1° and 2°, and attach different types of surfaces was used as the experimental course. Each subject was asked to perform nine trials on the platform (Figure 2) when it was configured to one of three surfaces (i.e., wood, blind guide, and Teflon drizzled with soapy water simulating normal, rough, and slippery road conditions, respectively) at level, 1°, or 2° cross-slope. The order of surface roughness and cross slope was randomized for each participant. Participants were instructed to start propelling their wheelchair straight from a resting position up to a comfortable pace until they reached the designated finish line. Data collection for all trials was initiated before initial hand-to-rim contact, and terminated before the wheelchair reached the marked finish line. There was a 5-minute rest period between each trial while the surface type or cross slope was configured.

Figure 1 SMART$^{\text{Wheel}}$

Figure 2 Propulsion platform with blind guide surface and 2° cross slope

2.3.3 Data Collection and Reduction

All data were collected at 240Hz via a Security Digital card on the SMART^{Wheel} (Cooper R.A., 1998; R. A. Cooper, Robertson, R. N., VanSickle, D. P., Boninger, M. L., Shimada, S. D., 1997). The data were then filtered and converted to a readable format with the SMART^{Wheel} software. The SMART^{Wheel} sign convention follows the right hand rule, with positive "x" forward, positive "y" up, and positive "z" point out of the wheel along the axle (R. A. Cooper, Robertson, R. N., VanSickle, D. P., Boninger, M. L., Shimada, S. D., 1997; A. M. Koontz, Cooper, R. A., Boninger, M. L., Yang, Y., Impink, B. G., Van der Woude, L. H., 2005). Positive moments were defined as counterclock-wise about the respective force vector. A stroke was defined as a propulsive contact. A cycle was defined as the period encompassing a propulsive contact and the subsequent recovery (R. E. Cowan, Nash, M. S., Collinger, J. L., Koontz, A. M., Boninger, M. L., 2009). Identification of contact and recovery phase was automatically recognized by a search algorithm and verified by visual inspection. The first cycle from a stationary position and the last cycle when participants approached the finish line were trimmed prior to key variable computation.

The following biomechanical variables were calculated or directly obtained from the SMART^{Wheel} bilaterally including stroke number, average velocity, push frequency, push angle, peak resultant force, peak wheel torque, and sum of work. These variables were calculated for each cycle and then averaged for each trimmed trial to provide a general representation of propulsion. Average velocity was the average linear velocity of the wheel during the cycle derived from the onboard encoder of the SMART^{Wheel}. Push frequency was calculated as 1/cycle time. Push angle (θ) was defined as the angular distance (degrees) traveled by the wheel during the propulsive moment portion of a contact. Resultant force (F_R) was defined as the vector sum of Fx, Fy, and Fz (Equation 1). Wheel moment (Mz) was defined as the moment along the axis of rotation responsible for angular acceleration of the wheel. Sum of work was calculated using Equation 2. A symmetry index for each variable was calculated by dividing the downhill side by the uphill side. A custom MATLAB program (Version 7.10 R2010a, The Mathworks Inc. MA, USA) was used to reduce the data, identify cycles, and compute biomechanical variables as described above.

Equation 1 Peak Resultant force

$$\text{Peak resultant force } F_R = \sqrt{Fx^2 + Fy^2 + Fz^2}$$

Equation 2 Sum of Work

$$\text{Sum of Work (J)} = \int Mz \; d\theta$$

2.3.4 Statistical Analysis

Statistics analysis was completed using SPSS statistical software (ver. 18.0, SPSS Inc. IL., USA). Distributions of variables were examined and transformations were made where necessary. To determine the impact of cross slope and surface roughness, each biomechanical variable on the downhill side was compared using a 3 (cross slope) × 3 (surface roughness) repeated-measures analysis of variance (ANOVA). Propulsion symmetry was also evaluated for each variable with a 3 (cross slope) × 3 (surface roughness) repeated-measures ANOVA. When significant main effects or interaction effect were found, post-hoc pairwise comparisons were performed using the Bonferroni adjustment to evaluate differences between conditions. To control for Type I error caused by multiple comparisons, α level for significance was adjusted at .01.

2.4 RESULTS

2.4.1 Participants

Fifteen subjects participated in this study including nine men and six women with an average age of 48 ± 9 years old. Nine of the 15 subjects had a spinal cord injury (SCI) ranging from L5/S1 to C6/7. Three subjects had multiple sclerosis and three subjects had lower extremity amputation. The number of years subjects have experienced disability was 17±10 years. All subjects used customized ultra-light wheelchairs (K0006) during the testing and all of them were able to complete the protocol.

2.4.2 Propulsion Biomechanics

The bilateral biomechanical variables (i.e., stroke number, average velocity, push frequency, push angle, peak resultant force, peak wheel torque, and sum of work) and their symmetry index are reported in Table 1 to Table 7. There was no main effect of surface roughness on push frequency ($F (2, 13) = .10$, $p = .903$, partial $\eta 2 = .02$), push angle ($F (2, 13) = 2.20$, $p = .150$, partial $\eta 2 = .25$), and velocity ($F (2, 13) = 2.60$, $p = .112$, partial $\eta 2 = .29$). There was no main effect of cross slope on push frequency ($F (2, 13) = .94$, $p = .41$, partial $\eta 2 = .13$), peak resultant force ($F (2, 13) = 2.04$, $p = .169$, partial $\eta 2 = .24$), and peak wheel torque ($F (2, 13) = 2.88$, $p = .092$, partial $\eta 2 = .31$). There was also no interaction effect between the two factors on all variables.

There was a main effect of cross slope on average velocity ($F (2, 13) = 17.37$, $p < .001$, partial $\eta 2 = .73$), stroke number ($F (2, 13) = 16.43$, $p < .001$, partial $\eta 2 = .72$) and sum of work ($F (2, 13) = 20.66$, $p < .001$, partial $\eta 2 = .71$). There was also a trend of main effect of cross slope on push angle ($F (2, 13) = 5.63$, $p = .017$, partial $\eta 2 = .46$). Pairwise comparisons indicated that subjects significantly decreased their propulsion speed on the 2° cross-slope (M= .70 m/s) than on the level surface (M= .80 m/s). Subjects also tended to reduce their push angle on the 2° cross-slope (M= 69.77) than on the level surface (M= 73.10). With the increase of cross-slope angles, stoke number significantly increased (M= 5.76 on the level surface, M= 6.36 on the 1° cross slope, and M= 7.38 on the 2° cross slope). Subjects also expended more work with increasing cross slope (M=84.02 J on the level surface, M=98.28 J on the 1° cross-slope, and M= 116.62 J on the 2° cross-slope).

There was a main effect of surface roughness on stroke number ($F (2, 13) = 17.58$, $p < .001$, partial $\eta 2 = .73$) and sum of work ($F (2, 13) = 8.70$, $p = .004$, partial $\eta 2 = .57$). There was a trend of main effect of surface roughness on peak resultant force ($F (2, 13) = 4.68$, $p = .029$, partial $\eta 2$ = .42) and peak wheel torque (F (2, 13) = 6.65, p= .010, partial $\eta 2$ = .51). Pairwise comparisons indicated that subjects pushed their wheelchair with more strokes on blind-guide strips (M= 6.91) and Teflon (M= 7.00) than wood surface (M= 5.58). Furthermore, subjects expended more work on blind-guide strips (M= 111.04 J) than wood surface (M= 90.66 J). Subjects also tended to push with less force and torque on Teflon (M= 102.34 N, M=20.99 Nm) than on blind-guide strips (M= 109.10 N, M=22.86 Nm) and wood (M= 110.00 N, M=23.21 Nm) surfaces.

In terms of propulsion symmetry, there was no main effect of surface roughness on all variables, and no main effect of cross slope on some variables including stroke number, velocity, push frequency, and push angle. There was also no interaction effect between surface roughness and cross slope on all variables. The result showed that peak resultant force (F (2, 13) = 10.52, p= .002, partial $\eta 2$ = .62), peak wheel torque (F (2, 13) = 8.47, p= .002, partial $\eta 2$ = .57), and sum of work (F (2, 13) = 13.27, p= .001, partial $\eta 2$ = .67) had significantly asymmetric pattern with increasing cross slope. On the 2° cross-slope, symmetry index for peak resultant force (M= 1.16), peak wheel torque (M= 1.28), and sum of work (M=1.61) are significantly greater than those on the level surface (M= 1.00 for peak resultant force; M= 1.06 for peak wheel torque; M= 1.10 for sum of work). Furthermore, on the 1° cross-slope subjects had significantly asymmetric work consumption (M= 1.30) than on the level surface (M=1.10).

Table 1 Stroke Number (SN), Mean (SD)

	Cross Slope 0°			*Cross Slope 1°*			*Cross Slope 2°*		
	Right	Left	SI	Right	Left	SI	Right	Left	SI
Blind	6.20	6.27	1.01	6.87	6.47	1.09	7.67	7.27	1.07

	(2.00)	(2.22)	(0.15)	(2.50)	(2.53)	(0.19)	(3.31)	(3.01)	(0.20)
Teflon	6.13	5.60	1.07	6.73	6.47	1.05	8.13	7.07	1.15
	(2.75)	(1.84)	(0.23)	(2.40)	(2.17)	(0.18)	(3.16)	(2.31)	(0.28)
Wood	4.93	4.73	1.04	5.47	5.33	1.03	6.33	5.67	1.14
	(2.60)	(1.87)	(0.25)	(2.44)	(2.02)	(0.28)	(2.35)	(2.06)	(0.27)

Table 2 Velocity (m/s), Mean (SD)

	Cross Slope 0°			*Cross Slope 1°*			*Cross Slope 2°*		
	Right	Left	SI	Right	Left	SI	Right	Left	SI
Blind	0.82	0.80	1.03	0.78	0.75	1.04	0.69	0.68	1.01
	(0.21)	(0.21)	(0.05)	(0.22)	(0.17)	(0.09)	(0.21)	(0.17)	(0.08)
Teflon	0.78	0.76	1.02	0.70	0.67	1.04	0.67	0.63	1.07
	(0.23)	(0.18)	(0.15)	(0.17)	(0.15)	(0.04)	(0.19)	(0.17)	(0.10)
Wood	0.79	0.78	1.01	0.77	0.76	1.04	0.72	0.70	1.04
	(0.11)	(0.10)	(0.09)	(0.14)	(0.20)	(0.21)	(0.23)	(0.23)	(0.11)

Table 3 Push Frequency (sec^{-1}), Mean (SD)

	Cross Slope 0°			*Cross Slope 1°*			*Cross Slope 2°*		
	Right	Left	SI	Right	Left	SI	Right	Left	SI
Blind	1.08	1.08	1.01	1.09	1.07	1.04	1.06	1.04	1.03
	(0.21)	(0.26)	(0.09)	(0.22)	(0.25)	(0.07)	(0.20)	(0.21)	(0.09)
Teflon	1.11	1.06	1.05	1.06	1.02	1.04	1.07	1.02	1.06
	(0.26)	(0.26)	(0.11)	(0.24)	(0.24)	(0.08)	(0.20)	(0.26)	(0.11)

Wood	1.09	1.12	0.98	1.05	1.05	1.01	1.05	1.07	0.98
	(0.26)	(0.22)	(0.15)	(0.22)	(0.22)	(0.09)	(0.27)	(0.29)	(0.13)

Table 4 Push Angle (deg), Mean (SD)

	Cross Slope 0°			*Cross Slope 1°*			*Cross Slope 2°*		
	Right	Left	SI	Right	Left	SI	Right	Left	SI
Blind	73.79	72.21	1.03	72.41	70.69	1.03	70.19	64.18	1.10
	(14.53)	(15.74)	(0.05)	(13.31)	(14.86)	(0.05)	(19.07)	(16.32)	(0.15)
Teflon	71.25	71.30	1.00	68.54	65.65	1.06	67.92	63.56	1.12
	(15.43)	(15.47)	(0.09)	(11.08)	(13.20)	(0.08)	(12.92)	(18.80)	(0.23)
Wood	74.27	73.94	1.02	74.20	72.64	1.04	71.20	65.97	1.11
	(13.63)	(13.65)	(0.18)	(17.15)	(19.50)	(0.14)	(15.98)	(19.48)	(0.15)

Table 5 Peak Resultant Force FR (N), Mean (SD)

	Cross Slope 0°			*Cross Slope 1°*			*Cross Slope 2°*		
	Right	Left	SI	Right	Left	SI	Right	Left	SI
Blind	106.10	107.18	1.01	107.03	103.58	1.06	114.16	100.94	1.16
	(28.67)	(29.91)	(0.20)	(27.97)	(27.31)	(0.23)	(28.97)	(21.15)	(0.29)
Teflon	97.19	103.23	0.99	102.09	99.02	1.13	107.74	99.25	1.13
	(26.36)	(32.27)	(0.28)	(30.11)	(35.98)	(0.33)	(34.47)	(31.59)	(0.33)
Wood	105.38	110.24	1.00	110.76	104.07	1.09	113.86	102.98	1.17
	(29.87)	(35.20)	(0.27)	(35.78)	(32.84)	(0.30)	(32.60)	(33.85)	(0.39)

Table 6 Peak Wheel Torque Mz (Nm), Mean (SD)

	Cross Slope 0°			*Cross Slope 1°*			*Cross Slope 2°*		
	Right	Left	SI	Right	Left	SI	Right	Left	SI
Blind	21.14	20.17	1.06	22.82	20.19	1.14	24.60	19.00	1.31
	(8.13)	(6.67)	(0.25)	(7.28)	(5.93)	(0.25)	(7.69)	(4.28)	(0.34)
Teflon	19.88	19.69	1.03	20.83	18.71	1.17	22.25	18.73	1.25
	(7.03)	(5.35)	(0.32)	(7.27)	(6.63)	(0.41)	(8.07)	(6.56)	(0.48)
Wood	21.94	21.08	1.09	23.56	19.97	1.25	24.13	19.92	1.28
	(8.37)	(7.03)	(0.38)	(7.32)	(6.93)	(0.46)	(8.43)	(6.77)	(0.50)

Table 7 Sum of Work (J), Mean (SD)

	Cross Slope 0°			*Cross Slope 1°*			*Cross Slope 2°*		
	Right	Left	SI	Right	Left	SI	Right	Left	SI
Blind	96.47	88.69	1.10	111.91	85.92	1.33	124.73	82.97	1.53
	(35.11)	(27.40)	(0.28)	(31.67)	(28.64)	(0.26)	(38.69)	(20.99)	(0.44)
Teflon	81.58	79.17	1.09	92.34	77.39	1.28	117.72	79.10	1.66
	(25.63)	(25.45)	(0.41)	(24.01)	(26.75)	(0.45)	(40.54)	(33.95)	(0.82)
Wood	73.99	70.14	1.10	90.59	72.03	1.29	107.40	68.16	1.65
	(26.50)	(21.81)	(0.41)	(31.46)	(21.61)	(0.37)	(34.76)	(17.70)	(0.67)

Note: Abbreviation: SD, Standard Deviation; SI, Symmetry Index.

2.5 DISCUSSION

The result of this investigation provides insight into how wheelchair propulsion mechanics is influenced by terrain features such as cross slope and surface roughness. Though these two terrain features were found not to interact with each other, they individually influenced the way wheelchair users propelled their wheelchairs. In this study, we chose to study small cross slopes (1° and 2°) with the purpose of investigating the appropriateness of the cross slope standard specified in the ADA Accessibility Guidelines. We found that participants had to use 10% more strokes on the 1° cross slope or 28% more strokes on the 2° cross slopes than on the level surface. Due to the increased stroke number with increasing cross slope, subjects also expended 17% more work on the 1° cross slope or 39% more work on the 2° cross slopes than on the level surface. In terms of propulsion force and moment, subjects tended to increase force and moment with increasing cross slope, however, the differences were not statistically significant due to the small sample size. When looking into the effect size, we noticed that subjects increased the peak resultant force and wheel moment on the 1° cross slope by about 3.6% (~3.7 N) and 6.7% (~1.4 Nm), respectively, and on the 2° cross slope by about 8.7% (~9.0 N) and 12.7% (~2.7 Nm), respectively. Even with more subjects being tested, we will be unlikely to detect statistically significant differences at greater effect size than aforementioned above. Unfortunately, there are no literatures or evidence suggesting the clinical impact of this magnitude of increase in peak resultant force and/or wheel torque. Rice et al. conducted a study where they developed a wheelchair propulsion training program and tested it on a case subject. At a self-selected speed, the case subject decreased mean resultant force by about 5.5N (I. Rice, Gagnon, D., Gallagher, J., Boninger, M., 2010). Richter et al. investigated the impact of medium to large cross slopes on wheelchair propulsion biomechanics and found that subjects pushed with significantly greater forces on the 3° and 6° cross-slopes by a factor of 1.2 and 1.4, respectively. They also found that the peak handrim force increases by an average of 3.9 N for each degree of cross slope (Richter, 2007), which is relatively consistent with our results even though their study was conducted on a treadmill course instead of over ground. In general, our study found that when traversing small gradient cross slopes, subjects chose to decrease their speeds, but maintain their push frequency. As a result, they had to increase the number of strokes and expended more work to cover the same distance. A number of ergonomic studies have strongly implicated frequency of task completion as a risk factor for repetitive strain injury or pain at the wrists (Loslever, 1993; Silverstein, 1987; Werner, 1998) and shoulder (Andersen, 2002; Cohen, 1998; Frost, 2002). The Clinical Practice Guideline on Preservation of Upper Extremity Function Following Spinal Cord Injury also recommends reducing the frequency of repetitive upper limb tasks and minimizing forces required to complete tasks (M. L. Boninger, Waters, R.L., Chase, T., Dijkers, M.P., Gellman, H., Gironda, R.J., Goldstein, B., Johnson-Taylor, S., Koontz, A., McDowell, S.L., 2005). Even though the changes in stroke number and propulsion force on small cross slopes were of relatively small magnitudes, the increased repetitiveness of upper-extremity motion and the amount of total work warrant attention if subjects are constantly exposed in such terrain conditions.

In terms of surface roughness, the results of significantly more strokes required on the rough and slippery surface and non-significant change on push frequency and velocity indicated that subjects tended to reduce the effective distance per push to compensate for the terrain change. The rough surface (blind-guide strips) made it difficult for users to travel in straight lines due to increased vibration, which resulted in 24% more strokes and 22% more overall work effort than the wood surface. The slippery surface (Teflon) caused wheelchairs to slip and reduced the effective distance travelled, which led to 20% more strokes than the wood surface. As subjects also tended to reduce propulsion force and moments to gain more control on the slippery surface, the overall work expended was increased but not statistically significant from the work on the wood surface. Again, the increased repetitiveness of upper-extremity motion warrants some attention if subjects need to traverse rough or slippery surfaces on a regular basis.

This study also examined the terrain impact on propulsion symmetry between two upper extremities. With increasing cross slope, the downhill side needed greater resultant force, wheel torque, and sum of work than the uphill side to accommodate the force pulling the wheelchair down the cross slope. The results indicate that propelling on the 2° cross-slope caused 16% difference on resultant force, 21% difference on wheel torque, and 46% difference on sum of work. Although not statistically significant, there is a trend that the downhill side needs more strokes, higher speed, and greater push angle than the uphill side. The findings were consistent with Hurd et al.'s study where they found significant side-to-side differences on propulsion moment, total force, tangential force, fractional effective force, time-to-peak propulsion moment, work, length of push cycle and power (under biomechanics laboratory and the general community setting) on the 2° cross-slope (Hurd, 2008a). Different from Hurd et al.'s study which also found significant propulsion asymmetry on several level surfaces such as smooth concrete, aggregate concrete, and outdoor ramp (Hurd, 2008a), our study found that the magnitude of propulsion asymmetry was dependent on cross-slopes only.

The push frequency, peak resultant force, and peak wheel moment in our study were higher than those in several previous studies on wheelchair propulsion biomechanics across different terrain conditions (Hurd, 2008a, 2008b). These studies used longer propulsion courses (10-30 meters) as opposed to the 5-meter course in our study (Hurd, 2008a, 2008b) and thus were able to examine the steady state performance of wheelchair propulsion. Our study was more of the start-up phases of propulsion where subjects tend to push faster and harder. Koontz et al. found that force and torque during start-up for all surfaces tested in their study were considerably higher compared with steady-state propulsion on a smooth level surface (A. M. Koontz, Cooper, R. A., Boninger, M. L., Yang, Y., Impink, B. G., Van der Woude, L. H., 2005).

2.6 LIMITATION

Although differences were noted in some biomechanical variables such as peak resultant force and wheel moment, they were not statistically significant due to the small sample size. Our subject population may not be representative of the general manual wheelchair user population, as they were recruited at the National Veterans Wheelchair Games where participants are likely to be physically active even though they are not professional athletes. In addition, the experimental course was too short to yield steady-state propulsion. Another limitation of the study is that we excluded the weight and axle position from the analysis. Based on previous studies that the weight distribution could significantly affect rolling resistance and pushing force (M. L. Boninger, Baldwin, M., Cooper, R. A., Koontz, A., Chan, L., 2000; M. L. Boninger, Souza, A. L., Cooper, R. A., Fitzgerald, S. G., Koontz, A. M., Fay, B. T., 2002; R. E. Cowan, Nash, M. S., Collinger, J. L., Koontz, A. M., Boninger, M. L., 2009). The magnitude of change might not be able to be compared between subjects. In addition, properly balanced centered of gravity controlled by axle position, camber, and steerable casters might reduce the downhill moment (Brubaker, 1986; Richter, 2007). The anterior configuration required less force than the posterior configuration on all type of surfaces (R. E. Cowan, Nash, M. S., Collinger, J. L., Koontz, A. M., Boninger, M. L., 2009). Looking into our study, all participants in the wheelchair game used ultra-light wheelchair that the performance of propulsion could be improved by customized adjustment. The result of this study might not be able to generalize the whole population of wheelchair users. Future work will focus on larger and more diverse groups of manual wheelchair users that allow us to compare the impact of cross slope across different types and levels of diagnoses. Also the protocol could be revised to include longer experimental courses and more realistic surface conditions.

2.7 CONCLUSION

Overall, this study shed a light on the relationship between different surface conditions and wheelchair propulsion mechanics. The ADA guideline for cross slope at 1.15° seems reasonable with relatively small increases in stroke number and propulsion force. However, while increasing to 2° cross slope, the biomechanical demands become greater and the increased repetitiveness of the upper limb motion and unbalanced efforts between the two upper limbs become more apparent. Rough or slippery surfaces also demand increased repetitiveness of the upper limb motion to compensate the decreased effective travel distances. As small cross slopes and slippery or rough surface are a part of everyday propulsion environments, the observed changes in propulsion biomechanics in this study should be considered in the prevention of upper limb pain and injury from daily overuse. The findings of the study may also help design better community and home in terms of facilitating pathway accessibility and minimizing propulsion demands.

2.8 ACKNOWLEDGE

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3.0 VALIDITY OF A WHEELCHAIR PROPULSION MONITOR (WPM)

3.1 ABSTRACT

Objective: To examine the validity of a wheelchair propulsion monitor (WPM) in estimating selected wheelchair propulsion biomechanics.

Design: One-group pretest-posttest design.

Setting: Biomechanics laboratory.

Participants: Convenience sample of 5 manual wheelchair users with SCI with an average age of 38±16 y/o; men, n= 3; women, n=2.

Intervention: Participants were recruited for a two-visit protocol where they pushed their own wheelchairs fitted with a SMART^{Wheel} at self-selected speed over level and sloped surfaces at the first visit, at the beginning of the second visit, and after a wheelchair propulsion training session at the second visit. A WPM comprised of three devices was attached to the subject's upper arm, underneath the wheelchair seat, and the wheelchair wheel to record upper limb motion and wheelchair movement.

Main Outcome Measures: Criterion biomechanical variables including stroke number, cadence, resultant force, and wheel torque were obtained via the SMART^{Wheel}. Estimated stroke number and cadence were calculated based on upper arm accelerations recorded by the WPM. Acceleration features of the upper arm derived from the WPM including resultant and three directions of acceleration, their standard deviations, and acceleration peak phase were used to predict propulsion force.

Results: The estimated stroke number and cadence by the WPM were consistent with the criterion measures by the SMART^{Wheel} (ICC= 0.99 for stroke number, ICC=0.97 for cadence) with less than 5% absolute percentage error for stoke number and 9% for cadence. The peak resultant force and wheel torque could be predicted to some extent by acceleration features on an individual subject basis. In addition, the estimated stroke number and cadence before and after the propulsion training session were statistically different, which was consistent with the changes in criterion stroke number and cadence.

Conclusion: This study demonstrated the preliminary validity of the WPM in estimating wheelchair propulsion characteristics in terms of stroke number and cadence. The WPM could potentially track upper limb movements for wheelchair propulsion in the natural environment of wheelchair users. Future studies should test more subjects and develop methods to merge data from the WPM's three devices to establish the validity of the WPM for real-world use.

Key Words: Acceleration; Biomechanics; Upper extremity; Wheelchairs; Spinal cord injuries

List of Abbreviation

Acc	Acceleration
SD	Standard Deviation
SCI	Spinal Cord Injury
SN	Stroke Number
WPM	Wheelchair Propulsion Monitor
WUSPI	Wheelchair Users Shoulder Pain Index

ADLs	Activity of Daily Living
MMP	Multimedia instructional program
ICC	Intraclass correlations
MWU	Manual Wheelchair Users

3.2 INTRODUCTION

Manual wheelchair users rely extensively on their upper limbs for mobility and activities of daily living. The long-term reliance on the upper limbs for performing daily activities has led to an increase in the prevalence of musculoskeletal injuries and pain. Between 49 % and 73% of manual wheelchair users have experienced carpal tunnel syndrome (Aljure, 1985; Burnham R.S., 1994; Sie, 1992), and between 30% and 73% of them have experienced rotator cuff tendinopathy or shoulder pain (Ballinger, 2000; Gellman, 1988; W. E. Pentland, Twomey, L. T., 1991). Any loss of upper limb function significantly affects mobility and independence of these individuals (K. A. Curtis, Roach, K. E., Applegate, E. B., Amar, T., Benbow, C. S., Genecco, T. D., Gualano, J., 1995; W. E. Pentland & Twomey, 1994; Silfverskiold, 1991).

There are ample reports indicated that pain and injury may be highly relevant in chronic SCI, and highly usage of upper limbs is blamed to be the cause. However, the frequency or intensity of upper limb activities that occur on a daily basis is unclear. Previous studies utilized vision systems such as Optotrak (M. L. Boninger, Baldwin, M., Cooper, R. A., Koontz, A., Chan, L., 2000) and VICON system (Gil-Agudo, 2010), and biomechanical analysis tools like a SMART^Wheel (Three Rivers Holdings Inc., Mesa, AZ) that can measure 6-dimensional propulsion forces and moments, to track upper limb motions for wheelchair propulsion. However, the cost and intricate setting of the vision systems and SMART^{Wheel} have limited their use within research laboratories and rehabilitation clinics.

Advances in miniature sensor technology have led to the development of wearable systems that can recognize and quantify user activity in the natural environment. Compared with ambulatory activity monitoring and recognition, there were fewer studies focusing on developing instrumentation and recognition software to monitor and classify upper limb usage and activities, especially for wheelchair users. Vega-Gonzalez et al. developed an upper-limb activity monitor relying on a pressure transducer that can depict not only movement or non-movement, but gave information about the position of the wrist with respect to the shoulder. However, the configuration would interfere with the subjects by restraining their movements and the sensor may become loose under large ranges of motion (Vega-González, 2005). Based on 10 non-impaired subjects and 10 chronic stroke patients for a period of eight hours, the results showed that the able-bodied participants used their dominant upper limb 10% more than their non-dominant upper limb and stroke patients used their unaffected upper limb twice as much as their affected upper limb (Vega-González, 2005). Nunn et al. used a commercial datalogger with connected sensors such as Electrocardiogram (ECG), piezoelectric respiratory band, pulse oximeter, and accelerometer to monitor patients with spinal cord injury during daily activity. Data were collected from subjects who were receiving treatment in a rehab hospital for a limited period of time. Simple analysis was performed to find time periods of significant activity and change (Nunn, 2005). Postma et al. found that wheelchair propulsion could be validly detected from a series of representative daily life activities by accelerometry-based activity monitors in patients with SCI (Postma, 2005). Tolerico et al., used a wheel rotation datalogger attached to the wheelchair wheel to quantify mobility characteristics and activity levels of manual wheelchair users in community settings (Tolerico, 2007). Coulter et al. placed a tri-axial accelerometer on the wheelchair wheel to track wheel revolutions, direction, and duration of movement (Coulter E.H., 2011).

Despite the aforementioned studies that used activity monitors to quantify upper limb activities, few studies have attempted to extract biomechanical variables from activity monitors that match the criterion measures collected by the laboratory-based devices such as vision systems or devices like a SMART^{Wheel}. Hiremath et al. estimated the temporal parameters of wheelchair propulsion including stroke time, propulsion time, and recovery time based on the three-dimensional acceleration at the third metacarpalphalangeal joint (3MP) derived from the vision system and compared the estimated values with those obtained from the SMART^{Wheel}. The results revealed high intraclass correlations of over 0.8 for all the temporal parameters (i.e. stroke, propulsion, and recovery time) over different surfaces (i.e. tile and carpet) (Hiremath, 2008). Ambur et al. used a wrist-worn accelerometry-based device called the eWatch to classify four wheelchair propulsion patterns of a single able-body subject based on extracted upper limb acceleration features. The average classification accuracy was in the range of 60-90% depending on surface type (Ambur, 2007). French et al. further expanded this work by including three normal subjects and the result of classifying four propulsion patterns was consistent with the previous study. A simpler binary classification scheme of arcing vs. non-arcing propulsion patterns was also explored, and the average classification accuracy reached 80%-90% depending on surface type (French, 2008).

Many laboratory-based biomechanical studies have identified key variables of wheelchair propulsion. In particular, the Consortium of Spinal Cord Medicine published a practice guideline, *Preservation of Upper Extremity Function Following Spinal Cord Injury: A Clinical Practice Guideline for Health Care Professionals*, which recommends reducing the frequency of repetitive upper limb tasks, minimizing forces required to compete tasks and minimizing extremes of wrist and shoulder motions (M. L. Boninger, Waters, R.L., Chase, T., Dijkers, M.P., Gellman, H., Gironda, R.J., Goldstein, B., Johnson-Taylor, S., Koontz, A., McDowell, S.L., 2005). Boninger et al. also found that lower peak forces, slower cadence, and a circular propulsive stroke in which the hand falls below the pushrim during recovery may help to prevent upper extremity injury among wheelchair users (M. L. Boninger, Koontz, A. M., Sisto, S. A., Dyson-Hudson, T. A., Chang, M., Price, R., Cooper, R. A., 2005). The purpose of this study was to conduct a preliminary performance analysis of a Wheelchair Propulsion Monitor (WPM) in estimating key biomechanical variables of wheelchair propulsion. The WPM integrated three devices including a wheel rotation datalogger, and an accelerometry-based device on the upper limb and underneath the wheelchair seat, respectively. The overall goal is to create an effective tool to monitor upper extremity usage and wheelchair propulsion characteristics in the natural environment of wheelchair users. The information on actual upper-limb usage will be helpful for clinicians and researchers to evaluate training outcomes and understand the etiology of upper limb injuries and pain in this population.

3.3 METHODS

3.3.1 Participants

A convenient sample of 5 subjects participated in this study thus far. Subjects were recruited through the IRB approved registries of the Human Engineering Research Laboratories (VA IRB# 0212005) and UPMC Department of Physical Medicine and Rehabilitation (Pitt IRB # 0304069). All subjects in the registries have provided informed consent to be contacted for future research studies. Subjects were included in the study if they 1) were 18 years of age or greater; 2) use of a manual wheelchair as a primary means of mobility; 3) have a Spinal Cord Injury. Subjects were excluded if they were unable to tolerate sitting for 2 hours, and/or have upper limb pain that limits mobility.

3.3.2 Experimental Protocol

Subjects were asked to pay at least two visits to the Human Engineering Research Laboratories (HERL). During the first visit, subjects completed a demographic survey and the Wheelchair Users Shoulder Pain Index (K. A. Curtis, Roach, K. E., Applegate, E. B., Amar, T., Benbow, C. S., Genecco, T. D., Gualano, J., 1995; K. A. R. Curtis, K. E. Applegate, E. B. Amar, T. Benbow, C. S. Genecco, T. D. Gualano, J., 1995). They were then asked to perform a series of consecutive activities according to a standard protocol in a semi-natural setting (e.g., hallway of the HERL, and HERL's activities of daily living lab). The protocol is shown in Table 8 and included wheelchair propulsion on a 30-meter level tile surface back and forward for three times, wheelchair propulsion up on a 12-meter 4° degree sloped tile surface for three times, and a series of activities of daily living that are representative of everyday life in manual wheelchair users. Subjects were asked to perform the activities in their own manner and at their own pace. The duration of each activity ranged from 5 to 10 minutes, and the total duration was about 40 minutes per subject. During the performance of activities, the subject's wheelchair was fitted with a SMART^{Wheel} that can measure 6-dimensional pushrim forces and moments, and a dummy wheel with the same dimensions as the SMART^{Wheel}. Simultaneous measurements with the SMART^{Wheel}, the WPM, and video recordings (as reference methods) were performed. After finishing the laboratory trial, subjects were instructed to leave HERL with the WPM attached to their wheelchair and the dominant upper arm, and go about their daily lives as usual for about two days.

The second visit was scheduled within two weeks from the first visit. During the second visit, subjects were asked to complete the wheelchair propulsion trials as described in the protocol of the first visit before and after a wheelchair propulsion training session. A multimedia instructional program (MMP) for wheelchair propulsion developed based on previous research studies and recommendations from a focus group (I. Rice, 2010; I. Rice, Gagnon, D., Gallagher, J., Boninger, M., 2010) was used to teach subjects appropriate propulsion techniques which emphasized reaching back, matching the speed of the hand to the speed of the pushrim, taking long strokes, and smoothly releasing the pushrim. Graphical overlays on the video together with audio input allowed for detailed explanations similar to an in-person presentation. Examples of good and bad techniques were also provided. The MMP was used in a previous study and significantly improved the propulsion techniques of manual wheelchair users (I. Rice, Gagnon, D., Gallagher, J., Boninger, M., 2010). After finishing the testing, subjects were instructed to leave HERL with the WPM for about two days. At the end of the study, subjects were asked to mail the WPM back to HERL or return the WPM to an investigator at a place of mutual agreement.

Table 8 Protocol overview

Visit I- Lab Trial
1. Consent form, demographics/wheelchair survey, pain questionnaires (WUSPI)
2. Real-life course propulsion
    1. Level propulsion (three times)
    2. Uphill propulsion (three times)
3. Mixed activity of daily living trial
    1. Push / Being pushed
    2. Open/ Close door
    3. Laundry
    4. Preparing meal
    5. Clothing
Visit I – Home Trial
1. 2-day field trial at home and community environment

Visit II – Lab Trial 1. Real-life course propulsion (three times) 2. Wheelchair propulsion training session 3. Real-life course propulsion (three times) **Visit II – Home Trial** 1. 2-day field trial at home and community environment

3.3.3 Wheelchair Propulsion Monitor (WPM)

The WPM integrated three devices including a wheel rotation datalogger attached to the wheelchair wheel, a wearable 3-axis accelerometer worn on the dominant upper arm, and the same 3-axis accelerometer attached underneath the wheelchair seat. The WPM monitors the wheelchair movement as well as the upper limb movement of the wheelchair users.

3.3.3.1 Wheelchair Rotation Datalogger (WRD)

The WRD developed at HERL tracks the number of wheel rotations, similar to a pedometer that tracks the number of steps. It is approximately 5cm in diameter and 3.8cm in depth (Figure 3). It is self-contained, lightweight, and powered by a 1/6D wafer-cell lithium battery, which enables the WRD to collect and store data up to three months. The WRD can be easily attached to the spokes of a manual wheelchair via two zip ties and thus requires no modifications to the wheelchair itself. The WRD measures the rotation of the wheelchair wheel through the use of three reed switches mounted 120° apart on the back of the printed circuit board and a magnet mounted at the bottom of a pendulum (Tolerico, 2007). As each reed switch is triggered, a date and time stamp of the event to the nearest tenth of a second is recorded. The time stamp data can be further processed to obtain distance traveled, speed, time of movement, and number of stops. The WRD has been used in previous studies to collect mobility characteristics of manual wheelchair users (Garrett, 2007; Tolerico, 2007).

Figure 3 Datalogger

3.3.3.2 SHIMMER

Shimmer (Shimmer Research, Dublin, Ireland) is a small wireless sensor platform that can record and transmit physiological and kinematic data in real-time. It is about 3cm in width, 5 cm in length, and 1.5cm in depth and about 60 gram. Shimmer used in this study contains a single tri-axial accelerometer and a power source which can continually collect data for up to four days and store the data on an onboard Security Digital Card. Two Shimmers were attached to the subject's dominant upper arm (Figure 4) and underneath the wheelchair seat with elastic straps (Figure 5), respectively. The upper limb Shimmer and the wheelchair seat Shimmer was configured at 20 and 60 Hz, respectively. Shimmer has been used in a number of studies to measure a person's posture, gait, and sit/stand transitions (Greene, 2010; McGrath, 2010; Patel, 2009; Twomey, 2010).

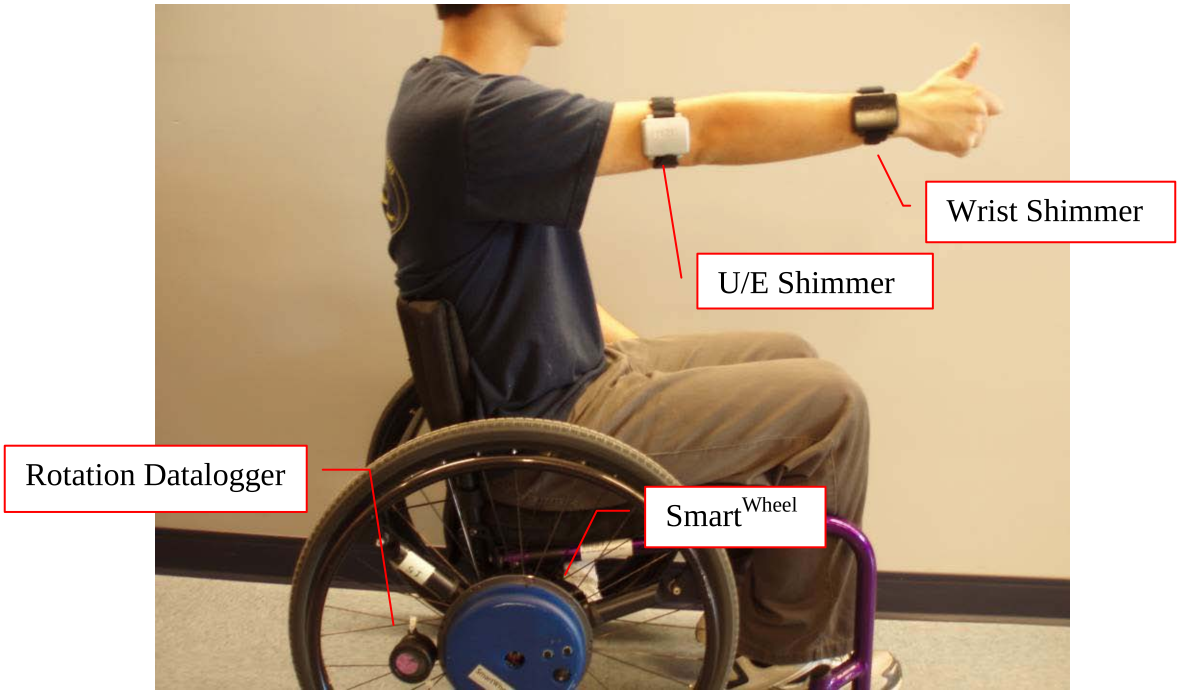

Figure 4 Instruments setup

Figure 5 Position for seat Shimmer

3.3.4 Data Reduction

3.3.4.1 SMART^{Wheel} data reduction

All biomechanical data were collected at 240Hz via a Security Digital card on the SMART^{Wheel} (Cooper R.A., 1998; R. A. Cooper, Robertson, R. N., VanSickle, D. P., Boninger, M. L., Shimada, S. D., 1997). The data were then filtered and converted to readable format with the SMART^{Wheel} software. The SMART^{Wheel} sign convention follows the right hand rule, with positive "x" forward, positive "y" up, and positive "z" point out of the wheel along the axle (R. A. Cooper, Robertson, R. N., VanSickle, D. P., Boninger, M. L., Shimada, S. D., 1997; A. M. Koontz, Cooper, R. A., Boninger, M. L., Yang, Y., Impink, B. G., Van der Woude, L. H., 2005). Positive moments were defined as counterclock-wise about the respective force vector. A stroke was defined as a propulsive contact. A cycle was defined as the period encompassing a propulsive contact and the subsequent recovery (R. E. Cowan, Nash, M. S., Collinger, J. L., Koontz, A. M., Boninger, M. L., 2009). Identification of contact and recovery phase was automatically recognized by a search algorithm and verified by visual inspection. The following biomechanical variables were calculated or directly obtained from the SMART^{Wheel} including stroke number, cadence, peak resultant force, and peak wheel torque. These variables were calculated for each cycle and then averaged for each propulsion trial to provide a general representation of propulsion. Push frequency was calculated as 1/cycle times. Resultant force (F_R) was defined as the vector sum of Fx, Fy, and Fz. Wheel moment (Mz) was defined as the moment along the axis of rotation responsible for angular acceleration of the wheel. A custom MATLAB program (Version 7.10 R2010a, The Mathworks Inc. MA, USA) was used to identify cycles, and compute biomechanical variables as described above.

3.3.4.2 WPM data reduction

In this preliminary study, we only looked into the wheelchair propulsion data collected from a Shimmer worn around the upper arm. The WPM sign convention was defined as AccX += superior, AccY += posterior, AccZ += lateral (right side). A 8^{th} order Butterworth low-pass filter with zero-lag, and 2-Hz cutoff frequency was applied to remove high frequency noise components affecting the data. To estimate the stroke number, we used the resultant acceleration (AccR) calculated as the vector sum of three directions of raw acceleration. A threshold was defined as the mean acceleration plus one standard deviation based on the first level propulsion trial during the first visit. The stroke number for each propulsion trial was then counted as the number of acceleration peaks over the established threshold. Figure 6 shows the filtered resultant acceleration signals of a wheelchair propulsion trial and the process of stroke number detection. Time for each propulsion trial was obtained by the lapse between the first and last strokes. Cadence was then calculated as the stroke number over time.

In order to predict peak resultant force and wheel torque, we also calculated a number of acceleration features including average peak acceleration and standard deviation in AccX, AccY, AccZ, and AccR for each trial, and average peak phase that considered the time lapse between the peak and valley resultant acceleration. A custom MATLAB program (Version 7.10 R2010a, The Mathworks Inc. MA, USA) was used to estimate stroke number and cadeence, and calculate the acceleration features as described above.

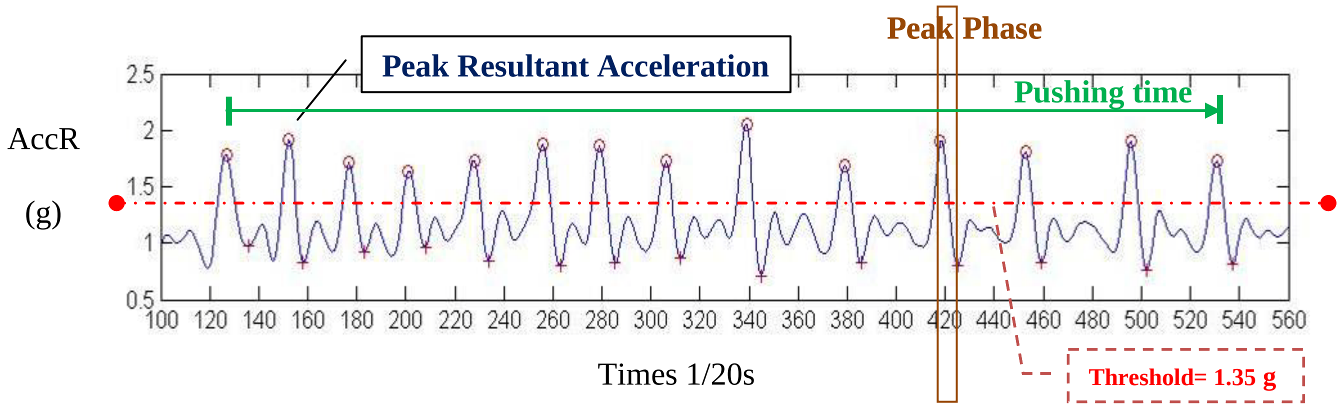

Figure 6 Resultant acceleration signals of a propulsion trial for Subject #05

3.3.5 Statistical Analysis

All statistical analyze were performed using SPSS software (ver. 18.0, SPSS Inc., Chicago, IL, USA) with the statistical significance at an alpha level of 0.05. Distributions of variables were examined and transformations were made where necessary. To determine the agreement between the estimated and criterion measures in terms of stroke number and cadence, the absolute difference and absolute percentage error were reported. Intraclass correlation coefficients (ICC (3, 1)) and the Bland and Altman plots were also used to assess the agreement. Each point on the Bland and Altman plot represents the mean (x-axis) and the difference (y-axis) of the criterion and estimated values for each propulsion trial of each subject. We used all the propulsion trials during the first and the second visit (before and after the training session) when accessing the agreement. To determine if the estimated stroke number and cadence can discriminate the training effect, Wilcoxon Signed Ranks Test were used to compare the criterion measures and estimate measures between the propulsion trials during the first visit and the after training trials during the second visit.

Due to the variability between subjects and the small sample size, separate stepwise multiple regressions were conducted for each subject to predict peak resultant force and wheel torque. The acceleration features described above were used as potential predictors. The two most significant predictors were chosen to construct the regression model for each subject and the adjusted R^2 was reported. A scatter plot was also used to display the trend and direction of relationship between the criterion variable and predictors for each subject.

3.4 RESULT

3.4.1 Participants

So far, a total of five MWUs with paraplegia participated in the study. There were three males and two females with a mean age of 38±16 years and weight of 161± 26 lb. The injury level of the subjects varied from L2 to T3. The number of years subjects have used a manual wheelchair was 14±11 years and all subjects used their wheelchair over six hours a day. Self-reported shoulder pain index was 9.1 ± 10.9 (where 0 indicates no pain and 150 indicate extreme pain). The top three common activities reported to cause shoulder pain were transferring from a wheelchair to the tub or shower, retrieving objects from an overhand shelf, and sleeping. All the five subjects completed all components of the study.

3.4.2 Estimation of Stroke Number and Cadence

The criterion and estimated measures on stroke number and cadence are provided in Table 9. Table 10 shows the absolute difference (Equation 3) and percentage error (Equation 4) between the criterion and estimated measures. The ICCs between the criterion and estimated measures are shown in Table 11. Figure 7 and Figure 8 are the Bland-Altman Plots which further illustrate the agreement between the criterion and estimated values.

Equation 3 Absolute Difference

$$\text{Absolute difference} = | V_{estimated} - V_{criterion} |$$

Note: V: Value

Equation 4 Percentage Error

$$\text{Percentage error} = \frac{| V_{estimated} - V_{criterion} |}{V_{estimated}} \times 100\%$$

Note: V: Value

Table 9 Criterion and estimated stroke number and cadence, mean (SD)

	Stroke Number				**Cadence**			
	Level		*Uphill*		*Level*		*Uphill*	
ID	Criterion	Estimated	Criterion	Estimated	Criterion	Estimated	Criterion	Estimated
1	28.1	28.1	24.4	23.3	.77	.81	1.04	1.03
	(2.0)	(2.2)	(1.8)	(2.1)	(.06)	(.09)	(.10)	(.13)
2	18.8	18.8	18.7	18.8	.77	.81	.79	.83
	(1.2)	(1.7)	(.9)	(.8)	(.06)	(.09)	(.04)	(.06)

3	18.5	18.5	18.3	18.6	.93	.96	1.22	1.28
	(1.5)	(1.5)	(1.6)	(1.3)	(.13)	(.10)	(.16)	(.18)
4	12.6	12.8	15.1	15.2	.60	.61	.95	.97
	(1.7)	(1.8)	(1.7)	(1.7)	(.05)	(.04)	(.09)	(.10)
5	17.1	17.1	14.9	14.9	.72	.78	.89	.96
	(2.8)	(2.7)	(.8)	(.8)	(.08)	(.08)	(.06)	(.04)

Note: SD: Standard Deviation

Table 10 Absolute difference and percentage error between the criterion and estimated values, mean (SD)

	Stroke Number				**Cadence**			
	Level		*Uphill*		*Level*		*Uphill*	
ID	AD	APE(%)	AD	APE(%)	AD	APE(%)	AD	APE(%)
1	.22	.79	1.13	4.61	.04	4.05	.03	2.94
	(.43)	(1.53)	(1.25)	(4.99)	(.02)	(2.68)	(.02)	(2.47)
2	.33	1.75	.33	1.79	.07	8.71	.04	5.14
	(.49)	(2.57)	(.71)	(3.87)	(.02)	(3.12)	(.02)	(2.60)
3	.22	1.24	.22	1.35	.07	6.85	.07	5.29
	(.43)	(2.38)	(.44)	(2.68)	(.04)	(4.26)	(.03)	(1.73)
4	.22	1.72	.11	.79	.03	4.74	.05	5.74
	(.43)	(3.38)	(.33)	(2.38)	(.02)	(2.88)	(.03)	(3.26)
5	.06	.40	0	0	.06	8.06	.07	7.79
	(.24)	(1.68)			(.03)	(3.92)	(.02)	(2.88)

Note: AD: Absolute Difference; APE: Absolute Percentage Error; SD: Standard Deviation

Table 11 Intraclass Correlation Coefficients on Stroke Number and Cadence

	ICC (3,1) SMART^{Wheel} with WPM		
Biomechanical Features	ICC	LB	UB
Stroke Number	0.99*	0.99	1.00
Cadence	0.97*	0.95	0.98

Note. Abbreviation: LB, lower bound; UB, upper bound.

*Correlations that was significant with p< 0.01

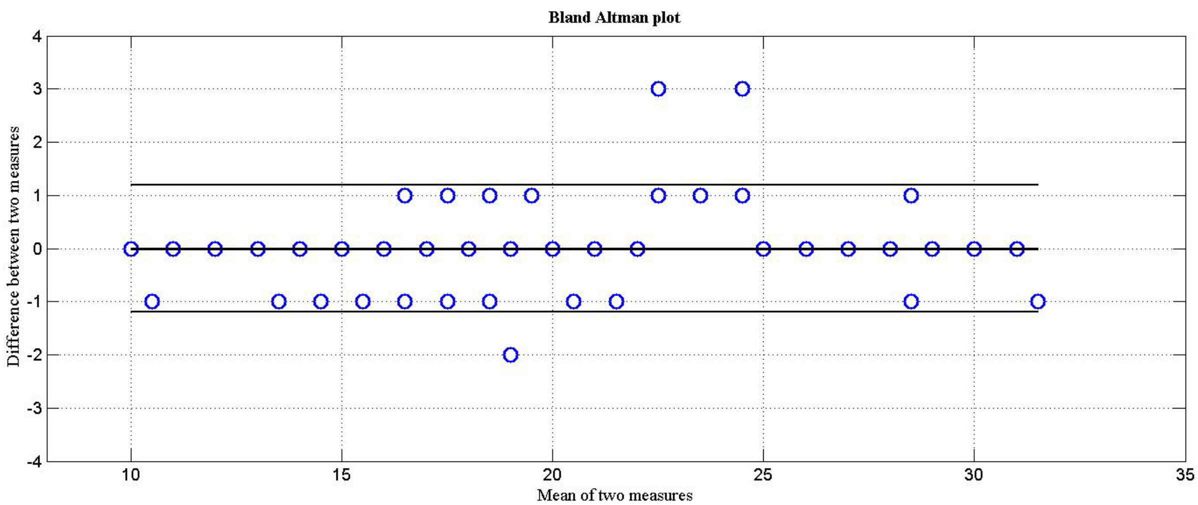

Figure 7 Bland Altman plot for Stroke Number

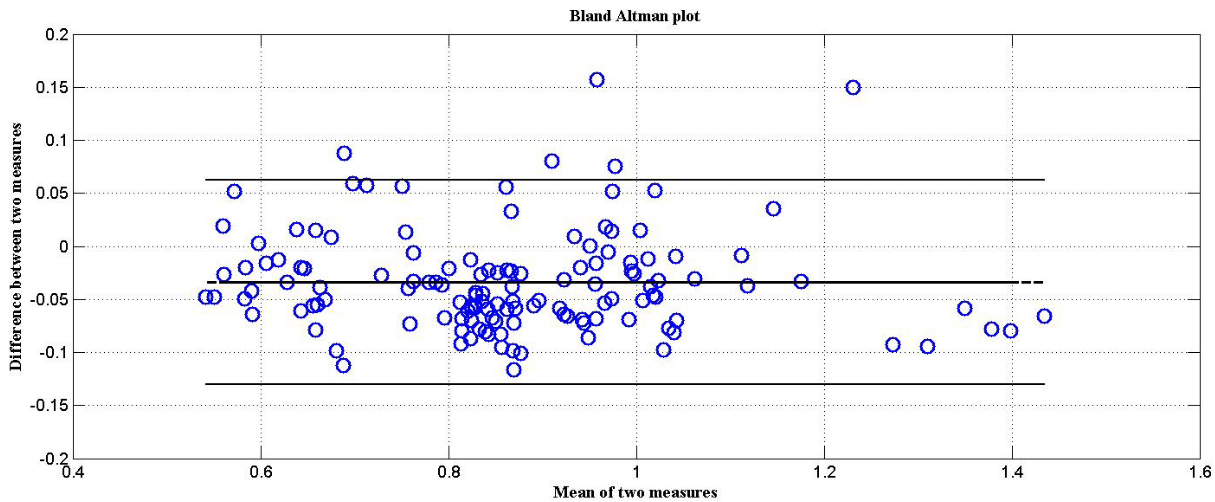

Figure 8 Bland Altman plot for Cadence

The immediate effects of propulsion training obtained from SMARTWheel and WPM were showed in Table 12. In terms of the ability of the estimated measures in discriminating the training effect, we found the criterion measures obtained from the SMARTWheel showed significantly decrease on cadence (Z= -2.03, p = 0.04) but not on stoke number after training on the level surface. Similarly, the estimated measures obtained from the WPM also showed a significant decrease on cadence (Z= -2.02, p = 0.04) but no statistical difference on stroke number. For the uphill condition, both the SMARTWheel and WPM had significant decrease on cadence (Z= -2.02, p = 0.04; Z= -2.03, p = 0.04; respectively), but no statistical difference on stroke number. Looking further into other criterion measures, there was a tread that participants increase push angle but reduce force and torque. However, the velocity did not change before and after training.

Table 12 Immediate effects of propulsion training (n=5)

	Criterion measures by SMART^{Wheel}	Estimated measured by WPM

		Pre-training	Post-training	*p*	Pre-training	Post-training	*p*
Stroke	Level	21.27± 5.76	18.63± 6.22	.50	21.33± 5.61	18.57± 6.25	.35
	Uphill	19.53± 4.15	17.90± 5.18	.35	18.93± 3.34	18.03± 5.07	.50
Cadenc	Level	.85± .15	.74± .15	.04	.87± .14	.76± .14	.04
	Uphill	1.08± .18	.89± .14	.04	1.09± .21	.93± .13	.04
Velocity	Level	1.28± .22	1.31± .18	.50			
	Uphill	1.08± .24	1.04± .20	.68			
PA	Level	102.86± 8.22	106.69± 14.16	.50			
	Uphill	102.45± 8.30	103.98± 9.15	.68			
Force	Level	51.07± 15.50	46.85± 20.78	.50			
	Uphill	75.29± 16.20	69.99± 22.38	.23			
Torque	Level	11.89 ±3.50	9.38± 4.28	.23			
	Uphill	17.53± 6.76	14.25± 8.50	.14			

Note: PA: Push Angle

3.4.3 Estimation of Peak Resultant Force and Wheel Torque

Table 13 to Table 14 shows the regression results for peak resultant force and wheel torque, respectively, for each participant. The results indicated that linear combination of some acceleration features was significantly related to the resultant force and wheel torque. However, each participant had specific regression model with different predictors. Figure 9 to Figure 10 showed the relation between criterion measures and specific acceleration features from different participants.

Table 13 Stepwise multiple regression models of Resultant Force

Predictors	Adjust R^2	Unstandardized coefficients		Standardized coefficients (β)	*P* value
		Estimate (SE)	95 % CI		
ID-01	R^2=84.9%	F (1,24) = 141.45, p<.01			
AccR		103.09 (8.67)	85.20- 120.98	.93	<.01
ID-02	R^2=73.6%	F (2,24) = 37.27, p<.01			
PeakPhase		6.77 (1.71)	3.25- 10.29	.53	<.01
AccR_SD		169.41 (53.71)	58.55- 280.26	.42	<.01
ID-03	R^2=61.1%	F (2,24) = 21.40, p<.01			
AccX		66.49 (15.13)	35.27- 97.71	.54	<.01
PeakPhase		-13.99 (3.24)	-20.68- -7.30	-.53	<.01
ID-04	R^2=67.4%	F (2,24) = 27.82, p<.01			
AccZ		153.28 (29.29)	92.83- 213.73	.66	<.01
AccX		33.67 (15.00)	2.70- 64.63	.29	.03
ID-05	R^2=87.8%	F (2,24) = 94.25, p<.01			
AccZ		101.27 (14.53)	71.28- 131.25	.70	<.01
AccR_SD		67.05 (22.05)	21.54- 112.56	.30	<.01

Note: SE: Std. Error

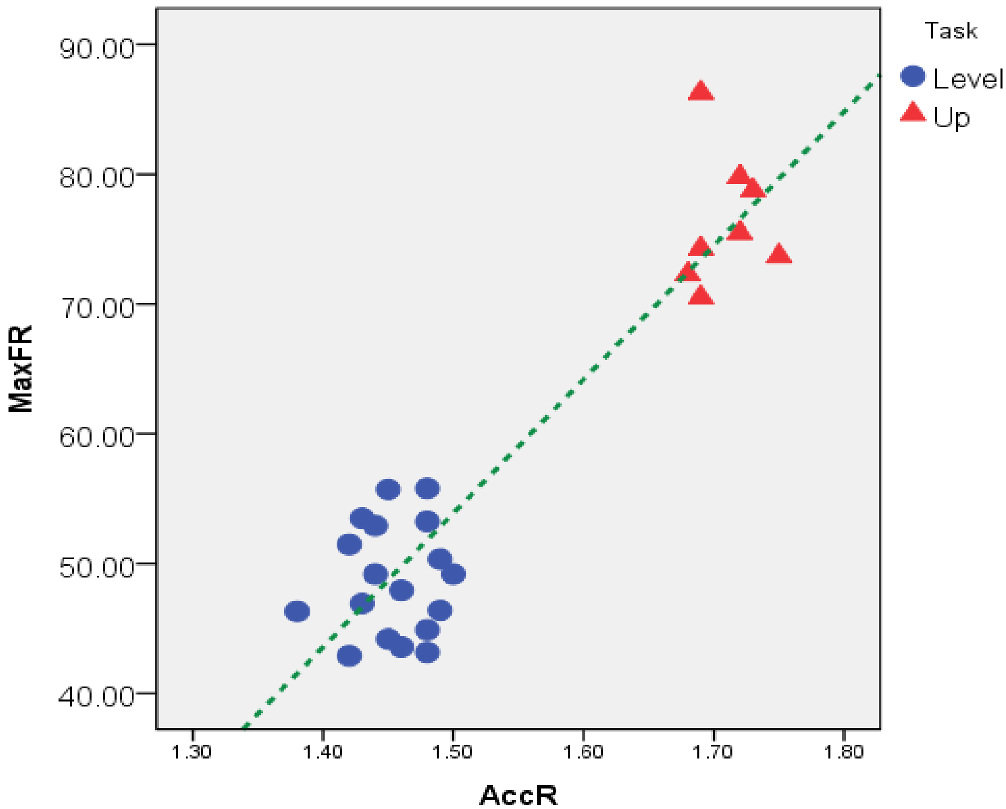

Figure 9 Correlation between peak resultant force (MaxFR) and AccR for Subject 01 for level and uphill propulsion

Table 14 Stepwise multiple regression models of peak wheel torque

Predictors	**Adjust R^2**	**Unstandardized coefficients**		**Standardized coefficients (β)**	***P* value**
		Estimate (SE)	**95 % CI**		
ID-01	R^2= 86.8%	F (2,23) = 82.27, p<.01			
AccR		36.98 (3.78)	29.16- 44.80	1.11	<.01
AccX_SD		-57.23 (26.65)	-112.36- -2.11	-.24	.04
ID-02	R^2= 66.0%	F (2,24) = 26.28, p<.01			
AccR_SD		17.11 (5.30)	6.18- 28.04	.49	<.01
PeakPhase		.46 (.17)	.12- .81	.42	.01

ID-03	R^2= 69.9%	F (2,24) = 31.13, p<.01			
AccX		22.36 (4.93)	12.17- 32.54	.53	<.01
AccZ		51.08 (11.88)	26.57- 75.59	.50	<.01
ID-04	R^2= 51.8%	F (2,24) = 14.95, p<.01			
AccY		-40.19 (8.27)	-57.25- -23.13	-.72	<.01
AccZ		27.17 (6.43)	13.90- 40.44	.63	<.01
ID-05	R^2= 91.3%	F (1,25) = 272.62, p<.01			
AccR		18.05 (1.09)	15.80- 20.31	.96	<.01

Note: SE: Std. Error

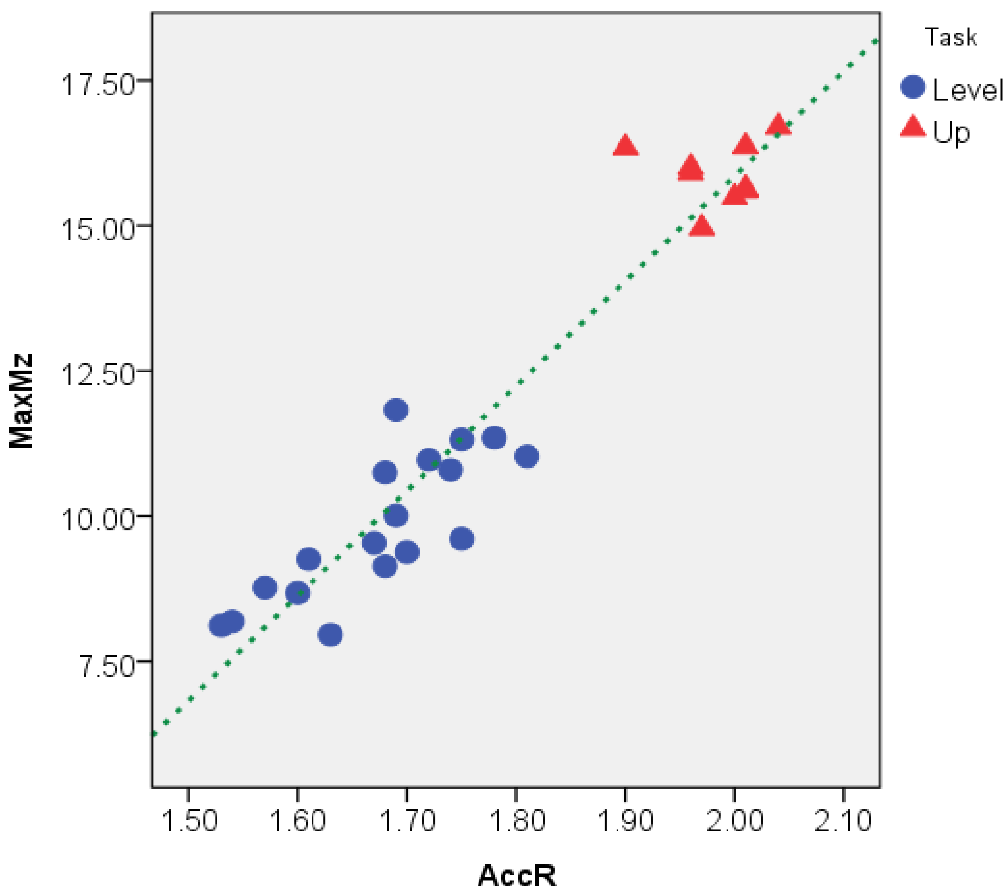

Figure 10 Correlation between peak wheel torque (MaxMz) and AccR for Subject 05 during level and uphill propulsion

3.5 DISCUSSION

This study evaluated the concurrent and discriminant validity of a wearable accelerometer for quantifying wheelchair propulsion characteristics. In terms of the concurrent validity, the estimated stroke number and cadence derived from the upper limb acceleration showed high intraclass correlations of 0.99 [0.99-1.00] and 0.97 [0.95-0.98], respectively, with the criterion measures by the SMART^{Wheel}. Previous studies had shown that an ICC value of 0.9 was deemed as excellent agreement if the lower bounds were greater than or equal to 0.75 (Lee, 1989). The absolute percent error between the estimated and criterion measures was less than 5% for stroke number and 9% for cadence. In terms of the discriminant validity, the changes in the estimated stroke number and cadence under similar velocity before and after the training session were consistent with the changes in the criterion measures, indicating that the estimated measures were able to detect the training effect when it existed. The ability of the wearable accelerometer in accurately detecting stroke number and cadence is essential for understanding the repetitiveness of upper limb movements that occur on a daily basis among manual wheelchair users. A number of ergonomic studies had strongly implicated frequency of task completion as a risk factor for repetitive strain injury or pain at the wrists (Loslever, 1993; Silverstein, 1987; Werner, 1998) and shoulder (Andersen, 2002; Cohen, 1998; Frost, 2002). The Clinical Practice Guideline on Preservation of Upper Extremity Function Following Spinal Cord Injury also recommends reducing the frequency of repetitive upper limb tasks (M. L. Boninger, Waters, R.L., Chase, T., Dijkers, M.P., Gellman, H., Gironda, R.J., Goldstein, B., Johnson-Taylor, S., Koontz, A., McDowell, S.L., 2005).

In addition to stroke number and cadence, previous research had also identified that high propulsion forces could be associated with risk of injury to the upper limbs (Andersen, 2002; M.

L. Boninger, Koontz, A. M., Sisto, S. A., Dyson-Hudson, T. A., Chang, M., Price, R., Cooper, R. A., 2005; Frost, 2002; Mercer, 2006; Roquelaure, 1997). When looking into the validity of the wearable accelerometer in predicting peak resultant force and wheel torque, we found that subjects required their own regression models with the adjusted R^2 ranging from 61.1% to 87.8% for peak resultant force and 51.8% to 91.3% for peak wheel torque, respectively. The acceleration features were able to predict the force and torque to some extent. However, the variability among subjects was high and there was no consistent regression model across subjects. It is possible that different propulsion patterns may cause varied impact on upper limb accelerations, as the acceleration signals are usually highly sensitive to orientation and position changes (Yang, 2009). By examining the video footage, we did notice that subjects were not consistent in their propulsion patterns. For example, subject #02, and #04 pushed on the wheel instead of pushrim frequently. Subject #03 was somewhat impatient and he pushed the wheelchair hastily. In Postma et al.'s study, they recruited 10 participants with SCI to see whether accelerometry-based activity monitor could validly detect wheelchair propulsion. The result indicated that the activity monitor comprised of six accelerometers had strong agreement (92%), sensitivity (87%), and specificity (92%) on detecting wheelchair propulsion. However, people with poor triceps strength had lower sensitivity than those with good triceps strength (Postma, 2005). Wheelchair users with good triceps strength had better cyclical movement patterns, which contributed to the better detection of wheelchair propulsion (Postma, 2005). With more subjects being recruited and tested, we will be able to have a better understanding on how the individual variability impacts the ability of the wearable accelerometer in predicting propulsion force and wheel torque.

While many papers have been published on developing and using instrumentation to monitor and classify activity, there is no specific instrumentation to our knowledge that is suitable for monitoring both the quantity and quality of upper limb movements of wheelchair users. Previous studies had attempted to use activity monitors to detect episodes of wheelchair propulsion (Postma, 2005) or the wheelchair movement in terms of traveling distance and speed (Coulter E.H., 2011; Tolerico, 2007). Although this study only analyzed the data from the wearable component of the WPM, the WPM which integrates three devices on the wheelchair and the upper limb of users has the potential to detect episodes of wheelchair propulsion, quantify wheelchair movements, as well as quantify the upper limb movement in terms of propulsion biomechanics. The WPM has the potential to provide clinical professionals and researchers with an indication of activity levels as well as propulsion skills of wheelchair users in their daily life. This information could also be used to evaluate and track the progress of interventions, and how wheelchair usage is related to the upper limb pain or injury.

This study is only a preliminary analysis of the validity of the WPM. The sample size is quite small (n=5). However, the preliminary analysis provided some insights into potential improvements of the protocol for further studies. Although the acceleration features seemed to be able to distinguish the propulsion efforts between the level and sloped surfaces, the sensitivity of these features was not clear. The limited variations of propulsion efforts might misguide the power of correlation. We plan to introduce more variations of the propulsion effort by changing target speed and surface condition. Also we plan to increase the number of propulsion trials for each subject so that we can split the data into a testing and validation set in order to evaluate individual regression models for each subject and better understand the predictive capability of the WPM on propulsion force and wheel torque.

3.6 LIMITATION

In this study, we replaced the subject' wheelchair wheels with a SMART^{Wheel} and a dummy wheel. Although these two wheels are designed in the same dimension, the weights are slightly different, which caused some turning tendency. The imbalance of pushing may affect the force direction and then change the way to push forward. Furthermore, we allowed participants to propel their wheelchair as their everyday use. However, we found some subjects were used to push on the wheel instead of the pushrim. The SMART^{Wheel} is a kinetic measurement tool that collects propulsion data based on contacting the pushrim. Therefore, this type of pushing restricts the measurement of force and moment, and thus diminished the correlation between the acceleration features and propulsion force or wheel torque. Future studies should consider using SMART^{Wheel} s on both sides and require participants to push on the pushrim. In this preliminary study, we only analyzed the acceleration signals from the upper arm Shimmer. However, different body segments may represent different levels of propulsion efforts. In Knorr et al.'s study, differences were shown on accelerometer features among body segments when qualifying upper limb movement in post-stroke patients (Knorr, 2005). Accelerometer data gathered from distal segments appear to provide more correlation with hand movement than features from data gathered from proximal segments (Knorr, 2005). In the future, we could combine the acceleration features derived from the dominant wrist with those from the dominant arm. In addition, wheelchair acceleration collected by a Shimmer underneath the seat may also provide a way to quantify certain biomechanical variables.

3.7 CONCLUSION

This preliminary study demonstrated the WPM has the ability to accurately detect stroke number and cadence, and detect a training effect when it exists. Specific acceleration features could be used to estimate individual resultant force and wheel torque. We anticipate that the WPM will become an accepted clinical tool for recording the amount and quality of functional upper-limb movements. This information could be useful for clinicians and researchers to evaluate training outcomes and understand the etiology of upper limb injuries and pain. Prevention of pain or injury in wheelchair users will have profound impact on manual wheelchair users, increasing their quality of life and decreasing healthcare costs associated with secondary injury.

3.8 ACKNOWLEDGE

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APPENDIX A

WHEELCHAIR USERS SHOULDER PAIN INDEX (WUSPI)

WHEELCHAIR USERS SHOULDER PAIN INDEX **Subject ID __________**

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 B

QUESTIONNAIRE PACKET

Questionnaire Packet

Development of measurement tools for propulsion training in the natural environment

COMPLETION LOG:	DATE:	INITIALS:	TIME:
Subject ID#: _______________ Data Collection	/ /		
Data Entry	/ /		
Verification	/ /		

4.0 Personal Data

Gender:
- ☐ Female (0)
- ☐ Male (1)

Age: ___________

Hand Dominance: R / L (Circle)

Weight: ___________ lb / Kg

Ethnic Origin:
- ☐ African-American (1)
- ☐ American Indian (2)
- ☐ Asian-American (3)
- ☐ Caucasian (4)
- ☐ Hispanic (5)
- ☐ Other (6): ________________

5.0 Onset of injury (Date mm/dd/yyyy): __________________

6.0 Spinal Cord Injury Level: __________________

Type of Spinal Cord Injury (Complete/Incomplete) :__________________

7.0 Type of wheelchairs:
- ☐ Standard Manual wheelchair (1)
- ☐ Light Weights (2)
- ☐ Ultra-light (3)
- ☐ Sport (4)
- ☐ Other (5): _____________

What is duration of using your wheelchair on average per day?
- ☐ Less than 2 hours
- ☐ 2-4 hours
- ☐ 4-6 hours
- ☐ 6-8 hours
- ☐ Over 8 hours

How long have you been using a wheelchair? _______________ years

How long have you been using your current wheelchair? __________________ years

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