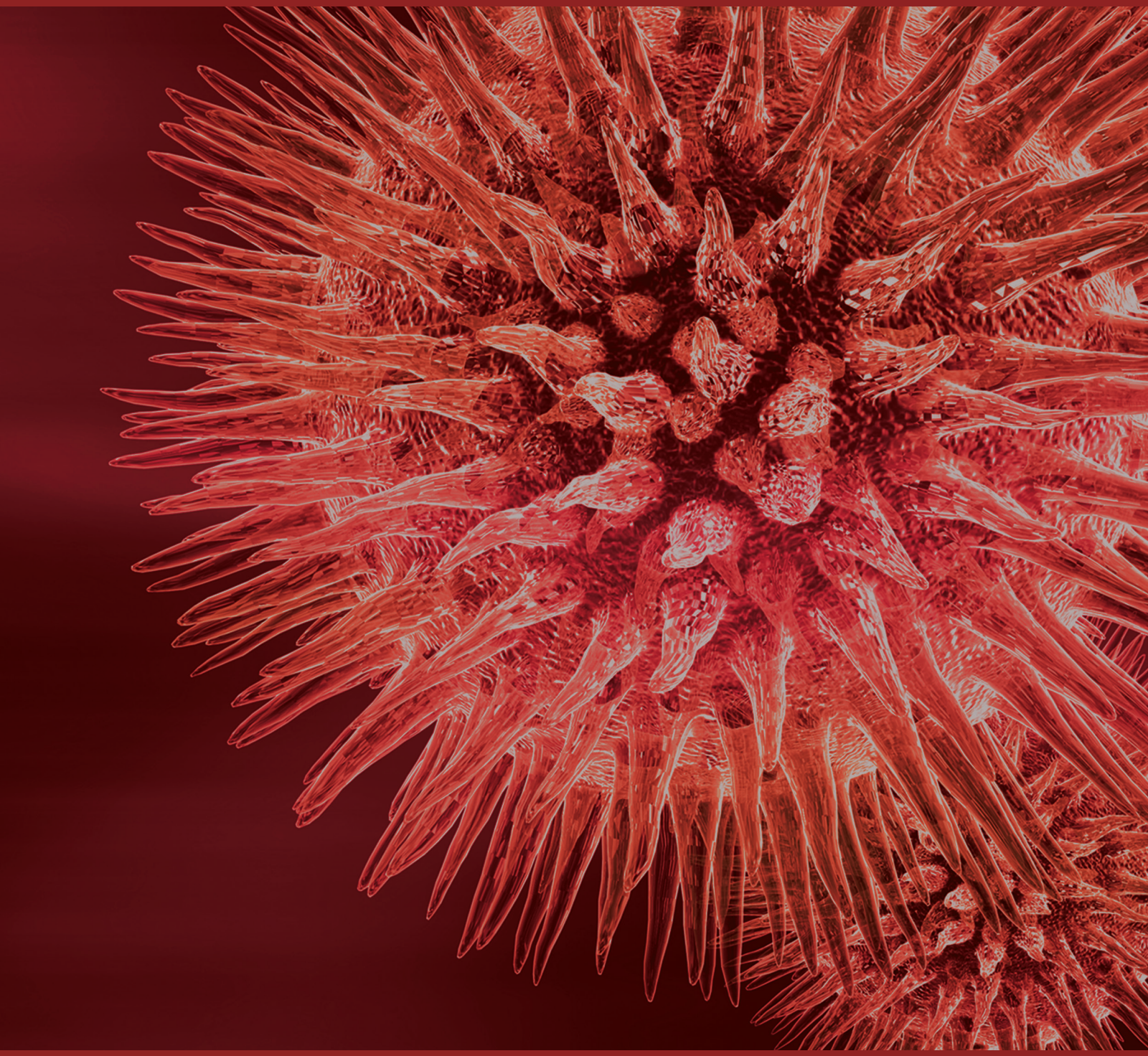
BioMed Research International

Wheeled Mobility

Guest Editors: Alicia M. Koontz, Dan Ding, Yih-Kuen Jan, Sonja de Groot, and Andrew Hansen



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Editorial

Wheeled Mobility

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Independence in mobility is one of the most important determinants of quality of life for individuals with disabilities [1]. While independent mobility is achievable with a wide variety of mobility-related technologies in existence today (e.g., prosthetic devices, powered orthotics, exoskeletons, etc.), wheeled mobility devices continue to make up the greatest portion of assistive devices in use. There are currently about 2.7 million wheelchair users in the United States (US) [2]. The powered and manual mobility market globally (with US being the largest regional market) is projected to grow exponentially due to aging baby boomers and increasing longevity [3]. This anticipated increased growth in the numbers of wheelchair users and that many wheelchair users continue to develop secondary disabilities along with experiencing other complications and barriers associated with wheelchair use brought us to this special issue.

This special issue features seven original research studies and seven development studies that advance our knowledge about secondary disabilities associated wheelchair use, measurement and training tools, and new assistive devices and adaptations. One ongoing thread of research in our field is to better understand how certain wheelchair features and tasks can lead to disabling secondary conditions such as pressure ulcers and upper limb pain. Y.-S. Yang et al. give us insight into injury risks associated with wheelchair standing by quantifying the amount of shear displacement that occurs between the body and backrest and seat, the ranges of motion of the lower limb joints, and the forces that act on the knee and foot during the seat-to-stand transition. Y.-S. Lin et al. show us that performing wheelchair push-ups for pressure relief reduces the shoulder joint's subacromial space and may predispose wheelchair users to shoulder impingement. This study also found that greater narrowing of the space after performing a repetitive rotator cuff task to simulate overuse was associated with increased years of wheelchair use and higher levels of shoulder pain. B. Slavens et al. describe in detail the propulsion techniques used by children and young adult manual wheelchair users with spinal cord injury. They found that the weight-normalized forces and moments these children and youth used to propel were similar to those reported in adult samples and, thus, identified children and youth as targets for interventions that help to prevent upper limb pain and overuse injuries. M. M. B. Morrow et al. in their comprehensive and detailed magnetic resonance imaging analysis of wheelchair users with shoulder pain tell us where the most common and exact locations of shoulder tendon tears are and describe other widespread abnormalities lending to a deeper understanding into patterns of

pathological shoulder findings among manual wheelchair users.

Other original research studies in this issue extend our understanding of the demands of manual wheelchair use and address either the personal or wheelchair attributes that affect the amount of effort required. D. Gagnon et al. show that pushing up a ramp increases shoulder mechanical and muscular effort and that success with steeper slopes is likely influenced by one's strength generating capacity and body mass index. A. Mandy et al. demonstrate that grip forces are reduced with a one-arm drive system, a type of adaptation often prescribed to individuals with hemiplegia, that "ties" the rear wheels together and can be driven and steered with one arm and foot compared to a similar type of system that does not "tie" the rear wheels together. F. Chenier and R. Aissaoui give us promising data on the use of carbon fibre in wheelchair frame construction in reducing the mechanical work load required to travel a certain distance while at the same time reducing the whole-body vibrations transmitted to the body from the ground.

The development of new robotic assistive devices and training programs and tools to enhance mobility and outcome measurement are additional areas of recent interest and growth. G. G. Grindle et al. describe a new wheelchair mounted robot arm called "StrongArm" capable of moving a person from their wheelchair to another surface and back eliminating the heavy lifting that would normally be required from a caregiver. E. M. Giesbrecht et al. detail the development and components of a new wheelchair skills home-based program tailored specifically for the needs of older adults. D. C. Kamaraj et al. establish the content validity of two new clinical tools, one specifically designed for screening for the appropriateness of power mobility and another for driving assessment. C.-Y. Tsai et al. find that wheelchair users who independently transfer and score higher (e.g., better) on the transfer assessment instrument, a new clinical scale used to assess a patient's transfer technique, exhibit the type of biomechanics that may protect them from developing upper limb pain and injuries.

The ongoing development, evaluation, or refinement of more accurate monitoring devices and measurement tools is important for quantifying the user performance and the interface between the user and the wheelchair. M. Ojeda and D. Ding discover that wearing a triaxial accelerometer around the upper arm combined with a wheel-mounted data logging system is a viable option to collect temporal data such as stroke frequencies and the cumulative exposure to forces on the upper limbs that occurs from daily wheelchair use. B. Mason et al. find that while using a highly portable miniature wheelchair data logger system to collect elite wheelchair activity works well for certain metrics, it falls short of being able to fully describe the intricacies of wheelchair performance on the court when compared to a less portable radiofrequency-based indoor tracking system. C.-W. Lung et al. call to our attention that the peak pressure index metric used to quantify seat interface pressure is sensitive to changes in wheelchair tilt and recline. Their study quantifies the amount of displacement in the metric and offers suggestions about how to remove this effect and allow for more accurate pressure readings.

We hope the science in this special issue will stimulate continued research that shares the same overarching goal of all this work which aims to help wheelchair users to become completely independent and highly functional in their devices, be protected from secondary complications associated with wheelchair use, be fully integrated into society, and be able to live active and healthy lifestyles.

Acknowledgment

We are sincerely grateful for all the unique and thoughtful contributions that were made by all the authors and our fellow colleagues in the field to this body of research on wheeled mobility.

Alicia M. Koontz
Dan Ding
Yih-Kuen Jan
Sonja de Groot
Andrew Hansen

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Research Article

Evaluation of Pediatric Manual Wheelchair Mobility Using Advanced Biomechanical Methods

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There is minimal research of upper extremity joint dynamics during pediatric wheelchair mobility despite the large number of children using manual wheelchairs. Special concern arises with the pediatric population, particularly in regard to the longer duration of wheelchair use, joint integrity, participation and community integration, and transitional care into adulthood. This study seeks to provide evaluation methods for characterizing the biomechanics of wheelchair use by children with spinal cord injury (SCI). Twelve subjects with SCI underwent motion analysis while they propelled their wheelchair at a self-selected speed and propulsion pattern. Upper extremity joint kinematics, forces, and moments were computed using inverse dynamics methods with our custom model. The glenohumeral joint displayed the largest average range of motion (ROM) at 47.1° in the sagittal plane and the largest average superiorly and anteriorly directed joint forces of 6.1% BW and 6.5% BW, respectively. The largest joint moments were 1.4% body weight times height (BW × H) of elbow flexion and 1.2% BW × H of glenohumeral joint extension. Pediatric manual wheelchair users demonstrating these high joint demands may be at risk for pain and upper limb injuries. These evaluation methods may be a useful tool for clinicians and therapists for pediatric wheelchair prescription and training.

1. Introduction

Among children under the age of 18, the wheelchair is the most widely used assistive mobility device impacting over 88,000 children, 90% of which use manual wheelchairs [1]. According to the 2012 Americans with Disabilities Report, approximately 3.7 million people use a wheelchair, with around 124,000 wheelchair users under the age of 21 and 67,000 under the age of 15 [2]. Despite this large number of pediatric wheelchair users, there is very limited information quantifying pediatric wheelchair mobility. This research will be valuable to the field of pediatric rehabilitation for improving clinical care to a developing and growing population of children and adolescents with spinal cord injury (SCI) and other orthopaedic disabilities. While a larger body of knowledge exists surrounding adult wheelchair biomechanics, it should not be assumed that the developing child experiences the same loading demands or is exposed to similar risk factors for overuse injuries, particularly during maturation. It is clear that pediatric onset SCI affects all aspects of quality of life, including mobility, participation, and function [3]; however, the impact of pediatric onset SCI on wheelchair biomechanics and the development of UE pain and pathology is unclear. Furthermore, it is unknown how each of the upper extremity joints (e.g., shoulder complex, elbow, and wrist) is affected during wheelchair mobility.

SCI is one of the leading causes of wheelchair usage in children and adolescents. The estimated annual incidence

of SCI in the United States is approximately 40 cases per million population or approximately 12,000 new cases each year [4]. There are approximately 273,000 persons with SCI in the United States [4], and this number is rising, the majority of which heavily relies upon manual wheelchairs for mobility and community participation. Manual wheelchair use requires the upper extremities for mobility, performing transfers and weight relief, and performing activities of daily living (ADLs). The upper extremity is not intended for this load magnitude or frequency. Adult manual wheelchair users with SCI have a reported prevalence of 1/3 to 1/2 with upper extremity pain and/or deterioration of function and injury, including destructive shoulder arthropathy, degenerative arthritis of the shoulder and elbow, rotator cuff tendonitis, coracoacromial pathology, and carpal tunnel syndrome [5–11]. More specifically, estimates of shoulder pain among manual wheelchair users with paraplegia range from 30% to 73% [6, 7, 10, 12, 13]. It has been shown that shoulder pain and degenerative changes, especially at the acromioclavicular joint, may develop prematurely in individuals with SCI due to overuse and altered mechanical stresses, particularly in those with high levels of manual wheelchair activity [5, 6]. It has also been determined that manual wheelchair users who propel at a faster cadence and experience greater peak resultant handrim forces relative to body weight also have greater incidence of impaired median nerve function [13, 14]. Due to longer-term wheelchair use from pediatric onset SCI, these injuries may occur earlier and reduce or severely limit independence, function, and quality of life in children. Further insight into wheelchair mobility biomechanics of pediatric wheelchair users is critical for ultimately preventing these complications and improving their quality of life.

Upper limb pain and pathologies have been associated with increased loading at extremes of joint excursions [15]. Manual wheelchair use in adults has shown high shoulder joint loading with forces ranging from 7% to 12% of body weight, of which our prior study of a child with SCI agrees [6, 16, 17]. High joint forces during manual wheelchair use have been shown to directly correlate to the cause of joint pain and injury [6]. Additionally, previous research has also investigated wheelchair stroke patterns in adults, the motion the hand makes during the stroke cycle, which only differentiates during the recovery phase, when the hand is not in contact with the handrim and is restricted to its movement. Four different propulsion patterns have been identified in adult users, by the hand's motion during the recovery phase: single-looping over propulsion (when the hands rise above the handrim), double-looping over propulsion (when the hands rise above and then fall below the handrim), semicircular (when the hands fall below the handrim), and arcing (when the hand follows the path of the pushrim). This research also showed that in adults, the semicircular pattern resulted in the lowest cadence to go the same speed and the greatest percentage of time spent in the contact phase, allowing the user to impart force to the handrim over a greater angle and longer time. Since both of these parameters have been linked to reduction of injury in adults, the semicircular pattern is the recommended technique for adult manual wheelchair propulsion [12, 13, 18]. A study by Mercer et al. supports further investigation of wheelchair design, prescription, training, and propulsion biomechanics that reduce shoulder joint forces and moments to preserve upper limb integrity [6]. Quantification of 3D joint dynamics in children is essential for uncovering the root of secondary injuries. In order to evaluate the pediatric population of manual wheelchair users, it is essential that biomechanical methods are applied that consider the unique musculoskeletal features of a child.

This study seeks to address the knowledge gap in biomechanical evaluation of pediatric wheelchair mobility by quantifying and characterizing upper extremity joint dynamics during manual wheelchair mobility. Specifically, we will quantify shoulder complex, elbow, and wrist joint demands during mobility to determine the ranges of motion, forces, and moments, which may play a key role in the clinical evaluation of joint integrity and wheelchair skills and performance. We hypothesize that proximal joint ranges of motion, joint forces, and joint moments will be significantly different than distal joint dynamics. This work presents biomechanical methods for pediatric evaluation that aim to minimize the risk of developing secondary complications associated with wheelchair use, such as upper extremity pain and overuse related upper limb injuries.

This work is the first of its kind to quantify wheelchair biomechanics in children. Better knowledge of how to evaluate UE dynamics during wheelchair propulsion may enhance our understanding of the onset and propagation of UE pain and secondary pathologies. This may lead to improvements in wheelchair prescription, design, training, and transitional care. Ultimately, the incidence of manual wheelchair use related pain and pathology might be reduced while maximizing function and quality of life of children with SCI.

2. Materials and Methods

2.1. Subjects. Twelve pediatric manual wheelchair users with spinal cord injury were evaluated at Shriners Hospitals for Children—Chicago. Subject characteristics are described in Table 1. IRB approval was obtained and an assent form or informed consent form was signed by the child and/or their parent/guardian. Subject inclusion criteria: under 21 years of age, chronic incomplete SCI diagnosis at least one year after injury, and manual wheelchair as primary mode of mobility.

2.2. Data Collection. Subject specific measurements were obtained and twenty-seven passive reflective markers were placed on bony anatomical landmarks and technical locations of the subject, including the suprasternal notch, xiphoid process, spinal process C7, acromioclavicular joint, inferior angle, trigonum spinae, scapular spine, acromial angle, coracoid process, humerus technical marker, olecranon, radial and ulnar styloids, and the third and fifth metacarpals [17]. Based on validated methods by Šenk and Chèze, due to the subcutaneous motion of the scapula, the trigonum spinae and inferior angle markers were only used during a static trial with the subject in anatomical position. During

Table 1: Subject characteristics for each subject and the calculated group averages and standard deviations.

Subject	Age (years)	Height (cm)	Weight (kg)	Gender	Arm dominance
1	9.8	137.1	31.9	Male	Right
2	7	121.9	26.5	Male	Left
3	8.7	128.0	28.2	Male	Right
4	20	167.6	51.1	Female	Left
5	18	129.5	61.2	Male	Right
6	18	177.8	54.0	Male	Left
7	17.7	67.8	49.0	Male	Left
8	16	170.2	63.1	Male	Right
9	11.3	154.2	34.7	Male	Right
10	17	151.3	41.3	Male	Left
11	8.6	124.4	31.7	Female	Right
12	6.5	119.3	28.5	Male	Right
Avg.	13.2	137.4	41.8		
St. dev.	5.0	29.9	13.4		

dynamic trials, the positions of these markers were calculated using a method based on rigid body theory, shown to be appropriate for scapular motion tracking especially during tasks with less than 120 to 150 degrees of arm elevation. This method had low RMS errors (5.4–10.3 deg), similar to those of commonly used tracker (3.2–10.0 deg) and acromion (4.8–11.4 deg) methods [19]. A SmartWheel (Outfront, Mesa, AZ), with an air tire, replaced the wheel on the dominant side of the subject's wheelchair for kinetic data collection; the SmartWheel companion wheel replaced the subject's wheel on the nondominant side. Plastic coated handrim attachments were not used on either wheel for any of the subjects as none of the subjects used coated handrims or gloves to assist with their propulsion.

The subject propelled his or her manual wheelchair along a 15 m walkway at a self-selected speed and self-selected propulsion pattern (Figure 1). A 10-camera Vicon MX system captured the 3D marker trajectories at 120 Hz, while simultaneously the SmartWheel collected the 3D forces and moments occurring at the hand-handrim interface at 240 Hz. Multiple trials were collected, with adequate rest provided to the subject as needed.

Figure 1: Subject preparing to begin motion analysis with the SmartWheel.

2.3. Upper Extremity Biomechanical Model. A custom bilateral pediatric upper extremity model was applied to the data to determine 3D joint angles, forces, and moments [17]. This biomechanical model comprises 11 segments, including thorax, clavicles, scapulae, upper arms, forearms, and hands. The joints of interest are three-degree-of-freedom thorax, wrist, glenohumeral, and acromioclavicular joints and two-degree-of-freedom sternoclavicular and elbow joints.

Segment coordinate systems (SCS) were determined for each of the model's 11 segments. Following ISB recommendations, the SCS axes are aligned such that the Z-axis points laterally towards the subject's right side, the X-axis points anteriorly, and the Y-axis points superiorly [20]. The joint angles were determined by the relative motion between two adjacent SCS, distal relative to proximal. A Z-X-Y Euler sequence is used to calculate the glenohumeral, elbow, wrist, and thorax joint angles, and a Y-X-Z Euler sequence is used for the acromioclavicular and sternoclavicular joint angle computation. The SCS follow the right-hand rule with the Z-axis as the flexion-extension axis, the X-axis as the abduction/adduction axis, and the Y-axis as the internal-external rotation axis.

2.4. Data Processing. Vicon Nexus was used to process the marker trajectories. Any gaps in the data were filled using the cubic spline interpolation feature and the resulting marker trajectories were filtered using a Woltring filter with a mean squared error setting of 20. The kinetic data from the SmartWheel was filtered using a low pass FIR filter. The kinetic data was then resampled to 120 Hz in MATLAB.

For each subject, ten wheelchair stroke cycles, compiled from multiple trials, were analyzed to produce average subject parameters, which were then used to compute the average group parameters of interest. Only strokes occurring during

steady-state propulsion were included in the analyses; the start-up and stopping pushes were excluded. Average time series data of the joint angles, forces, and moments were all time-normalized to percent of the wheelchair stroke cycle. The stroke cycles were separated into two phases (contact and recovery) based on total force applied to the handrim, with the contact phase subdivided into periods of propulsive contact (propulsion) and nonpropulsive contact (initial contact and release) as determined by the moment about the wheel axle [21]. Contact phase angle, the angle the wheel moved during hand contact with the handrim, was determined by subtracting the wheel angle at the end of the contact phase from the wheel angle at the start of contact phase. The propulsion period angle, the angle the wheel moved during propulsive hand contact, was similarly defined during the propulsion period. These definitions for wheelchair stroke cycle phases and periods, and the angles, follow those described by Kwarciak et al. [21] (Figure 2). The stroke pattern was determined using the sagittal plane motion of the marker on the third metacarpal, plotting the vertical position versus fore-aft position [12].

Peak joint angles and angular ranges of motion were identified and calculated. All force data were normalized to percent body weight (%BW) and all moment data were normalized to percent body weight times height (%BW × H). Peak joint forces and moments were also identified. Two sample t-tests were used for statistical comparisons amongst group average joint ranges of motion and average peak dynamics.

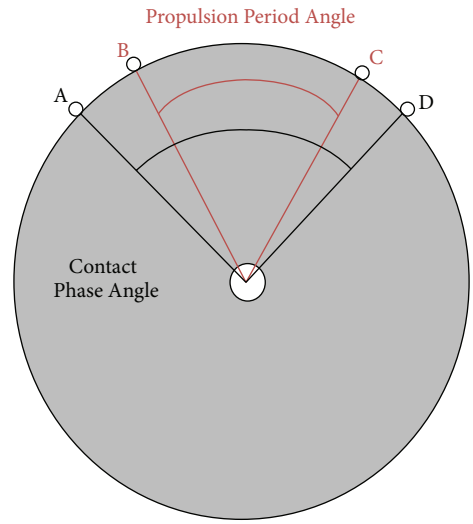

Figure 2: Contact phase angle and propulsion period angle. The contact phase angle begins at Point A, when a force is detected on the handrim, indicating hand contact, and ends at Point D, when a handrim force is no longer present. The propulsion period angle begins at Point B, when a propulsive moment about the wheel axle is present, and ends at Point C, when the propulsive moment is no longer detected.

3. Results

3.1. Temporal-Spatial Parameters. The average propulsion speed was 1.23 m/s ± 0.26 m/s, ranging from 0.79 m/s to 1.6 m/s, with an average cadence of 1.1 strokes/sec ±0.2 strokes/sec. The average contact phase occurred from 0% to 35.8% stroke cycle with a range of 25% to 45% stroke cycle. Within the contact phase, the initial contact period occurred on average from 0% to 3.6% stroke cycle, the propulsion period on average occurred from 3.6% to 34.1% stroke cycle, and the release period occurred on average from 34.1% to 35.8% stroke cycle. The average contact phase angle was 85.6° ± 15.7°. The average propulsion period angle was 72.6° ± 11.9°. Lastly, the average peak resultant handrim force was 10.1% BW ± 3.7% BW.

Additionally, there was a wide array of propulsion patterns utilized by the children. One subject used the single-looping overpropulsion (SLOP) pattern, 3 subjects used the double-looping overpropulsion (DLOP) pattern, and 3 subjects used the recommended semicircular (SC) pattern [12]. The remaining five subjects used a mixture of patterns making the primary pattern unidentifiable.

3.2. Joint Kinematics. Group mean joint angles of the thorax, sternoclavicular, acromioclavicular, glenohumeral, elbow, and wrist joints were characterized over the wheelchair stroke cycle. These mean joint angles in each plane of motion, along with +/− one standard deviation, are depicted in Figures 3 and 4. The mean peak angles and ranges of motion (ROMs) of each joint were also identified and computed over the stroke cycle. The joint ROMs are depicted in Figure 5. Additionally, multiple parameters were statistically significantly different ($P < 0.05$) from one another, as seen in Table 2.

3.3. Joint Kinetics. Group mean joint forces and moments of the glenohumeral, elbow, and wrist joints were characterized over the wheelchair stroke cycle. The group's mean joint forces (+/− one standard deviation) along each axis are depicted in Figure 6. The mean group joint moments (+/− one standard deviation) in each plane of motion are depicted in Figure 7.

Additionally, average peak joint forces and moments were identified and shown in Figures 8 and 9, respectively. Statistically significant differences ($P < 0.05$) were found between the multiple parameters, as seen in Table 3. The glenohumeral joint forces were statistically significantly higher than the wrist joint forces directed superiorly, laterally, and posteriorly. The wrist joint forces in the anterior and inferior directions were significantly greater than those at the GH joint. Additionally, the GH joint experienced significantly higher joint forces directed superiorly and posteriorly than the elbow joint, while the elbow joint experienced significantly higher inferiorly directed force than the GH joint. Lastly, the elbow joint experienced significantly higher forces than the wrist in the superior, lateral, and posterior directions. When evaluating joint moments, the GH joint experienced significantly greater moments in flexion, abduction, external rotation, and extension than the wrist joint, while the elbow was only significantly greater than the wrist in the extension moment. Additionally, the GH joint experienced significantly higher moments than the elbow joint in internal rotation,

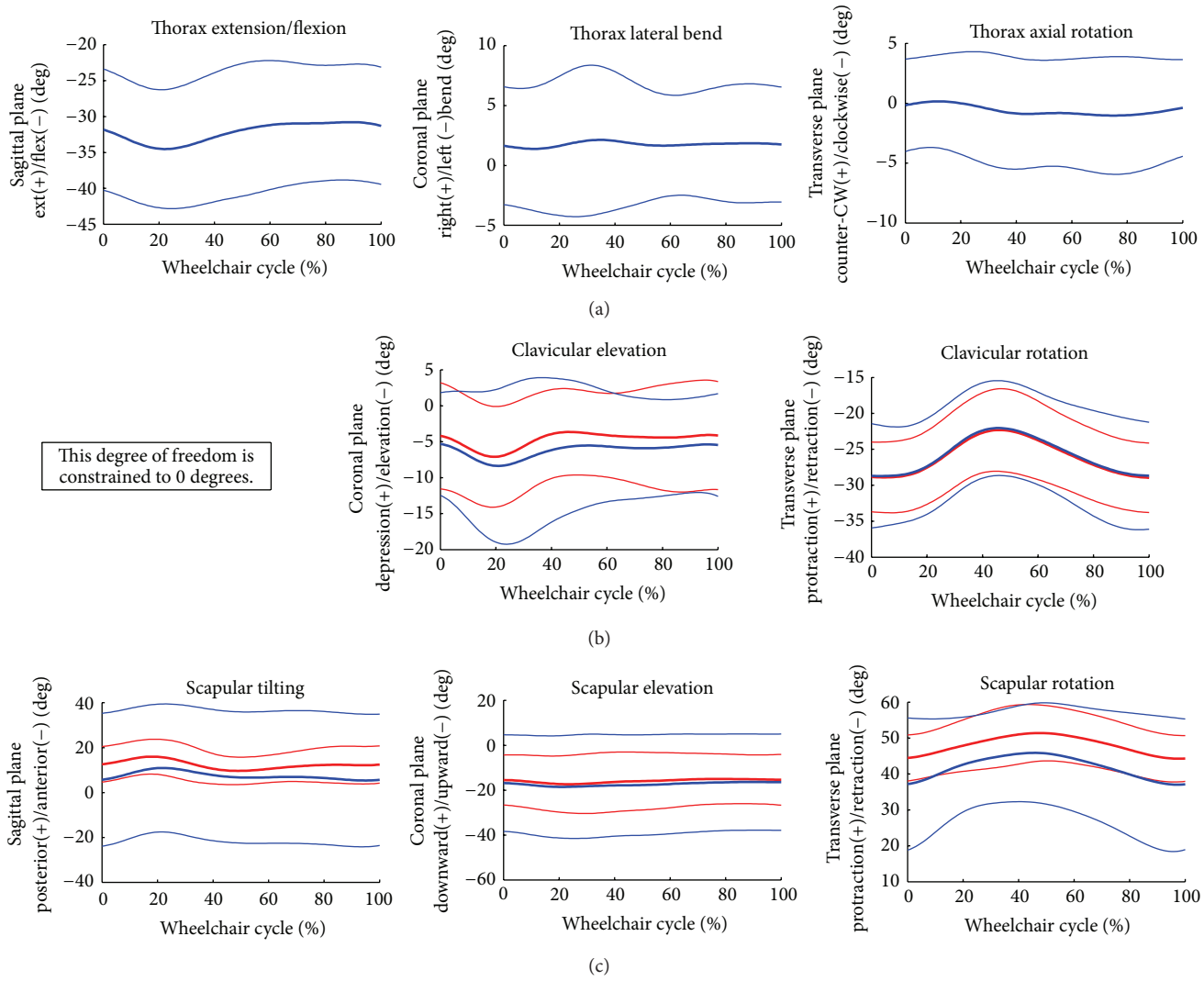

Figure 3: Group average joint kinematic data of the thorax, sternoclavicular, and acromioclavicular joints. Mean (bold) and +/− one standard deviation joint kinematics of the thorax: top row, the sternoclavicular (SC) joint: middle row, and the acromioclavicular (AC) joint: bottom row (dominant side: blue, nondominant side: red).

abduction, external rotation, and extension. Both the elbow and wrist joints experienced significantly higher flexion moments than the GH joint.

4. Discussion

Our novel biomechanical model defined the UE, including the thorax, shoulder complex (humerus, scapula, and clavicle segments), elbow, and wrist for a comprehensive bilateral characterization [17]. The goal of this work was to quantify pediatric wheelchair mobility using new modeling techniques and evaluation approaches for the pediatric and young adult user.

The average self-selected speed of the children and young adults (1.23 m/s) was comparable to the self-selected speed of adult manual wheelchair users with SCI (1.07 m/s) in a study by Collinger et al. [16]. While self-selected speeds are important for evaluating typical, everyday activity, future work investigating pediatric propulsion should also consider assessing steady-state target speeds since biomechanical parameters may vary with speed [16].

The temporal-spatial parameters were compared with two other studies involving adult manual wheelchair users with SCI (Table 4) [22, 23]. The children and young adults in this study exhibited on average low cadences, which are recommended for injury prevention. However, they also experienced peak resultant, weight-normalized, handrim force similar to those of adult manual wheelchair users. While these force levels have been correlated to long-term pain and pathology in adult users, it is still unknown what effect similar loading would have on pediatric and young adult users. Additionally, while the average propulsion period angles were similar between this study and that of Gil-Agudo et al. (where this term is called "propulsion angle") and Kwarciak et al., the contact phase angles appear to be much smaller among pediatric and young adult users than the

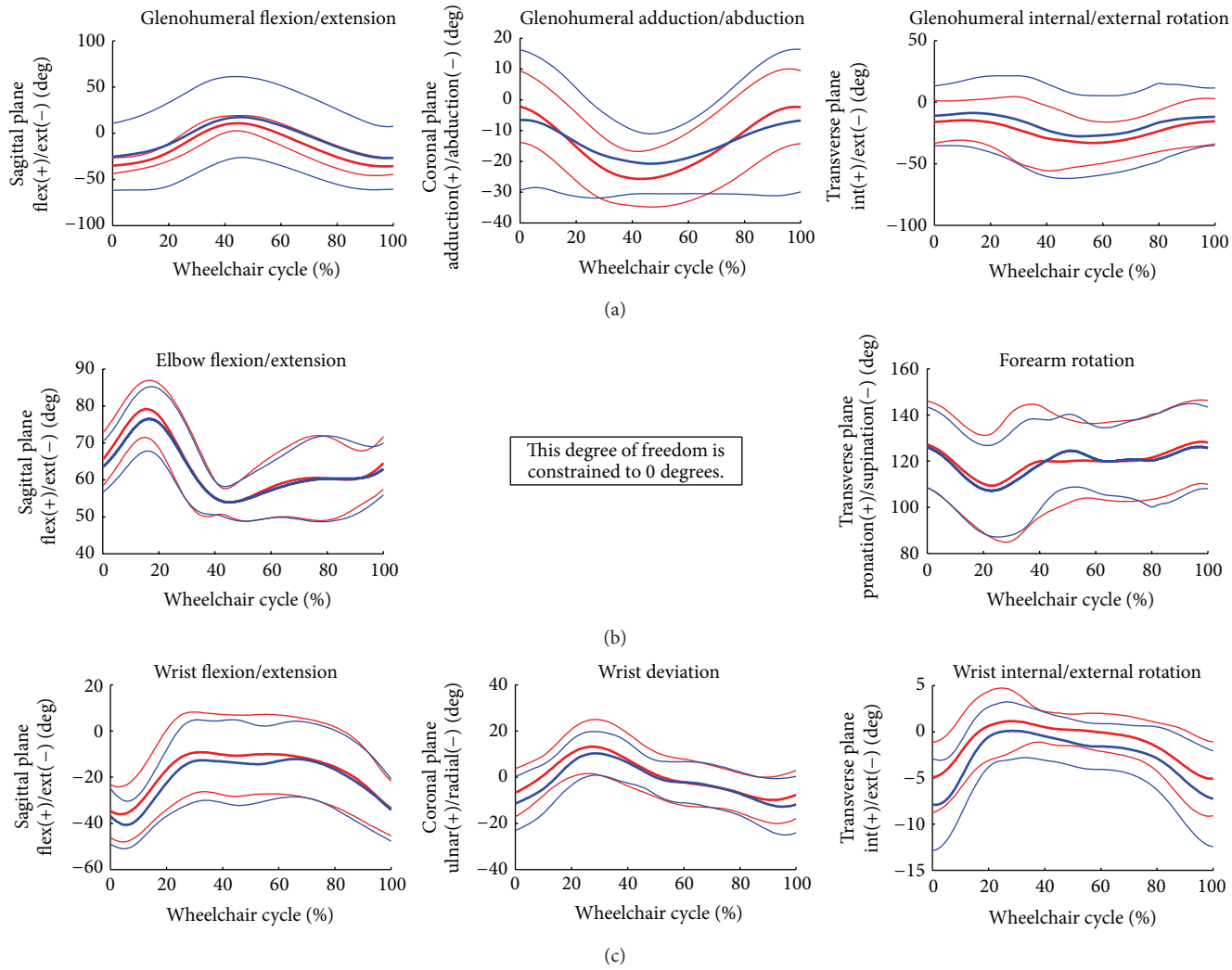

Figure 4: Group average joint kinematic data of the glenohumeral, elbow, and wrist joints. Mean (bold) and +/− one standard deviation joint kinematics of the glenohumeral (GH) joint: top row, the elbow joint: middle row, and the wrist joint: bottom row (dominant side: blue, nondominant side: red).

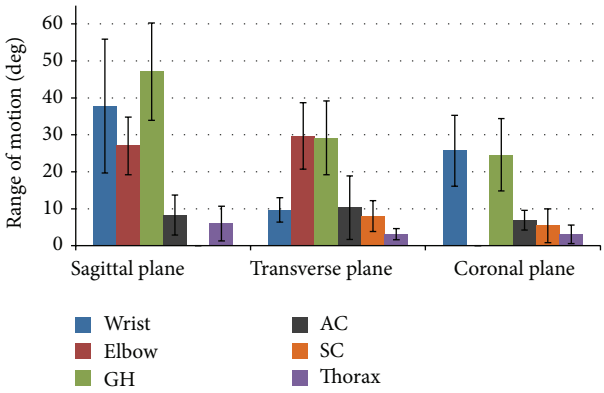

Figure 5: Group average joint ranges of motion (and standard deviation bars) for the thorax, sternoclavicular (SC), acromioclavicular (AC), glenohumeral (GH), elbow, and wrist joints in the sagittal, transverse, and coronal planes.

adults in the Boninger et al. (where this term is called "push angles") and Kwarciak et al. studies. Further evaluation is required to determine if smaller stature, greater efficiency, or other factors are the reason for this difference. Inconsistencies among cadence values as they relate to velocity, as our subjects had higher speed than the adults investigated in Gil-Agudo et al.'s study with lower cadence, further indicate a need to investigate temporal-spatial parameters of children and young adult manual wheelchair users [22, 23]. The correlation among these parameters, as well as biomechanical metrics, to other factors of clinical history, such as age, diagnosis and injury level, time since injury, and time of manual wheelchair use, may be particularly insightful.

The average relative time spent in the contact phase of the stroke cycle (35.8%) fell within the range commonly reported for adult manual wheelchair users of 30% to 50% [24]. It has been shown that increased relative time of the contact phase is

Table 2: P values for t-tests comparing all joint ranges of motion for each plane of motion. Italic cells indicate statistically significant differences ($P < 0.05$).

		Wrist	Elbow	GH	AC	SC
Elbow	Sagittal	0.076				
	Transverse	*<0.001*				
GH	Sagittal	0.168	*<0.001*			
	Transverse	*<0.001*	0.899			
	Coronal	0.783	*N/A*			
AC	Sagittal	*<0.001*	*<0.001*	*<0.001*		
	Transverse	0.834	*<0.001*	*<0.001*		
	Coronal	*<0.001*	*N/A*	*<0.001*		
SC	Transverse	0.270	*<0.001*	*<0.001*	0.411	
	Coronal	*<0.001*	*N/A*	*<0.001*	0.339	
Thorax	Sagittal	*<0.001*	*<0.001*	*<0.001*	0.27514	
	Transverse	*<0.001*	*<0.001*	*<0.001*	*0.015*	*0.002*
	Coronal	*<0.001*	*N/A*	*<0.001*	*0.002*	0.164

Table 3: P values for t-tests comparing peak joint forces for all directions and t-tests comparing peak joint moments for all rotations. Italic cells indicate statistically significant differences ($P < 0.05$).

	Force			Moment		
	Direction	Wrist	Elbow	Direction	Wrist	Elbow
Elbow	Anterior	0.161		Adduction	0.296	
	Superior	*<0.001*		Internal	0.389	
	Lateral	0.092		Flexion	0.145	
	Posterior	*<0.001*		Abduction	0.513	
	Inferior	0.921		External	0.980	
	Medial	0.289		Extension	*<0.001*	
GH	Anterior	*0.033*	0.279	Adduction	0.453	0.521
	Superior	*<0.001*	*<0.001*	Internal	0.153	*0.043*
	Lateral	*0.019*	0.128	Flexion	*0.009*	*0.001*
	Posterior	*<0.001*	*<0.001*	Abduction	0.052	0.085
	Inferior	*0.046*	0.063	External	0.076	0.075
	Medial	0.629	0.402	Extension	*<0.001*	*0.002*

Table 4: Average (St. dev.) subject characteristics and temporal-spatial parameters (NR means not reported).

Parameter	Children and young adults		Adults			
	This study	Kwarciak et al., 2009 [21]	Boninger et al., 2000 [22]		Gil-Agudo et al., 2010 [23]	
Age (yrs)	13.2 (5)	40.7 (11.3)	35.2 (9.3)		37.5 (9.8)	
Height (cm)	137.4 (29.9)	NR	176 (9.9)		170 (10)	
Weight (kg)	41.8 (13.4)	76.6 (16.4)	75.3 (17.9)		70.1 (10.9)	
Diagnosis	SCI	SCI	SCI		SCI	
			Speed 1	Speed 2	Speed 1	Speed 2
Velocity (m/s)	1.2 (0.26)	1.08 (NR)	0.98 (0.13)	1.65 (0.18)	0.833 (NR)	1.11 (NR)
Cadence (stroke/sec)	1.05 (0.19)	NR	1 (0.2)	1.3 (0.3)	1.1 (0.2)	1.2 (0.3)
Contact phase angle (deg)	85.6 (15.7)	98.38 (NR)	100.9 (16.5)	110.7 (14.7)	NR	NR
Propulsion period angle (deg)	72.6 (11.9)	80.64 (15.68)	NR	NR	66.3 (16.5)	69.7 (16.8)
Peak resultant force (%BW)	10.1 (3.7)	NR	9.6 (3.5)	13.9 (4.5)	8.6 (2.1)	10.6 (4.1)

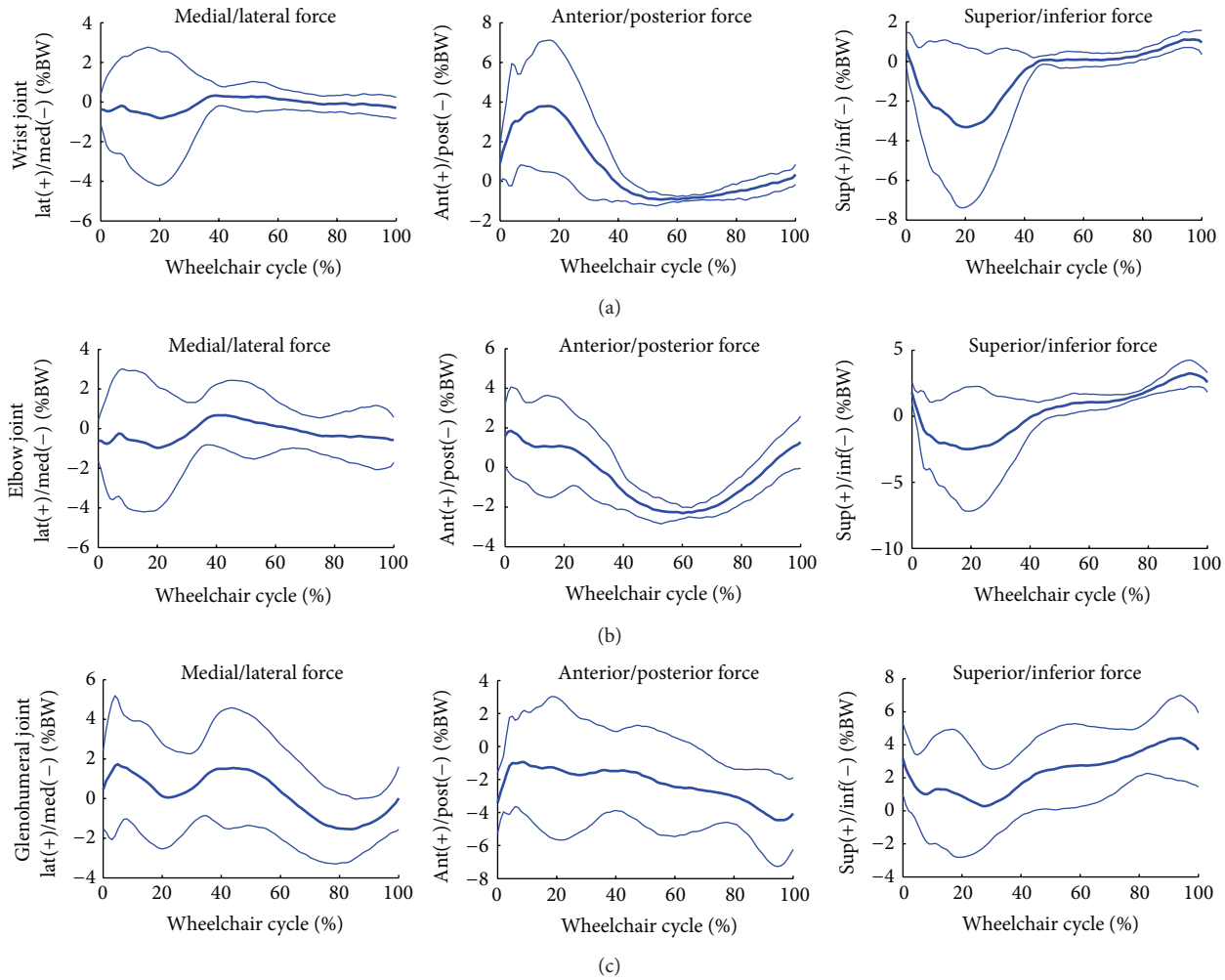

Figure 6: Group average joint force data of the dominant side wrist, elbow, and glenohumeral joints. Mean (bold) and +/− one standard deviation joint forces of the wrist joint: top row, elbow joint: middle row, and the glenohumeral (GH) joint: bottom row of the subjects' dominant side.

indicative of more challenging tasks, such as propelling with increased resistance or up a ramp [24], and while the propulsion evaluated here was not considered challenging, there were a few subjects whose relative time in the contact phase was around 45%. Perhaps this measure is indicating that this task was more demanding for these children. Likewise, we found some children whose relative time in contact phase was around 25%, slightly below the commonly reported range. As it is recommended to take long, smooth propulsive strokes [13], this shortened contact time with the handrim could result in higher force and rate of force application and thus be related to higher joint demands. Additionally, the model captured times of nonpropulsive moments on the handrim, indicating a braking effect, or nonefficient movements. The presented biomechanical methods may be used for evaluation of efficiency and training.

Propulsion patterns used by the pediatric and young adult population in this study were varied and subjects often either switched patterns between trials or did not clearly use one of the four common patterns described in the literature [12, 18], thus making identification of the primary pattern difficult. While the semicircular propulsion pattern is the recommended pattern for reduced joint loading and cadence [13], it has not yet been shown if this pattern is appropriate for the pediatric population. The semicircular pattern may not be appropriate or attainable for the pediatric propulsion due to physical limitations, wheelchair prescription or set-up (either originally or due to the growth of the child), or improper training. Given the patterns observed in this study, further investigation is warranted on appropriate propulsion patterns in the pediatric population of manual wheelchair users.

Overall, joint ranges of motion ranged from 3.1° to 47.1°, with the largest ROM at the glenohumeral joint during flexion/extension. Significant differences were found between joint ROMs of the glenohumeral, elbow, and wrist joints during internal-external rotation and flexion/extension. Peak joint forces ranged from 1.31% BW posteriorly at the wrist to 6.08% BW superiorly at the glenohumeral joint. Moments ranged from 0.1% BW × H of wrist extension to 1.36% BW × H of elbow flexion. These forces and moments are of

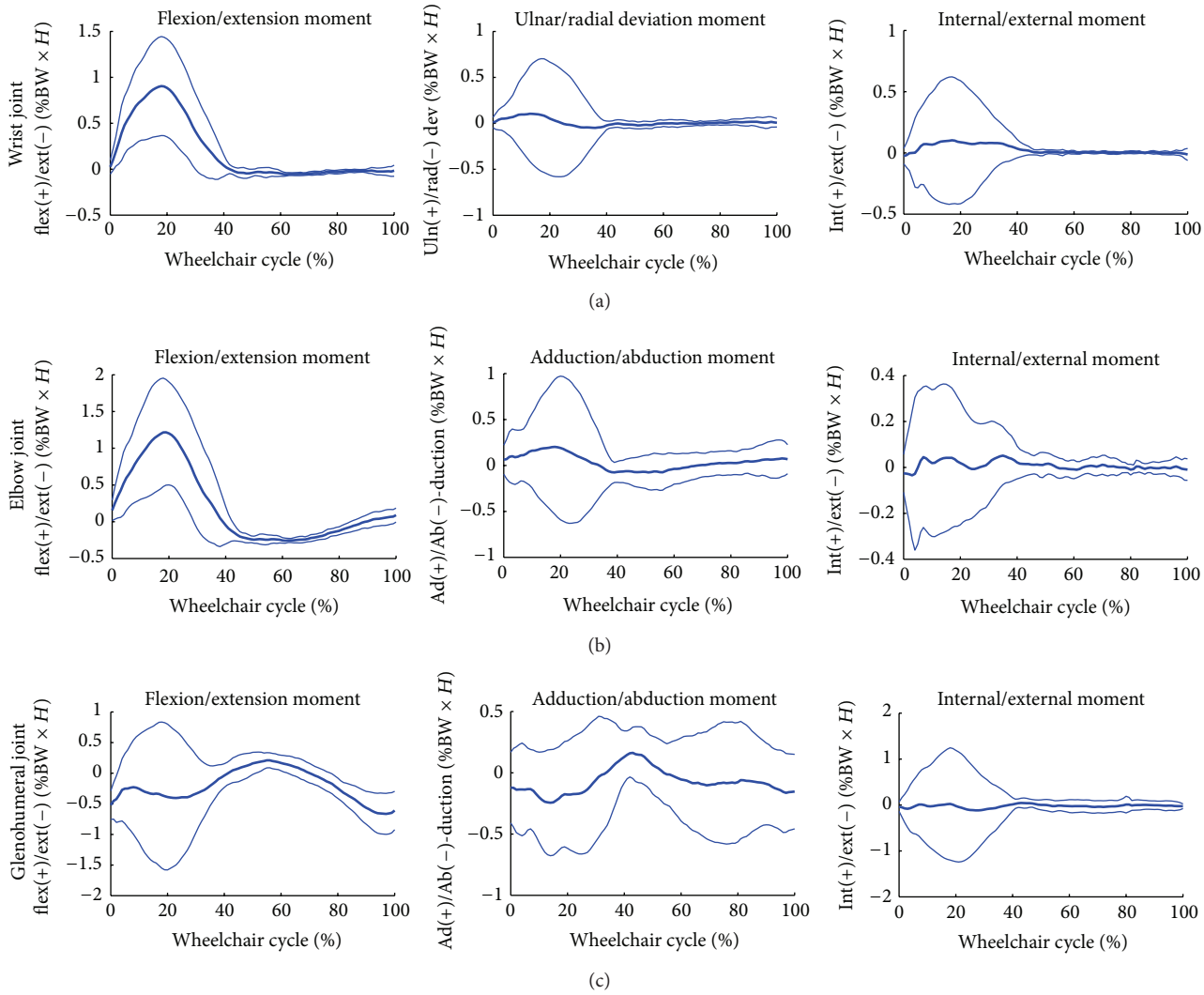

Figure 7: Group average joint moment data of the dominant side wrist, elbow, and glenohumeral joints. Mean (bold) and +/− one standard deviation joint moments of the wrist joint: top row, elbow joint: middle row, and the glenohumeral (GH) joint: bottom row of the subjects' dominant side.

concern in the pediatric population since they are similar to the magnitudes reported in adults [6, 23, 25].

These findings support continued quantitative evaluation of joint biomechanics for the prevention of pain and overuse injuries, of which these children may be at risk. The shoulder joint demonstrated the highest ROMs and forces compared to the distal elbow and wrist joints, as hypothesized. However, joint moments proved to be the highest at the elbow, followed by the glenohumeral joint and wrist joint. This may be due to mechanical inefficiency, lack of adequate training, and/or asymmetry. In addition, the variation of stroke patterns and speed may have impacted our group means. This further supports subject specific analyses in the future. This work has potential to be applied for in-depth quantitative evaluation of these factors. While much work has been conducted examining adult biomechanics during wheelchair use and the effects on upper limb injuries, particular consideration should be given to children. It is important that biomechanical evaluation methods are applied when prescribing, training, and providing long-term, transitional care of pediatric wheelchair users. Further work is underway to investigate the effects of age, duration of wheelchair use, muscular and soft tissue contributions, and level of injury. Ultimately, this work may lead to the development of guidelines for optimal pediatric wheelchair mobility.

5. Conclusions

This study presents findings for three-dimensional (3D) evaluation of joint dynamics of the shoulder complex, elbow, and wrist in children with SCI. Currently, pediatric wheelchair mobility biomechanics have not been reported. Evaluation methods for effective quantification of upper extremity joint dynamics during wheelchair propulsion have been presented.

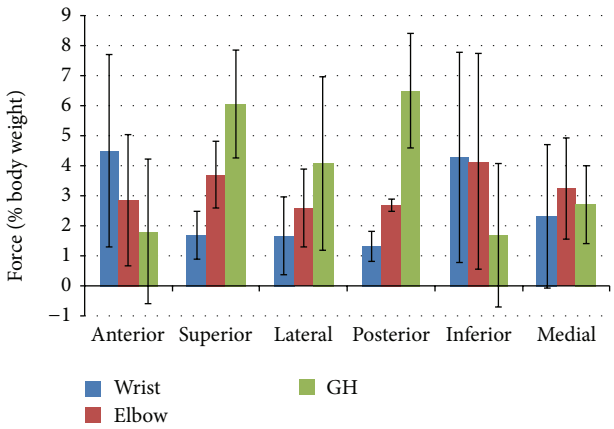

Figure 8: Group average peak joint forces (and standard deviation bars) of the subjects' dominant side glenohumeral, elbow, and wrist joints along the anterior-posterior, superior-inferior, and lateral-medial axes.

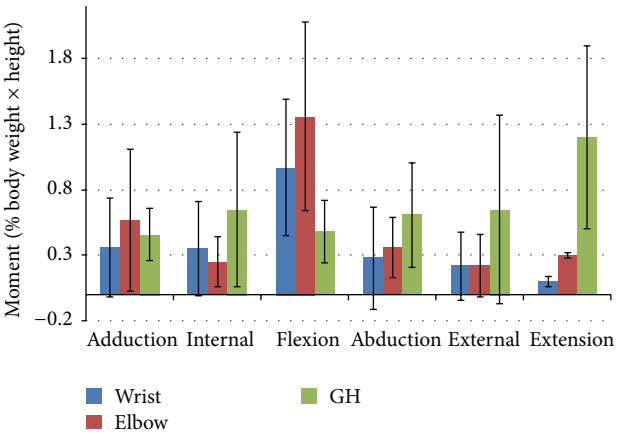

Figure 9: Group average peak joint moments (and standard deviation bars) of the subjects' dominant side glenohumeral, elbow, and wrist joints for adduction-abduction, internal-external, and flexion-extension rotations.

A population of 12 children with SCI was characterized. Joint ranges of motion and forces were found to be the greatest at the glenohumeral joint. Magnitude and frequency of joint demands are of concern for long-term manual wheelchair use in those with pediatric onset SCI. Parameters including range of motion, peak joint force, and peak joint moment should be assessed in 3D for all pediatric wheelchair users for rehabilitation planning. Further evaluation techniques and characterization should be investigated for prevention of pain and upper limb injuries. Particular attention should be given to this population of interest in regard to longer-term duration of wheelchair use and changes during development and maturation. Additional work is underway to correlate clinical history, pain, and functional outcomes to joint dynamics for further insight to provide patient-centered rehabilitation. Ultimately this work may lead to the development of pediatric guidelines for optimal wheelchair propulsion.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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Research Article

Design and User Evaluation of a Wheelchair Mounted Robotic Assisted Transfer Device

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Purpose. The aim of this study is to describe the robotic assisted transfer device (RATD) and an initial focus group evaluation by end users. The purpose of the device is to aid in the transfers of people with disabilities to and from their electric powered wheelchair (EPW) onto other surfaces. The device can be used for both stand-pivot transfers and fully dependent transfers, where the person being transferred is in a sling and weight is fully on the robot. The RATD is fixed to an EPW to allow for its use in community settings. *Method*. A functional prototype of the RATD was designed and fabricated. The prototype was presented to a group of 16 end users and feedback on the device was obtained via a survey and group discussion. *Results*. *Thirteen out of sixteen (83%)* participants agreed that it was important to develop this type of technology. They also indicated that user, caregiver, and robotic controls were important features to be included in the device. *Conclusions*. Participants in this study suggested that they would be accepting the use of robotic technology for transfers and a majority did not feel that they would be embarrassed to use this technology.

1. Introduction

The ability of people with mobility impairments to live in their homes and communities with maximal independence often hinges, in part, on their ability to transfer or to be transferred by an assistant. In order to help people with mobility impairments that cannot transfer independently live at home and participate in life's activities, insurance or government agencies may provide for personal attendant care services and in some cases provide stipends for family members providing these services. Further, independent transfers are a common source of upper extremity injuries and joint degeneration that often leads to the need for assistance with transfers over time [1]. Recent research has also shown that many people who can perform independent transfers need assistance when the height differential between transfer surfaces is greater than 75 mm or the gap between surfaces is greater than 150 mm [2]. For people who use power wheelchairs and need human and/or mechanical assistance with transfers, the options are limited. During dependent transfers with a human assistant, there is a high risk of injury (both acute and cumulative) to both the wheelchair user and the assistant, especially over the long term [1].

Between 1973 and 1987, 770 wheelchair-related accidents that led to death were reported to the US Consumer Products Safety Commission. 8.1% of these accidents were caused by falls during transfers [3]. Between 1986 and 1990, there were an estimated 36,000 wheelchair-related accidents in the USA that resulted in a visit to the emergency department. 17% of these accidents were due to falls during transfers [4]. In 2003, more than 100,000 wheelchair-related injuries were treated in US emergency departments, showing an upward trend in the number of injuries over time [5].

When caretakers assist in transferring wheelchair users, there is an additional risk of injury to the caretaker. In one study, of the 48 accidents reported by the 174 participants, 15.5% involved attendants [6]. There were more than 1,325,000 home care workers or clinicians in the United States in 2004. This group is expected to grow by 56% from 2004 to 2014 [7]. Lower back injuries are a major risk for this group, and one estimate found that 10.5% of back injuries in the United States are associated with transferring patients. In one study investigating bed to chair transfers, it was found that healthcare workers experience up to 3500 N of compressive forces during a single transfer [8]. In another study where lifts were implemented in a hospital to assist with patient transfers, it was found that over a 3-year period, there was a 70% decrease in claims cost at the intervention facility. The cost of compensation for injuries at this facility also decreased, with a 241% increase in the comparison facility [9]. Numerous studies have indicated that the US population will continue to age in the coming decades [10, 11] and that the prevalence of disability and impairment has remained high but stable [12].

There are approximately 1.5 million people in the United States who have disabilities that require them to use a wheelchair. One study found that 60% of people reported shoulder pain since beginning their wheelchair use. In comparison, only about 4.7% of the general population report regular shoulder pain [13]. Sitting pivot transfers (SPTs) are ranked among the most strenuous daily tasks of wheelchair users. Repetitions of this task over time can be detrimental to the shoulder and elbow joints of wheelchair users [14].

There are variations in wheelchair users' movements during transfers dependent on their level of injury. When patients transfer themselves from a wheelchair to another surface, most of their weight is initially supported by their trailing upper extremity. As they lose contact with the seat, weight is shifted to the leading arm [15]. During wheelchair transfers, large forces are placed on the shoulder and elbow joints. The leading shoulder encounters higher displacement and velocities than the trailing one [16]. This can cause damage in the leading arm to be accelerated and the onset of pain in this arm to occur sooner.

When wheelchair users are transferred by other people, the biomechanics of the transfer take on a different form. Strain is still placed on the wheelchair users shoulder joints, although it is more evenly distributed across the sagittal plane. There is also an additional factor of strain placed on the lower back of the person assisting with the transfer. One study found that a pivot transfer puts 112 lbs of force onto the clinician assisting with the transfer and raises their risk of developing a lower back disorder to 38.8% [7].

One technique that is used in many healthcare facilities is to move patients using ceiling lifts. In one study where lifts were added to an extended care unit, 71.4% of care staff reported that it became their preferred method of transferring patients and 96% believed that the ceiling lifts made lifting residents easier [17]. While these lifts effectively transfer people without placing as much strain on the caretaker, they are often not used because they are time consuming. In many cases, legislation concerning the implementation of lifts is focused on the caretakers' comfort and safety as opposed to the patients'. In rare cases, these lifts can even subject the patient to bruising or skin tearing. Another major concern when transferring patients using a lift system is that the patient may feel that being moved around in such a manner is undignified [18].

Few high tech devices for transfers are reported in the literature. One such device is the Home Lift, Position, and Rehabilitation (HLPR) chair, developed by researcher at NIST which aims to be able to lift wheelchair users, rotate them, and place them on a toilet, chair, or bed. It has been used to help evaluate how current and future standards could be applied to the HLPR and future robotic transfer devices [19, 20].

The aim of this paper is to describe the design, function, and a focus group evaluation of a novel device for assisting with transfers called the Robotic Assistive Transfer Device (RATD). The purpose of the RATD is to aid in the transfers of people with disabilities to and from their electric powered wheelchair (EPW) onto other surfaces such as a bed, shower bench, toilet, or another chair. The device can be used for both stand-pivot transfers, where the person has some ability to stand and places some weight on the ground, and fully dependent transfers, where the person being transferred is in a sling and weight is fully on the robot. The RATD is fixed to an electric powered wheelchair to allow for its use in both home and community settings. The overarching objective of this study is to engineer solutions to allow people who use power wheelchairs that require assistance (human or mechanical) while transferring to be able to transfer in their own homes, in the homes of friends/family, and in the community at large (e.g., hotels, restaurants, and shopping malls) in a safe, comfortable, efficient, and convenient manner.

The rationale for the conception of this device grew out of the described literature and through previous work on the Personal Mobility and Manipulation Appliance (PerMMA) [21–24]. PerMMA was developed as a test platform for assistive bimanual manipulation and advanced interfaces. While PerMMA and other assistive manipulators were capable of moving small household objects to aid in activities of daily living (ADLs), transfers to preform ADLs like bathing and toileting were not possible with existing hardware. The need for a strong, but less dexterous robotic arm, was perceived.

2. Methods

2.1. Design. The RATD's design allows for 5 powered degrees of freedom (DOF): two rotary joints, two prismatic joints, and track and carriage subsystem that allows the robot to translate around the seat frame of the wheelchair. When coupled to an EPW, the RATD has 7 overall DOFs. The design of the track and carriage is adapted from previous work on the PerMMA [21] robot and allows the RATD to be used on either side of the EPW seat, greatly increasing its workspace. It also allows the RATD to be stowed behind the seat without adding any width to the EPW when not in use. Proceeding from the carriage to the end effector, the first joint is the shoulder, which rotates internally toward the user or externally away

from the user, which is shown in the left and center panel of Figure 3. The shoulder is connected to the proximal segment that contains a prismatic joint. This segment is along the axis of rotation of the shoulder and extends the robots workspace vertically. The proximal segment is connected to the distal segment by an elbow joint, as seen in Figure 2. The distal segment also contains a prismatic joint that allows the end effector to extend away from the elbow.

The robot is powered electromechanically by a combination of planetary gear motors and linear actuators. The carriage is moved around the track using a 24 V, 0.52 A planetary gear motor with a 100 : 1 gear ratio, which is connected to spur gear that propels it along a rack machined in the center of the face of the track. Mechanically, the shoulder joint is a 1.25 inch diameter steel shaft that is fixed to the proximal segment and connected to the carriage with a tapered bearing. It is actuated by a 24 V, 2.2 A planetary gear motor, with a 326 : 1 fixed to the carriage that has a spur gear that pushes another spur gear attached to proximal segment. Proximal and distal segments are identical in construction and are made up of two concentric hexagonal bodies that are able to slide past each other. The bodies are composed of nylon plastic shells created using selective laser sintering (SLS), stainless steel threaded rods, and aluminum end plugs. The combination of elements provides the bodies with strength; the double walled nylon shells provide the compressive strength and the stainless steel threaded rods provide the tensile strength. The aluminum end caps allow threaded rods to be held and tensioned. The concentric bodies are coupled together with a 2500 N linear actuator (Linak, L30) with 250 mm stoke length. Pins inserted through the end plugs and through the clevis ends of the actuator hold the assembly together. An elbow joint connects the proximal and distal segment to each other. A linear actuator (Linak, L30) crosses the joint and powers the elbow to move from 35 degrees to 100 degrees vertically. All three actuators have a spline and nut that prevents them from being back driven. Attached to the end of distal segment is a load cell and handle. Also, attached to the distal segment is a double hook on a swivel, which is used to hang the loops of a transfer sling.

The RATD is equipped with force and position sensors. The position of each joint is tracked using a microcontroller equipped, absolute encoder with digital output (Model A2, US Digital, Vancouver, WA). Two absolute inclinometers with digital output (Model A2T, US Digital, Vancouver, WA) are placed on the base of the wheelchair to determine the angle at which the wheelchair is sitting with respect to gravity. The encoders and inclinometers are able to be daisy-chained to form a network called a Serial Encoder Interface (SEI) bus, which allows data from multiple devices to be transmitted using only four lines. Force sensing is done in two places: at the base of the proximal segment and at the handle. The 6 DOF load cell (Model Omega, ATI-IA, Apex, NC) at the base of the proximal segment can withstand high torque and serves as the primary measurement tool for load on the arm. The second 6 DOF load cell (Model Delta, ATI-IA, Apex, NC) is located between the end of distal segment and the handle. Its primary purpose is to serve as an input device for controlling the arm in conjunction with the handle.

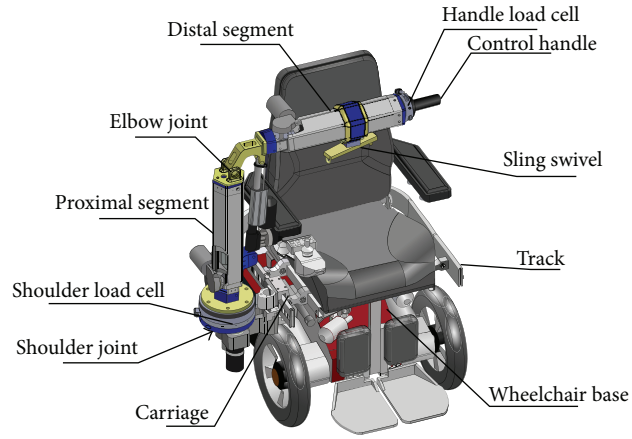

Figure 1: An annotated solid model showing the key mechanical features of the RATD.

The core electronic components that drive the arm consist of a single board computer (SBC) (Model Cobra, VersaLogic, Tualatin, OR), an analog to digital converter board (Model VCM-DAS-2, VersaLogic, Tualatin, OR), an SEI bus to USB converter (Model SEI-USB, US Digital, Vancouver, WA), and a custom designed relay board, as shown in Figure 4. The SBC provides the programmability, memory storage, and data bus capability to the system. The relay board is used to translate low current digital logic signals from the SBC into high current switching needed to control the motors and linear actuators that power the robot's joints. In addition to receiving computer based signals, the relay board is also capable of accepting inputs from a mechanical switch array to drive each joint. The analog to digital converter is used to digitize the signals from the load cells for use in the control algorithm. Similarly, the SEI to USB converter receives the signals from the encoder network and allows them to be read through a USB port on the SBC to be used in control algorithms. The electronics are powered via a DC-DC converter, which steps wheelchair batteries from 24 v down to ±12 v and 5 v.

The device can be controlled by two different methods by the caregiver: a switch pad or through a force sensing handle method called Direct Interaction. For the switch pad, the carriage and the 4 DOF of the arm are controlled individually with two switches for each DOF, one for each direction of motion. The hardware for the RATD does not have the ability to perform proportional speed control, so motor motion is either on or off. Direct interaction uses a load cell to receive caregiver force inputs through the control handle, visible in Figure 1. The force inputs are mapped to different DOF in an intuitive way to move the RATD. Once one DOF is activated, it locks out the other DOF until the force is removed. The algorithm and force mapping are described in detail by Jeannis et al. [25].

The framework for conceptualizing the safety aspects of RATD is made up of 4 layers. The first of these layers consists of mechanical features, including shrouding of pinch points; rounded edges of metal and plastic surfaces; padding in strategic areas; and compliance, which allows the robot

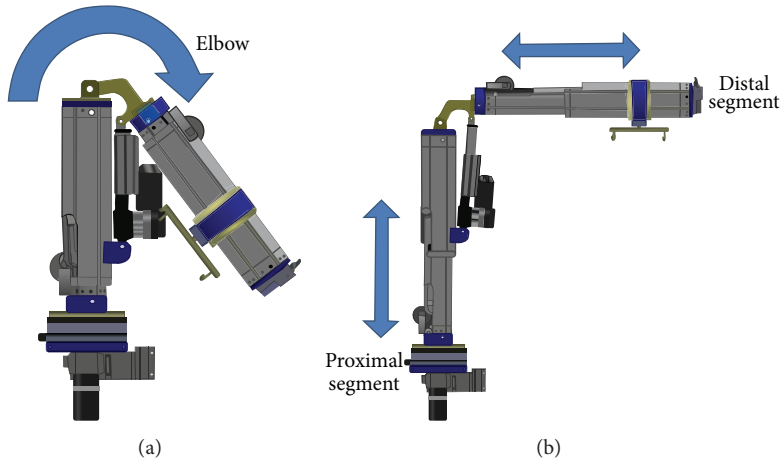

Figure 2: A solid model showing RATD's axis of motion for the shoulder, proximal segment, and distal segment joints.

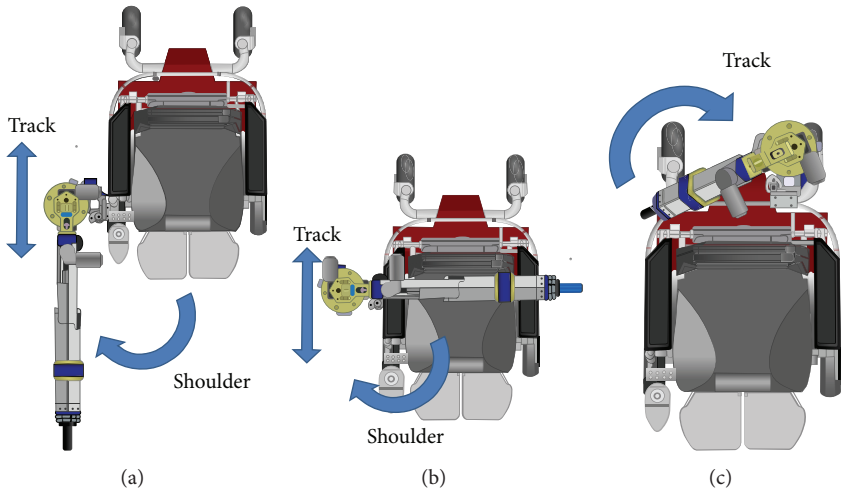

Figure 3: A solid model showing RATD's axis of motion for the shoulder and track joints.

to elastically bend under certain loading conditions. The second layer includes electronic features, including limit switches, hard force limits, hard speed limits, and user initiated emergency stops. The third layer is made up of software features, which allows for the programming of soft force limits, soft speed limits, keep-out zones, and the ability to limit the rate of loading. The fourth layer consists of the human caregiver, who has the ability to observe and make decisions regarding safety.

Stability during transfers is a key safety aspect of the RATD. Force sensors, computational ability, known kinematics, and controlled actuation allow for the use of a dynamically calculated stability boundary that limits the workspace based on payload. A quasistatic mathematical model of the RATD attached to a C500, developed and verified by Wang et al. [26], accounts for the COM of the C500 and the payload on the RATD and determines the safe workspace boundary for that payload. An additional algorithm tracks the kinematics of the RATD and can prevent it from moving into a region that would cause a tip-over.

A demonstration protocol was created to determine if the RATD could perform transfers. A 185-pound extrication dummy known as "Survivor" (Dummies Unlimited, Inc., Pomona, CA) was used as a surrogate for the wheelchair user. Four surfaces were identified: a mat table, a shower bench, a toilet in a restricted space, and soft chair with arms. The transfers were performed to and from each object and with both the key pad and Direct Interaction. The tasks were evaluated as pass-fail, the transfer was considered complete when Survivor was situated in the middle of the surface, and the time period was unlimited.

2.2. Focus Group Protocol. In order to obtain qualitative feedback regarding the concept for the RATD, a focus group was conducted. A convenience sample of 18 participants was recruited at the 2011 National Veteran Wheelchair Games in Pittsburgh, PA. In order to participate, participants had to report that they used some type of wheeled mobility as primary means of mobility. This was to include people who normally transferred dependently and independently

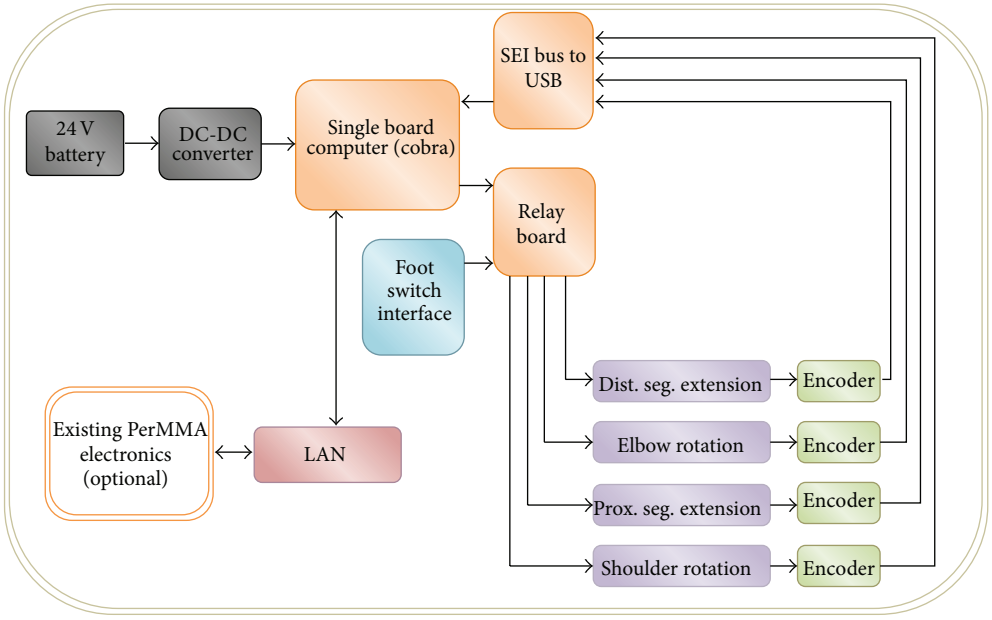

Figure 4: A block diagram describing RATD's motors, sensors, and associated electronics.

to better determine who might use the device and in what context. After obtaining written informed consent, each person was asked to fill out a presurvey that asked questions regarding their demographic information, types of assistive technology (AT) they used, and their satisfaction with that AT. Following the presurvey, the participants were shown a live demonstration of the RATD and an explanation of the device by the design team. Participants were given the opportunity to ask the design team questions. A moderator, who was not involved with the design of the device, then led a group discussion of the device. The moderator probed the group as to what features of the device they like or disliked, what features they would like to see added, and, if they would use the device, in what context would they use the RATD. The conversation was recorded using a digital recorder. Following the group discussion, the participants were asked to fill out a postsurvey that asked questions related to the RATD and gave an additional opportunity to make general comments about the device. The postsurvey contained a set of questions in which the participants were given a design feature related to the RATD and were asked to rate on a 7-point Likert scale if the feature would make them less likely to want the device (1) or more likely to want the device (7). It also contained a second set of questions in which the participants were given a statement and asked to what extent they disagreed (1) or agreed (7) with the statement on a 7-point Likert scale. The focus group lasted about an hour and a half, from start to finish.

For the purpose of analysis, the Likert scale responses were collapsed. For the question on product features, responses of 1 and 2 were categorized as "less likely," 3, 4, and 5 as neutral, and 6 and 7 as "more likely." For the statement questions, responses of 1 and 2 were categorized as "disagree," 3, 4, and 5 as "neutral," and 6 and 7 as "agree." The responses were compiled using Excel and a descriptive analysis of the data was completed using SPSS.

Table 1: The disability and frequency of participants.

		Count
Disability	SCI	9
	Amputation	1
	MS	2
	TBI	1
	TBI and amputation	1
	Back injury	1
	Hemiparalysis	1

3. Results

Figure 5 shows a sequence of photographs that demonstrate a caregiver using the RATD to transfer a wheelchair user from an EPW to a mat table. The RATD was able to perform all the transfers during the demonstration protocol.

In the stowed position, the RATD fits within the footprint of the C500 and can fit through any doorway that a C500 without an RATD can fit through, as shown in Figure 6.

Of the 18 participants recruited, 16 finished the study and an analysis was performed using data from only the participants that finished the study. The group consisted of 11 males and 5 females, all of whom were Veterans. The participants were an average of 20 ± 13 years post onset of disability. 8 participants used manual chairs and 8 participants used powered mobility. The types of disabilities represented in this study are given in Table 1.

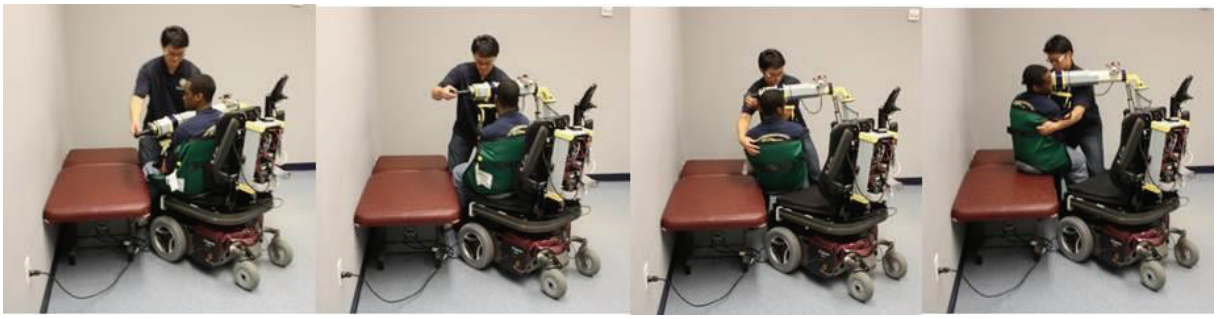

Figure 5: A sequence of photographs of the RATD being used to transfer a person from an electric powered wheelchair to a mat table, by a caregiver.

Table 2: The responses to the survey questions related to product features.

Product feature	Responses (n = 16 unless noted)		
	Less likely to want it	No difference	More likely to want it
(A1) A transfer device attached to a power wheelchair.	1 (6%)	9 (56%)	6 (38%)
(A2) A transfer that can be controlled by a caregiver.	1 (6%)	5 (31%)	10 (63%)
(A3) A transfer device that can be controlled by a computer program. (n = 14)	1 (7%)	6 (43%)	7 (50%)
(A4) A transfer device that can be controlled by the user.	1 (6%)	5 (31%)	10 (63%)

When asked "How much money out of pocket would you pay for the RATD?" the participants responded with an average of \$1407.69 ± 2416.42 and a range of \$0–8,000. A histogram showing the distribution is given in Figure 7. Three participants declined to answer this question. When asked if having a transfer device attached to a wheelchair would make them more or less likely to want it, 1 (6%) responded with less likely, 9 (56%) responded with no difference, and 6 (38%) responded with more likely. When asked if having a transfer device controlled by a caregiver would make them more or less likely to want it, 1 (6%) responded with less likely, 5 (31%) responded with no difference, and 10 (63%) responded with more likely. When asked if having a transfer device controlled by a computer program would make them more or less likely to want it, 1 (7%) responded with less likely, 6 (43%) responded with no difference, and 7 (50%) responded with more likely, with two participants declining to answer the question. When asked if having a transfer device controlled by the user would make them more or less likely to want it, 1 (6%) responded with less likely, 5 (31%) responded with no difference, and 10 (63%) responded with more likely. A summary of these responses is given in Table 2.

The results of the survey pertaining to agreement with a particular statement are summarized in Table 3.

Three notable themes were brought up during the group discussion. The first was that the device would be especially good for travel. The RATD would minimize the amount of equipment that would need to be transported and that it would be easier to adapt to bathrooms that have less than ideal accessibility. The second is that the device should also be available with a user interface, so that persons with a disability could transfer themselves without a caregiver. It was noted in the discussion that the RATD could provide a range of transfer assistance from dynamically adjustable grab bars, through stand-pivot transfers, to fully dependent sling transfers. The participants suggested that those needing less assistance would likely want to control the RATD themselves. Lastly, the participants indicated dissatisfaction with current sling technology for dependent transfers and that the RATD might open up new possibilities for improved slings or harnesses for both dependent and stand-pivot transfers.

4. Discussion

4.1. Design. During the demonstration protocol the RATD was able to be used to transfer Survivor from the wheelchair to all the surfaces and back. The initial position of the wheelchair was critical to being able to complete the transfer. This suggests that either the RATD workspace needs to be increased or additional aids are needed to locate the EPW efficiently near the surface. While no strict scientific evaluation was performed, Survivor behaved well as a transfer surrogate. In the future, if a transfer surrogate can be validated and standardized, it will greatly aid algorithm development and progress in the field.

The number and types of degrees of freedom were selected deliberately when creating the RATD and much information was drawn from prior work with the PerMMA project. While working with PerMMA it was observed that humans can control prismatic joints better than rotational joints. Since human control is an important aspect of the RATD concept, effort was made to include multiple prismatic joints to reduce cognitive load on the caregiver. The arm portion of the system has 4 DOFs, which may seem counterintuitive, since this leaves the arm highly constrained. However, unlike most robotic arms, which are designed for finer manipulation, the task of moving a person is gross movement and the typical wrist-like DOFs in other

Table 3: The responses to the survey questions related to agreement or disagreement with a statement.

Statement	Responses (n = 16 unless noted)		
	Disagree	Neutral	Agree
(B1) I would choose to use the RATD.	4 (25%)	9 (56%)	3 (19%)
(B2) Using the RATD would make my life easier.	3 (19%)	9 (56%)	4 (25%)
(B3) Leaning to use the RATD would be easy for me.	1 (6%)	6 (38%)	9 (56%)
(B4) I would be anxious about using the RATD.	6 (38%)	8 (50%)	2 (13%)
(B5) It would be embarrassing to be seen using the RATD. (n = 15)	11 (73%)	3 (20%)	1 (7%)
(B6) It would be easier to just get another person to help rather than use the RATD.	6 (38%)	5 (31%)	5 (31%)
(B7) It is important that we develop technology that can do this.	0 (0%)	3 (19%)	13 (81%)

(a) (b)

Figure 6: The RATD in its stowed position. (a) It is shown while passing through a doorway and (b) it is shown from above.

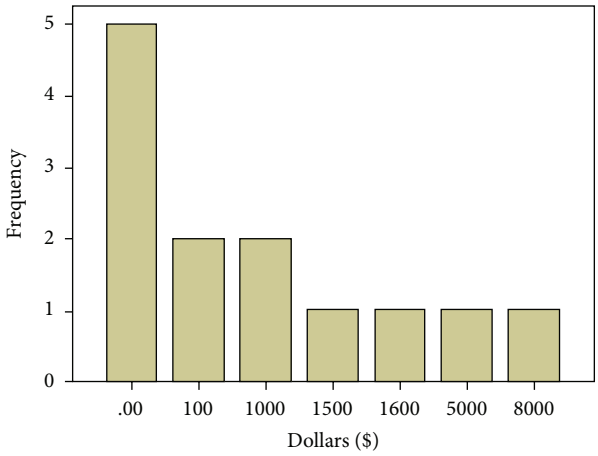

Figure 7: Histogram of how much the participants were willing to pay out of pocket for RATD in dollars.

robotic arms are not necessary. Fewer DOFs reduce control complexity and save physical space, which is paramount on a mobile device.

4.2. Focus Group. The responses to the survey yielded some notable results. In regard to product feature Table 2 A1, the majority of the participants were either neutral or supportive of the idea of having a transfer device attached to a power wheelchair, with only small minority objecting to this idea. This suggests that there is not a categorical bias against having a combination mobility and transfer device. Product features Table 2 A2, A3, and A4 were aimed at determining what types of controls the participants were comfortable with, especially contrasting computer/robotic control of the device versus the more traditional user or caregiver control that is used on typical assistive devices. Given that the responses to all three types of controls were similar, this suggests that people are not categorically biased against computer programs controlling their device and that several control methods are likely necessary to accommodate different people and the different contexts for which they might use a transfer device.

The responses to statements Table 3 B3, B4, and B5 also suggest that the participants would be accepting this robot technology. With statement Table 3 B3, the majority of the participants agreed that they would be able to learn how to use the RATD, which is contrary to the common perception of robots as complicated. Possible explanations for this might be that people are growing more comfortable with high tech devices or that the limited number of inputs and prismatic joints make the RATD more manageable to operate. With statement Table 3 B4, the majority of participants suggested that they would not be anxious or would have neutral

feelings when using the RATD, which again may be contrary to common perceptions of robots. This may reflect that participants are at least willing to trust a robotic transfer device but may be cautious while doing so. With statement B5, a strong majority of participants indicated that they would not be embarrassed to use the RATD, suggesting that the participants do not perceive any negative social bias toward the device.

The response to statement Table 3 B6 suggests a possible weakness of the RATD. The group was split on whether seeking additional caregiver help would be easier than using the device. While evidence strongly indicates that transferring without properly used equipment is dangerous, this response suggests that humans are still considered an alternative to transfer technology by people with disabilities. Until transfer technology overcomes the speed and adaptability of humans, this perception will likely persist and is a key challenge for developers of transfer devices.

In order to better interpret the results, some discussion of the participants is warranted. While all the participants used wheeled mobility, some had the ability to independently transfer some needed partial assistance, and others were completely dependent on caregivers for transfers. For survey questions such as Table 3 B1, the participants' ability to transfer likely influenced their response. Future work should focus specifically on people who need some sort of assistance for transfers and in what context they would use the device. However, a strong majority agreed with statement Table 3 B7 that a transfer device with RATD capabilities was important to develop. This suggests that while some of the participants might not have a current need for the device, they could see that others might be able to benefit from it or that they might be able to benefit from it as their abilities change in the future. In regard to how much the user would be willing to pay out of pocket, most of the participants indicated that they would pay little or no money out of pocket for the device. This suggests that the participants expect 3rd-party payers to fund the device.

As noted in the methods, a convenience sample was used and this may have led to participants having a preferential bias toward new technology and may have given more positive answers than the general population. Also, the data was collected from individuals at a recreational event and this again may have resulted in a biased sample toward people who are active and may have excluded those who are not able to be. Including the less active might have resulted in identifying other transfer-related barriers. Lastly, while precautions were taken to minimize the role of the design team, they were present in the room for the presentation and discussion. The participants' perception of the design team may have influenced the participants' attitude toward the device in positive or negative way.

It should be noted that this study has several limitations including small sample size, a relatively homogenous sample, and the inherent limitations of qualitative data. In regard to the homogeneous sample, all participants were Veterans, were predominately male, and all had acquired conditions. However, the population that may benefit from this device is likely very heterogeneous. Future design development should focus on improving controls for caregivers; creating user controls; further refinement of algorithms for tip-over stability to include nonlevel surfaces; and optimizing the device for cost, size, aesthetics, and reliability. Future experimental studies should focus on comparing the device to existing technology and the role of caregivers.

Conflict of Interests

The authors report no conflict of interests.

Acknowledgments

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Research Article

Trunk and Shoulder Kinematic and Kinetic and Electromyographic Adaptations to Slope Increase during Motorized Treadmill Propulsion among Manual Wheelchair Users with a Spinal Cord Injury

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The main objective was to quantify the effects of five different slopes on trunk and shoulder kinematics as well as shoulder kinetic and muscular demands during manual wheelchair (MWC) propulsion on a motorized treadmill. Eighteen participants with spinal cord injury propelled their MWC at a self-selected constant speed on a motorized treadmill set at different slopes (0°, 2.7°, 3.6°, 4.8°, and 7.1°). Trunk and upper limb movements were recorded with a motion analysis system. Net shoulder joint moments were computed with the forces applied to the handrims measured with an instrumented wheel. To quantify muscular demand, the electromyographic activity (EMG) of the pectoralis major (clavicular and sternal portions) and deltoid (anterior and posterior fibers) was recorded during the experimental tasks and normalized against maximum EMG values obtained during static contractions. Overall, forward trunk flexion and shoulder flexion increased as the slope became steeper, whereas shoulder flexion, adduction, and internal rotation moments along with the muscular demand also increased as the slope became steeper. The results confirm that forward trunk flexion and shoulder flexion movement amplitudes, along with shoulder mechanical and muscular demands, generally increase when the slope of the treadmill increases despite some similarities between the 2.7° to 3.6° and 3.6° to 4.8° slope increments.

1. Introduction

There has been a growing interest in motorized treadmill manual wheelchair (MWC) propulsion in recent years in rehabilitation research environments, and to a lesser extent in clinical practice, since it seems to closely duplicate overground MWC requirements [1]. Motorized treadmill MWC propulsion also allows for propulsion, in a restricted space, during short (e.g., high intensity interval training) and prolonged (e.g., cardiorespiratory fitness training) periods of time at different speed or slope parameters. Moreover, unlike propulsion on a roller ergometer or a dynamometer, MWC propulsion on a motorized treadmill allows for greater freedom of MWC movements and some inertial effect exposure linked to the acceleration/deceleration of the wheelchair and head-trunk-upper limb segments. Hence, motorized treadmill MWC propulsion is a promising therapeutic alternative founded on the principle of repetitive task-specific training and anticipated sensorimotor adaptations. The quality and quantity of evidence currently available on motorized MWC propulsion do not inform rehabilitation professionals about how to vary parameters that can be easily modulated (e.g.,

the speed or slope of the treadmill) and their effects during motorized treadmill MWC propulsion. Stronger evidence is needed to optimize assessment and training protocols.

Propelling a MWC up slopes on a motorized treadmill has been found to increase upper limb demand in a few recent studies. Richter et al. [2] found among MWC users with a spinal cord injury (SCI), propelling at a self-selected velocity on a motorized treadmill, that the speed was about 1.5 and 2.7 times slower when pushing up 3° and 6° slopes, respectively, in comparison with the level surface (0° slope). Additionally, they also reported that the peak total force at the handrim was about 1.7 and 2.2 times higher when pushing up 3° and 6° slopes, respectively, in comparison with the level surface (0° slope) despite the reduced treadmill speed. Yang et al. [3] found among MWC users with a SCI, propelling at an imposed steady speed of 0.9 m/s on a motorized treadmill, that peak shoulder flexion increased by 9° while the total and tangential forces applied at the handrim were about 2.09 and 2.38 times higher when pushing up a 3° slope (approximately a 1 : 20 ratio) in comparison with the level surface, whereas the mechanical efficiency was only found to be 1.1 times higher. More recently, Gagnon et al. [4] found among MWC users with a SCI, each propelling at a self-selected steady speed on a motorized treadmill up slopes set at 0°, 2.7°, 3.6°, 4.8°, and 7.1°, that the total and tangential forces applied at the handrim were at least 2 times greater as the slope became progressively steeper. The greatest change observed was between 0° and 2.7°, while similarities were observed between 2.7° and 3.6°. Such differences also support the need to gain a better understanding of the effects of steeper slopes on trunk and shoulder kinematics, shoulder kinetics, and shoulder muscular demand during uphill propulsion on a treadmill set at a steady speed among a group of experienced MWC users.

In addition, few studies have investigated the effects of varying slopes during overground MWC propulsion over the past decade. Among those, Chow et al. [5] investigated the effects of various slopes (i.e., 0°, 2°, 4°, 6°, 8°, 10°, and 12°) on trunk kinematics, handrim kinetics, and upper limb and upper trunk muscular demand among MWC users with a spinal cord injury. Overall, forward trunk flexion and muscular demand (i.e., triceps brachii, anterior deltoid, and pectoralis major) were found to progressively increase as the slope became steeper. van Drongelen et al. [6] compared shoulder net joint moments between overground level and uphill (i.e., 3° slope) propulsion among MWC users with paraplegia and tetraplegia. The resultant net shoulder moments were about 2 times higher during uphill compared to overground level MWC propulsion. Arabi et al. [7] compared relative mechanical demand during uphill propulsion across three slopes (i.e., 2.7°, 4.8°, and 5.7°) among a group of able-bodied individuals who used a MWC. They confirmed that the relative mechanical demand significantly increased as the slope became steeper and reached 16.1%, 25.7%, and 31.1% of their maximum isometric voluntary force generating capability for the three slopes tested, respectively. Last, Kulig et al. [8] compared shoulder net joint moments between level overground and uphill (slope = 4.7°) MWC propulsion in individuals with paraplegia. They showed that the peak net shoulder flexion, adduction and internal rotation moments were 2.2, 2.2, and 2.7 times greater, respectively, when ascending a slope compared to level overground wheelchair propulsion.

The aim of this study was to quantify the trunk and nondominant shoulder kinematic changes along with the nondominant shoulder joint moments and electromyographic changes during MWC propulsion on a motorized treadmill set at a self-selected natural speed on five different slopes (i.e., 0°, 2.7°, 3.6°, 4.8° and 7.1°). It was expected that the nondominant shoulder and trunk range of motion and nondominant shoulder joint moments and muscle electromyographic activity would gradually and significantly increase with each slope increment while the speed remained constant.

2. Material and Methods

2.1. Description of Participants. A convenience sample of 17 men and 1 woman who sustained a SCI (American Spinal Injury Association Impairment Scale [9] (AIS) = A, B, C or D) volunteered to participate in this study (Table 1). Participants were included in the study if they had sustained a SCI at least three months before the study, had been discharged from initial intensive inpatient rehabilitation, were living in the community and used their MWC for at least 4 hours per day. Participants also had to master basic and advanced wheelchair skills, including the capability to propel up a 9-metre long access ramp meeting building code standards in the province of Quebec in Canada (i.e., maximum slope of 1 : 12 for slopes of a maximum length of 9 metres) [10]. Participants were excluded if they presented associated neurological conditions, musculoskeletal impairments/pain, cardiorespiratory/vascular conditions, or any other impairments or disabilities that might have interfered with the performance or safety of the experimental tasks. The self-reported Wheelchair User's Shoulder Pain Index (WUSPI) questionnaire [11, 12] was completed (group's mean score = $0.89 \pm 1.05/10$) and reviewed by a physical therapist who asked specific questions whenever pain was rated as interfering with the performance of wheelchair mobility to further verify that pain will not limit their ability to specifically complete the experimental tasks. During a telephone interview with the potential participants, the rehabilitation research coordinator reviewed the inclusion and exclusion criteria to determine eligibility before scheduling the clinical and laboratory assessments. All participants gave their written consent to participate in the study after being informed of the objectives and nature of their participation in the study. The Research Ethics Committee of the Centre for Interdisciplinary Research in Rehabilitation of Greater Montreal (CRIR #715-0312) approved the present study.

2.2. Clinical Evaluation. Each participant underwent a clinical assessment, completed by a physical therapist, in order to collect their personal characteristics, measure their anthropometric parameters (height, weight, length, and circumference of body segments), characterize the severity of the sensory and motor impairments (ASIA Impairment Scale [13]),

Table 1: Description of participants.

Participants	Gender	Age years	Height m	Mass kg	Time since injury years	Neuro-logical level	ASIA*			WUSPI*		
							AIS*	Sensory/ 224	Motor/ 100	Mean/ 10	Propulsion 10 min/10	Slope/ 10
1	M	44.3	1.84	80.3	10.6	T7	A	117	50	0.19	0.0	0.0
2	M	46.4	1.70	80.2	4.6	T10	B	140	50	0.68	0.1	0.5
3	M	32.2	1.92	95.9	5.3	T10	A	140	50	0.00	0.0	0.0
4	M	35.8	1.80	77.1	11.8	T6	D	194	81	1.25	1.9	2.2
5	M	33.2	1.95	72.3	7.8	T12	C	162	56	0.19	0.0	0.0
6	M	52.6	1.77	108.9	18.7	T9	A	132	50	1.14	2.4	3.4
7	M	59.9	1.88	99.8	5.0	T10	A	140	50	0.34	0.4	2.2
8	M	44.0	1.72	68.4	22.1	T4	B	183	35	0.07	0.9	0.0
9	M	41.2	1.78	72.7	6.1	C7	C	56	44	1.23	2.7	2.7
10	M	28.4	1.85	66.6	10.6	T12	A	154	50	0.63	3.8	1.9
11	M	39.0	1.76	101.8	2.8	T10	A	72	50	3.65	5.5	6.8
12	M	49.1	1.70	76.8	4.4	T7	A	88	52	0.97	1.5	2.4
13	M	55.7	1.80	103.1	4.9	T3	A	88	50	0.31	1.8	1.6
14	M	32.8	1.75	61.9	8.9	T4	A	95	50	0.10	0.4	0.4
15	F	28.1	1.65	47.5	4.8	T11	A	148	50	0.20	2.0	1.0
16	M	33.0	1.65	66.5	5.3	T6	A	53	50	0.10	0.0	0.3
17	M	52.7	1.73	78.2	8.9	T12	B	172	63	3.12	2.6	4.8
18	M	25.8	1.83	59.2	4.9	T7	A	112	50	1.88	4.7	2.6
Mean		**40.8**	**1.78**	**78.7**	**8.2**			**124.8**	**51.7**	**0.89**	**1.7**	**1.8**
SD		**10.3**	**0.09**	**17.0**	**5.1**			**42.3**	**9.0**	**1.05**	**1.7**	**1.9**

*Gender: M = male, F = female; AIS = ASIA Impairment Scale: A = no motor or sensory function is preserved below the neurological level, B = sensory function is preserved but motor function is not preserved below the neurological level, C = Motor function is preserved below the neurological level and more than half of key muscle functions below the single neurological level of injury have a muscle grade less than 3/5 and D = Motor Incomplete. Motor function is preserved below the neurological level and at least half of key muscle functions below the neurological levle of injury have a muscle grade > or = 3/5; ASIA = American Spinal Injury Association; WUSPI = Wheelchair User's Shoulder Pain Index.

confirm the absence of debilitating U/L musculoskeletal impairment (i.e., WUSPI [11, 12], U/L joint ranges of motion, U/L static manual muscle testing), and confirm U/L nondominance [9].

2.3. Motorized Treadmill Wheelchair Propulsion. At the start of the laboratory assessment, each participant was given a five-minute familiarization period of motorized treadmill propulsion at various slopes that differed from those investigated in the present study during which rest periods were allowed to avoid fatigue. The motorized dual belt instrumented treadmill (Bertec Corporation, Columbus, Ohio, United States) (width = 0.84 m; length = 1.84 m) was adapted for safe MWC propulsion. The MWC was anchored with elastic bands to a bilateral frictionless gliding safety system preventing excessive antero-posterior and rotational movements of the MWC (Figure 1). The imposed speed of the treadmill was adjusted for each participant to mimic the self-selected natural propulsion speed measured during a timed performance-based 20 meter MWC propulsion test. This last test was performed three times with a two-minute rest taken between trials to compute the self-selected natural propulsion speed. Thereafter, each participant first propelled their own MWC on the motorized treadmill with a level ground (0°) and then randomly at four different slopes: 2.7°, 3.6°, 4.8°, and 7.1°, reflecting an increase from one unit of height to 20, 16, 12, and 8 units of length, respectively. For each angle tested, two trials lasting a maximum of one minute (i.e., 20 consecutive pushes) and separated by a two-minute rest period were recorded. During each trial, the last 10 complete consecutive propulsion cycles recorded were used to compute the measurements of interest (i.e., trunk and shoulder kinematics, shoulder kinetics, and shoulder muscular demand) and were essential to confirm the successful completion for each slope tested (i.e., two trials/slope). This study design was selected to minimize systematic errors related to the testing (e.g., learning) and temporary maturation effects (e.g., fatigue) associated with the experimental protocol and to conclude that the findings of the present study do not result from these potential threats to internal validity. Participants rated their perceived nondominant localized U/L effort using a 10 cm visual analog scale ranging from *"no effort"* (0 cm) to *"maximum effort"* (10 cm) during the rest periods.

2.4. Trunk and Shoulder Kinematics. To capture the 3D movements of the trunk, the nondominant U/L and

(a) (b) (c)

Figure 1: Illustration of three slopes tested during motorized treadmill MWC propulsion: (a) 0° slope, (b) 3.6° slope, and (c) 7.1° slope.

the MWC, a total of 27 skin-fixed light-emitting diodes (LEDs) were placed on specific anatomical landmarks while four LEDs were fixed to the MWC frame [14]. The 3D coordinates of each LED within the laboratory coordinate system were collected at 30 Hz with a motion capture system incorporating four synchronised camera units (Optotrack 3020 and Optotrack Certus; Northern Digital Inc., Waterloo, Ontario, Canada, http://www.ndigital.com). Supplementary bony landmarks, wheelchair, and treadmill reference points were digitised to determine principal axes of segments and locate articular joint centres for the trunk and nondominant U/L, wheelchair position and treadmill slope. Before initiating the experimental tasks, three abduction-adduction and three flexion-extension active movements were recorded to locate the shoulder articular centre with respect to the scapula using a quadratic sphere fitting procedure [15, 16]. All marker trajectories were visually inspected and interpolated when coordinates were missing using a linear or a cubic spline method. The marker trajectories were then filtered using a 4th order zero-lag Butterworth filter with a cutoff frequency set at 6 Hz. The recommendations formulated by the International Society of Biomechanics [17] were used to determine segmental coordinate systems (head, trunk, arms, forearms, and hand). Relative motion between the humerus and clavicle, used as a surrogate rigid segment for the scapula that articulates with the humerus [17], was computed using a $ZX'Y''$ cardanic rotation sequence to avoid gimbal lock and to interpret reconstructed shoulder movements according to three anatomical movements commonly described in clinical practice (i.e., flexion/extension, abduction/adduction, internal and external rotations) [18]. The relative trunk forward inclination angle (i.e., forward trunk flexion) was computed as the motion of the vertical axis of the trunk, defined by a unit vector created with the midpoint between the eighth thoracic vertebra and the xiphoid process to the midpoint between the seventh cervical vertebra and the sternal notch, with respect to the vertical axis of the laboratory coordinate system. For the trunk forward flexion/extension, and the shoulder flexion/extension, adduction/adduction, and internal/external rotation, the minimal and maximal movements along with their total excursion were the main outcome measures.

2.5. Handrim Kinetics. Each participant's MWC was equipped bilaterally with 24″ or 26″ instrumented wheels (SmartWheel (SmartWheel, Out-Front (formerly Three Rivers Holdings, LLC), Mesa, Arizona, United States, http://www.out-front.com/)) to measure the three dimensional components of the total force applied at the handrim during MWC propulsion at a sampling frequency of 240 Hz [19]. While these instrumented wheels did not alter axle position or other rear wheel spatial characteristics (e.g., orientation and diameter of the handrim), they slightly increased wheelchair width and weight (4.8 kg/instrumented wheel) and may have affected rolling resistance (i.e., urethane tire). Three dimensional handrim kinetic data were filtered with a 4th order Butterworth filter and a cutoff frequency of 20 Hz and then downsampled to 30 Hz to fit the kinematic data using a custom MATLAB (MATLAB, MathWorks, Natick, Massachusetts, United States, http://www.mathworks.com/) routine.

2.6. Shoulder Kinetics. Shoulder net joint moments were computed using an inverse dynamic method [20]. The data entered into a custom-made MATLAB algorithm included

the anthropometric characteristics as well as the U/L kinematics and the pushrim kinetics with respect to the lab coordinate system. Shoulder net joint moments were then expressed in the same coordinate system used to express shoulder joint kinematics and normalized against the body mass of each participant. In fact, moderate to high associations were found between body mass and mean and peak shoulder net joint moments when propelling with no slope ($r = 0.554$ and 0.577) and with slopes of varying degrees ($r = 0.713$ to 0.809). The peak and mean shoulder net joint moments in the sagittal, frontal, and transverse plane were the main outcome measures.

2.7. Shoulder Muscular Demand. The electromyographic activity of the anterior and posterior portions of the deltoid along with the clavicular and sternal heads of the pectoralis major was recorded at the nondominant upper extremity at a sampling frequency of 1200 Hz using a portable telemetric system (NORAXON USA Inc.; Scottsdale, Arizona; Telemyo 900). Skin preparation and the placement of the surface electrodes (BlueSensor M, AMBU, Ballerup, Danmark) (Ag/AgCl sensor –13.2 mm^2 active surface area) were made in accordance with SENIAM recommendations (refer to http://www.seniam.org/). Following baseline noise removal, all EMG signals recorded were visually inspected before being filtered with a 4th order zero-lag Butterworth bandpass filter with low and high cut-off frequencies set at 30 and 500 Hz, respectively. Thereafter, EMG patterns were full-wave rectified and filtered with a 6 Hz low-pass filter to generate EMG linear envelopes for each muscle studied. The muscular utilization ratio (MUR (%)) was calculated for each muscle studied by normalizing the amplitude of the EMG signals recorded during the experimental tasks, against the peak EMG signal recorded over a 0.5 second period during one of the two static maximum voluntary contractions (MVC). Muscle-specific manual resistance was applied by a trained physiotherapist to generate the MVCs while participants remained seated in their own wheelchair. Meanwhile, another research associate manually provided additional trunk and wheelchair stability to participants. For each muscle studied, the peak and mean MURs as well as an indicator of muscle work (IMW), were calculated using the integral of the MUR data, were the main outcome measures. All EMG signal processing was performed digitally using a custom-developed MATLAB algorithm.

2.8. Statistical Analyses. Descriptive statistics (mean ± SD) were calculated for the demographic and clinical characteristics of all participants as well as for the kinematic, kinetic, and muscular demand outcome measures. For these last outcome measures, 10 propulsion cycles were averaged per trial resulting in a total of 20 propulsion cycles for each slope tested. The kinematic, kinetic and muscular demand data recorded during the push phase of each cycle analyzed were also time-normalized over 100% (i.e., 100 data points) to generate a profile for each participant and a mean group profile. Shapiro-Wilk tests confirmed that the kinematic, kinetic and electromyographic outcome measures for all slopes tested were normally distributed and justified the use of parametric statistical tests. One-way repeated-measure analyses of variance (ANOVA) with one within-subjects factor (slopes of 0°, 2.7°, 3.6°, 4.8°, and 7.1°) using a general linear model was used to determine the effect of the slopes on the kinematic and kinetic and electromyographic outcome measures and an eta-squared value was used to confirm if the proportion of the total variability attributable to the slope factor (i.e., effect size) was small (>0.02), moderate (>0.13), or large (>0.26). Whenever an ANOVA revealed significant differences (main effect; $P < 0.05$) after the result of the Mauchly's test of sphericity of the covariance matrix was taken into consideration, Student's t-tests for paired samples were computed (*post hoc* tests) with a Bonferroni correction setting the significance level at $P \leq 0.0125$ ($P \leq 0.05/4$ pairwise comparisons) as a result of the four possible slope increments (i.e., 0° to 2.7°, 2.7° to 3.6°, 3.6° to 4.8° and 4.8° to 7.1°). All statistical analyses were performed with SPSS Statistics 17.0 software for Windows. Note that the kinematic and electromyographic data were only collected and computed at the nondominant U/L since quasi-symmetrical U/L movement strategies and efforts were assumed in order to safely propel on a linear trajectory on the motorized treadmill [21] and since the nondominant U/L strength generating capability is generally weaker than the one at the dominant U/L, possibly resulting in higher relative demand at the nondominant U/L during the performance of a symmetrical functional task.

3. Results

3.1. Completion Rate. At a mean natural and constant self-selected propulsion speed of 1.17 ± 0.18 m/s [min = 0.91 m/s; max = 1.65 m/s], all participants (completion rate = 100%) were able to propel themselves on the 0° slope and up the 2.7° slope (Table 2). The completion rate reached 88.9% (N = 16/18 participants), 77.8% (N = 14/18 participants), and 55.6% (N = 10/18 participants) for the 3.6°, 4.8°, and 7.1° slopes, respectively (Table 2).

3.2. Temporal Parameters. Table 3 summarizes the mean duration of the push and recovery phases and the total duration of a propulsion cycle in seconds for the different treadmill slopes. The average durations of the push phase were similar for all tested slopes (ANOVA; $P = 0.267$), whereas the average duration of the recovery phase declined as the slope became steeper (*post hoc* tests; $P \leq 0.043$). The total duration significantly decreased as the slope became steeper (*post hoc* tests; $P \leq 0.001$), except during the 2.7° to 3.6° slope increment that remained similar.

3.3. Trunk and Shoulder Kinematics. The trunk and shoulder movement patterns are illustrated in Figures 2(a), 2(b), 2(c), and 2(d), whereas the minimum, maximum, and excursion of the trunk and shoulder movement amplitudes are summarized in Table 4. The slopes of the treadmill significantly influenced most minimum, maximum, and excursion trunk and shoulder movement amplitudes. At the trunk, all

Table 2: Description of self-selected comfortable propulsion speed, experimental tasks completed, and rate of perceived exertion.

Participants	Self-selected speed (m/s)	Slopes					Rate of perceived exertion (/10)				
		0°	2.7°	3.6°	4.8°	7.1°	0°	2.7°	3.6°	4.8°	7.1°
1	1.11	√	√	√	√	√	0.9	7.4	5.5	8.1	9.5
2	1.65	√	√	√	√		0.1	2.2	0.3	0.3	—
3	1.05	√	√	√	√	√	1.3	1.3	3.9	4.4	7.1
4	1.18	√	√	√	√	√	0.8	1.7	3.7	4.8	7.8
5	1.20	√	√	√	√	√	1.7	4.8	2.5	5.9	7.9
6	0.91	√	√	√			3.0	6.3	7.7	—	—
7	1.48	√	√				0.9	9.7	—	—	—
8	1.16	√	√	√			0	0.3	0.9	—	—
9	1.04	√	√	√	√		1.8	3.8	3.3	4.2	—
10	1.27	√	√	√	√	√	0	0	0	0	0
11	1.39	√	√	√	√	√	1.0	1.5	3.3	2.7	3.2
12	0.99	√	√	√	√		1.1	4.1	4.0	7.5	—
13	1.06	√	√				4.2	8.8	—	—	—
14	1.25	√	√	√	√	√	1.2	5.2	6.7	6.4	8.5
15	1.03	√	√	√	√		0.0	0.6	1.9	3.6	—
16	1.07	√	√	√	√	√	0.7	3.4	4.6	6.6	9.2
17	1.06	√	√	√	√	√	0.7	6.2	2.6	6.5	8.9
18	1.11	√	√	√	√	√	2.9	4.6	6.1	6.6	7.3
Mean	**1.17**	n = **18**	n = **18**	n = **16**	n = **14**	n = **10**	**1.2**	**3.9**	**3.5**	**4.7**	**6.8**
SD	**0.18**						**1.1**	**2.9**	**2.2**	**2.5**	**3.0**

Table 3: Group average (SD) mean temporal parameter measured of the push phase measured at the nondominant handrim at the five different slopes tested.

	Slopes				
	0°	2.7°	3.6°	4.8°	7.1°
	Temporal parameters (s)				
Push phase	0.48 ± 0.08	0.48 ± 0.08	0.49 ± 0.08	0.48 ± 0.07	0.48 ± 0.06
Recovery phase	0.59 ± 0.22	0.27 ± 0.10	0.26 ± 0.09	0.22 ± 0.08	0.18 ± 0.05
Total (cycle)	1.07 ± 0.23	0.75 ± 0.16	0.75 ± 0.14	0.70 ± 0.13	0.66 ± 0.10

minimum, maximum, and excursion movement amplitudes significantly increased as the slope became steeper, except for minimum and maximum values during the 2.7° to 3.6° slope increment that remained similar. The greatest maximum forward trunk flexion (60.9°), which was accompanied by the greatest forward trunk excursion (22.4°), was reached during the 7.1° slope. At the shoulder, the maximum shoulder flexion movement amplitude significantly increased as the slope became steeper, except for the 3.6° to 4.8° slope increment that remained similar, whereas the minimum shoulder flexion movement amplitude (i.e., shoulder extension) significantly decreased during that same period. As a result, the shoulder flexion excursion remained comparable despite slope increments. The minimum, maximum, and excursion shoulder abduction movement amplitudes remained comparable as the slope became steeper. The minimum, maximum and excursion shoulder internal rotation movement amplitudes also remained comparable as the slope became steeper, with the exception of the minimum and excursion values, which significantly increased during the 0° to 2.7° slope increment.

3.4. Shoulder Kinetics. The net shoulder flexion, abduction and internal rotation moment patterns are illustrated in Figures 2(e), 2(f), and 2(g) while their mean and maximum values are summarized in Table 5. The greatest maximum shoulder moments were found during flexion across all slopes except for the 7.1° slope when the internal rotation generated the greatest moment. All mean and maximum shoulder moments were significantly influenced by the slopes of the treadmill. The mean and maximum flexion moments significantly improved as the slope increased, except for the 3.6° to 4.8° and 4.8° to 7.1° slope increments. The mean adduction moments only significantly improved as the slope increased between 0° and 2.7°, whereas the peak mean value only significantly improved as the slope increased between

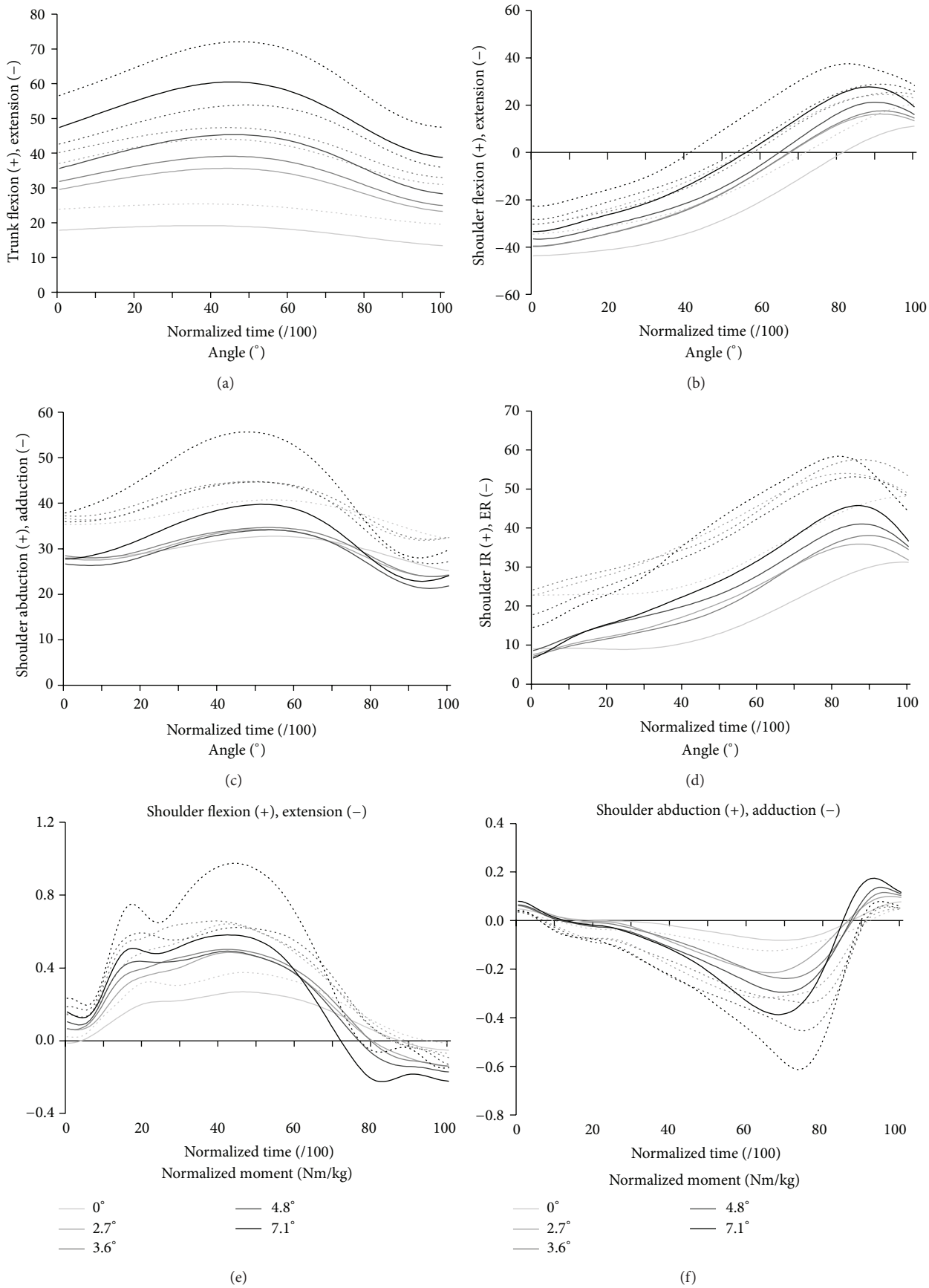

Figure 2: Continued.

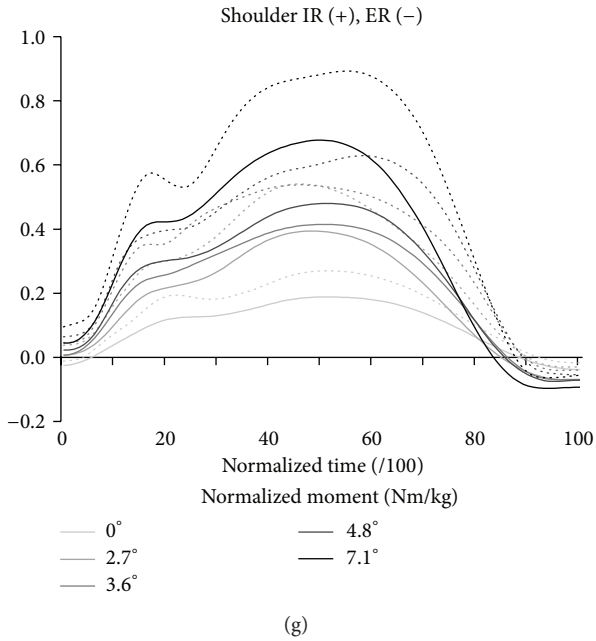

Figure 2: Group averaged time-normalized profile (solid line) and standard deviation (dotted line) of the shoulder and trunk kinematics (a, b, c, and d) and weight-normalised shoulder moments (e, f, and g) during the push phase for the five slopes tested at self-selected natural speed.

the 0° to 2.7°, 3.6° to 4.8°, and 4.8° to 7.1° slope increments. The mean and maximum internal rotation moments significantly increased as the slope became steeper, except for the 3.6° to 4.8° slope increment.

3.5. Shoulder Muscular Demand. The MUR patterns of the muscles studied are illustrated in Figure 3, while their mean and maximum values are summarized in Table 6. The mean IMWs of the muscles studied are summarized in Table 6 and illustrated in Figure 4. For all muscles studied, their mean and maximum MURs, as well as their indicator of muscle work value, significantly increased as the slope became steeper, except for the posterior deltoid that remained comparable between the 2.7° to 3.6° slope increment.

4. Discussion

This study quantified trunk and nondominant shoulder kinematic changes along with nondominant shoulder joint moments and electromyographic changes during MWC propulsion on a motorized treadmill set at a self-selected natural speed on five different slopes (i.e., 0°, 2.7°, 3.6°, 4.8°, and 7.1°). Overall, the MWC users with a SCI increased forward trunk flexion and peak shoulder flexion while also increasing shoulder mechanical and muscular efforts to adapt to slopes that progressively increased during simulated uphill MWC propulsion on a motorized treadmill.

4.1. Trunk and Shoulder Movement-Related Adaptations. The movement-related adaptations occurring at the trunk and shoulder partially support the hypothesis that their outcome measures would gradually and significantly increase with each slope increment while the speed remained constant. At the trunk, the maximum forward trunk flexion and the total trunk excursion increased significantly as the slope became steeper, except for the 2.7° to 3.6° slope increment. Chow et al. [5] obtained somewhat comparable results in terms of trunk kinematics although no difference was revealed when comparing 4° and 8° slopes. This may be explained by the fact that participants propelled at self-selected speeds that decreased progressively in their protocol as the slope increased. Nonetheless, the increased maximum forward trunk flexion coupled with the increased forward trunk excursion may allow MWC users to move their centre of mass further and faster anteriorly and to maintain its projection in front of the rear wheel axle in order to prevent backward tilt and falls as the slope increases. This may also explain why maximum forward trunk flexion and forward trunk excursion became greater as the gravitational effects became harder to overcome with steeper slopes. The decreasing success rate with a steeper slope may be explained in part by the fact that some participants were classified as being overweight (body mass index > 25) or class I obese (body mass index = 30.0–34.9) with associated abdominal obesity that limited their ability to increase forward trunk flexion to accommodate for the steeper slopes. In fact, 55.6% of participants who were unable to propel up the 7.1° slope were overweight ($N = 3$) or obese ($N = 2$), whereas only 30% of participants who were able to propel up the 7.1° slope were overweight ($N = 2$) or obese ($N = 1$). Hence, abdominal circumference may deserve additional attention when investigating wheelchair propulsion technique or manual wheelchair skills such as uphill propulsion.

Table 4: Group average (SD), total excursion (°), and maximum and minimum (°) kinematic values measured at the trunk and nondominant shoulder at the five different slopes tested as well as the results of the ANOVA and post hoc comparison test.

		Slopes					ANOVA	Eta-squared	Post hoc comparisons			
		0°	2.7°	3.6°	4.8°	71°			0° versus 2.7°	2.7° versus 3.6°	3.6° versus 4.8°	4.8° versus 7.1°
Joint angle (°)		Trunk—flexion (+), extension (−)							Trunk—flexion (+), extension (−)			
	Excursion	5.88 (2.12)	12.67 (3.40)	14.56 (2.77)	17.61 (3.96)	22.37 (6.00)	<0.001	0.907	<0.001	0.006	<0.001	0.003
	Maximum	19.27 (6.23)	35.92 (8.45)	39.45 (8.31)	45.79 (8.53)	60.93 (11.30)	<0.001	0.889	<0.001	0.031	0.002	0.005
	Minimum	13.39 (6.14)	23.26 (7.75)	24.90 (8.05)	28.18 (7.71)	38.57 (8.59)	<0.001	0.910	<0.001	0.028	<0.001	<0.001
		Shoulder—flexion (+), extension (−)							Shoulder—flexion (+), extension (−)			
	Excursion	55.67 (9.20)	58.17 (10.92)	59.63 (9.01)	60.19 (6.98)	64.99 (8.95)	0.019	0.296	0.072	0.183	0.473	0.217
	Maximum	11.31 (9.34)	18.00 (8.20)	19.35 (7.62)	22.59 (7.30)	30.88 (7.45)	<0.001	0.849	<0.001	0.001	0.054	<0.001
	Minimum	−44.36 (9.35)	−40.17 (9.36)	−40.27 (9.15)	−37.60 (8.34)	−34.10 (10.36)	<0.001	0.651	<0.001	0.006	<0.001	<0.001
		Shoulder—abduction (+), adduction (−)							Shoulder—abduction (+), adduction (−)			
	Excursion	9.35 (3.43)	12.27 (7.16)	13.48 (6.61)	14.88 (7.56)	19.65 (11.88)	0.012	0.337	0.010	0.363	0.158	0.047
	Maximum	33.25 (8.00)	34.81 (10.65)	35.40 (10.26)	34.88 (10.57)	40.60 (15.43)	0.083	—	—	—	—	—
	Minimum	23.90 (6.27)	22.53 (7.26)	21.92 (6.66)	20.00 (5.45)	20.94 (5.70)	0.038	—	—	—	—	—
		Shoulder—internal rotation (+), external rotation (−)							Shoulder—internal rotation (+), external rotation (−)			
	Excursion	24.86 (8.20)	30.55 (10.24)	33.35 (8.83)	34.52 (9.44)	43.28 (11.34)	<0.001	0.623	0.004	0.040	0.765	0.013
	Maximum	31.57 (16.82)	37.89 (18.46)	40.15 (19.17)	43.02 (12.23)	49.69 (8.71)	<0.001	0.665	<0.001	0.029	0.855	0.033
	Minimum	6.71 (13.57)	7.34 (15.10)	6.80 (16.94)	8.50 (9.18)	6.41 (7.70)	0.152	—	—	—	—	—

Table 5: Group average (SD) mean and peak weight-normalised moments (Nm/kg) measured at the nondominant shoulder at the five different slopes as well as results of the ANOVA and post hoc comparison tests.

		Slopes					ANOVA	Eta-squared	Post hoc comparisons			
		0°	2.7°	3.6°	4.8°	71°			0° versus 2.7°	2.7° versus 3.6°	3.6° versus 4.8°	4.8° versus 7.1°
Normalised moments (Nm/kg)		Shoulder—flexion (+), extension (−)							Shoulder—flexion (+), extension (−)			
	Mean	0.140 (0.056)	0.226 (0.067)	0.248 (0.081)	0.233 (0.083)	0.242 (0.151)	0.027	0.288	<0.001	0.003	0.433	0.966
	Peak (maximum)	0.292 (0.109)	0.512 (0.152)	0.546 (0.167)	0.559 (0.128)	0.733 (0.302)	<0.001	0.588	<0.001	0.004	0.867	0.049
		Shoulder—abduction (+), adduction (−)							Shoulder—adduction (+), abduction (−)			
	Mean	−0.016 (0.023)	−0.060 (0.050)	−0.069 (0.046)	−0.090 (0.043)	−0.103 (0.036)	<0.001	0.623	<0.001	0.414	0.147	0.366
	Peak (minimum)	−0.088 (0.041)	−0.257 (0.105)	−0.278 (0.104)	−0.369 (0.137)	−0.494 (0.133)	<0.001	0.797	<0.001	0.149	0.002	0.002
		Shoulder—internal rotation (+), external rotation (−)							Shoulder—internal rotation (+), external rotation (−)			
	Mean	0.097 (0.036)	0.189 (0.050)	0.222 (0.064)	0.252 (0.065)	0.344 (0.101)	<0.001	0.726	<0.001	0.004	0.191	0.010
	Peak (maximum)	0.196 (0.078)	0.415 (0.144)	0.456 (0.129)	0.530 (0.148)	0.780 (0.184)	<0.001	0.801	<0.001	0.001	0.103	0.001

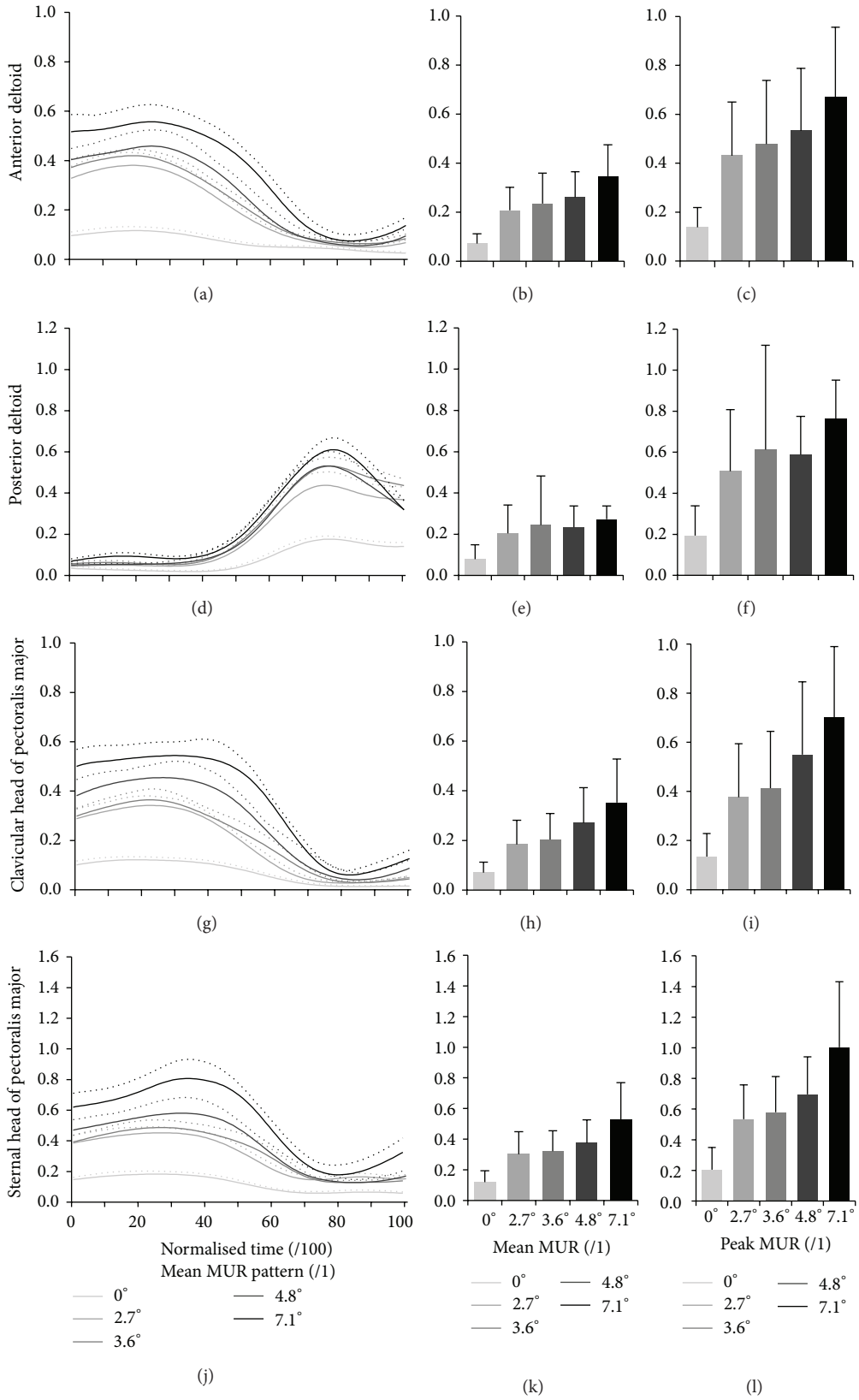

Figure 3: Group averaged time-normalized mean profile (solid line) and standard deviation (dotted line) (a, d, g, and j), as well as group-average (SD) mean (b, e, h, and k) and peak (c, f, i, and l) muscle utilization ratio (MUR) during the push phase for the five slopes tested at self-selected natural speed.

Table 6: Group average (1 SD), mean muscle utilisation ratio (MUR), peak MUR (/1), and indicator of muscle work (IMW) values for each muscle assessed at the nondominant shoulder at five different slope angles as well as results of the ANOVA and post hoc comparison tests.

	Slopes					ANOVA	Eta-squared	Post hoc comparisons			
	0°	2.7°	3.6°	4.8°	71°			0° versus 2.7°	2.7° versus 3.6°	3.6° versus 4.8°	4.8° versus 7.1°
	Anterior deltoid							Anterior deltoid			
Mean MUR (/1)	0.07 (0.04)	0.21 (0.09)	0.23 (0.12)	0.26 (0.10)	0.34 (0.13)	<0.001	0.831	<0.001	0.002	<0.001	0.001
Peak MUR (/1)	0.14 (0.08)	0.44 (0.23)	0.48 (0.26)	0.54 (0.26)	0.68 (0.28)	<0.001	0.836	<0.001	0.002	<0.001	<0.001
IMW	7.18 (3.98)	20.70 (9.43)	23.48 (12.37)	26.06 (10.43)	34.48 (13.04)	<0.001	0.829	<0.001	0.003	<0.001	0.002
	Posterior deltoid							Posterior deltoid			
Mean MUR (/1)	0.08 (0.07)	0.20 (0.14)	0.24 (0.24)	0.23 (0.10)	0.27 (0.06)	<0.001	0.907	<0.001	0.144	<0.001	<0.001
Peak MUR (/1)	0.20 (0.14)	0.52 (0.31)	0.62 (0.51)	0.60 (0.19)	0.77 (0.19)	<0.001	0.894	<0.001	0.027	0.001	0.005
IMW	7.95 (6.92)	20.36 (13.84)	24.43 (23.81)	23.35 (10.47)	27.34 (6.48)	<0.001	0.906	<0.001	0.143	<0.001	<0.001
	Clavicular fibers of pectoralis major							Clavicular fibers of pectoralis major			
Mean MUR (/1)	0.07 (0.04)	0.18 (0.10)	0.20 (0.10)	0.27 (0.14)	0.35 (0.18)	<0.001	0.754	<0.001	0.003	<0.001	0.005
Peak MUR (/1)	0.14 (0.09)	0.38 (0.22)	0.42 (0.23)	0.57 (0.31)	0.71 (0.29)	<0.001	0.814	<0.001	0.009	<0.001	0.001
IMW	7.00 (4.26)	18.37 (9.77)	20.41 (10.46)	27.04 (14.32)	35.28 (17.57)	<0.001	0.752	<0.001	0.003	<0.001	0.005
	Sternal fibers of pectoralis major							Sternal fibers of pectoralis major			
Mean MUR (/1)	0.12 (0.08)	0.30 (0.15)	0.32 (0.13)	0.37 (0.15)	0.53 (0.24)	<0.001	0.781	<0.001	0.005	0.001	0.006
Peak MUR (/1)	0.21 (0.15)	0.54 (0.23)	0.59 (0.23)	0.71 (0.26)	1.01 (0.42)	<0.001	0.829	<0.001	0.003	<0.001	0.002
IMW	11.94 (7.53)	30.23 (14.77)	32.27 (13.23)	37.53 (15.11)	52.80 (24.06)	<0.001	0.780	<0.001	0.005	0.001	0.006

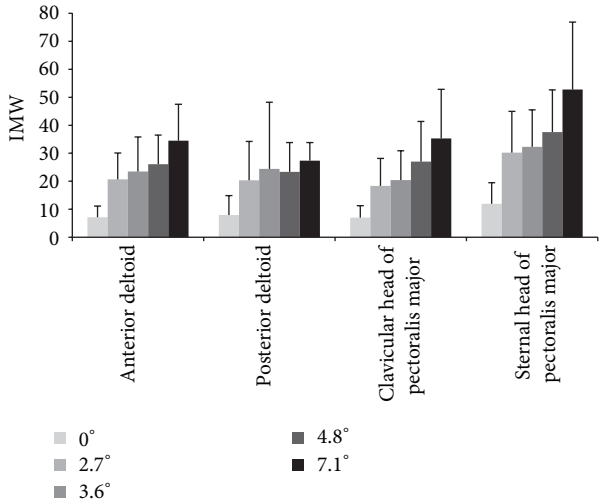

Figure 4: Group averaged (SD) mean indicator of muscle work (IMW) during the push phase for the five slopes tested at self-selected natural speed.

At the shoulder, the maximum shoulder flexion movement amplitude significantly increased as the slope became steeper, except for the 3.6° to 4.8° slope increment that remained similar, whereas the minimum shoulder flexion movement amplitude (i.e., shoulder extension) significantly decreased during that same period. As a result, the shoulder flexion excursion remained comparable despite slope increments. It is possible that increased maximum shoulder flexion was needed to accommodate for the increased forward trunk flexion and to apply most of the force tangentially on the handrim to preserve mechanical efficiency. The fact that only the duration of the recovery period drastically decreased as the slope became steeper can also explain, in part, the relatively stable shoulder flexion-extension excursion (i.e., similar push phase durations across slopes). This finding is consistent to some extent with the work of Yang et al. [3] who found comparable values of movement-related adaptation at the shoulder when comparing level overground and uphill (i.e., 3° slope) MWC propulsions. However, the shoulder flexion-extension excursion significantly increased on average by 9.37° and the shoulder extension remained similar in their study contrary to the results of the present study. These discrepancies may be attributed to the fact that participants used test wheelchairs that were not anchored to a safety system when propelling up a 3° slope at 0.9 m/s on a motorized treadmill, the level surface propulsion (0° slope) was performed overground, and the shoulder kinematic calculation differed from the work by Yang et al. [3]. Nevertheless, all these results confirmed kinematic adaptations at the shoulder as the slope progressed. Lastly, the increased forward trunk flexion coupled with an increased shoulder flexion that modifies the orientation of the force generated at the shoulder may increase posterior shoulder joint forces and explain the elevated muscular demand at the posterior deltoid occurring towards the end of the push phase [22].

4.2. Shoulder Joint Mechanical and Muscular Effort Adaptations. The shoulder joint moments adaptations partially support the hypothesis that their outcome measures would gradually and significantly increase with each slope increment while the speed remained constant. For two out of the three shoulder net joint moments investigated, most of the mean and peak values for the shoulder flexor and internal rotator moments progressively increased as the slope became steeper, aside from some outcome measures during the 2.7° to 3.6° or the 3.6° to 4.8° slope increment. The relative muscular demand adaptations and muscular work computed for all muscles investigated fully support this hypothesis since their outcome measures gradually and significantly increased with each slope increment, while the speed remained constant. Hence, once the results of these two approaches are combined, it is clear that the shoulder mechanical and muscular effort increases are key contributors to the adaptation process associated with the steeper slopes. Moreover, these results are in line with the perceived U/L efforts expressed by the participants and may explain, in part. The decreasing success rate with a steeper slope as shoulder strength generating capability most likely becomes a determinant for propelling on steeper slopes.

4.3. Implications for Clinical Practice. With the use of a slope or of a combination of slopes, when a MWC user propels himself/herself on a motorized treadmill, therapists may be able to offer task-specific high-intensity short duration interval training programs to increase U/L strength, particularly at the shoulders. Cautiousness is advised with this practice since the risk exposure (i.e., increased shoulder mechanical and muscular demands) will progressively and significantly increase as the slope becomes steeper and will vary according to the strength-generating capability of each MWC user. In addition, since the risk exposure could trigger the development or exacerbation of secondary impairments at the U/L, particularly at the shoulders, such a program should also be accompanied by proper warm-up and cool-down periods as well as by antagonist muscle strengthening to prevent muscle strength imbalance. Alternatively, therapists may also offer task-specific cardiorespiratory fitness training programs when a MWC user propels on a motorized treadmill with no slope or minimal slope (<2.7°) with minimal demands during a prolonged period of time (i.e., cardiorespiratory fitness training).

4.4. Limits of the Study. The present study included a relatively small sample size ($n = 18$) of experienced MWC users who have completed their rehabilitation process which may limit the strength of the evidence and potential to generalize the results with new MWCs, respectively. The fact that the participants used their personal wheelchairs during the study warrants consideration as optimal wheelchair positioning and configuration parameters most likely differs across participants and impacts the outcome measures of interest in the present study and the risk of the MWC tilting or falling backwards when propelling up a slope. The use of the instrumented wheels which slightly increase the width and weight

of the wheelchair along with rolling resistance may have modified participants' propulsion technique (e.g., increased shoulder abduction) and fatigue level (e.g., increased U/L effort) in comparison to propelling with their own wheels. The self-selected natural treadmill speed determined for each participant, maintained across all slopes tested in an effort to isolate the effect of speed, also requires consideration since MWC users tend to reduce their speed when propelling uphill in daily life, particularly on steep slopes [2]. Finally, the kinematic and kinetic and electromyographic variables solely focused on the trunk and nondominant shoulder prevent a full understanding of U/L adaptations (i.e., elbow and wrist not studied) during motorized treadmill MWC propulsion across different slopes.

5. Conclusion

This study confirms that MWC users with a SCI increase forward trunk flexion and peak shoulder flexion while also increasing shoulder mechanical and muscular efforts to adapt to slopes that progressively increase during simulated uphill MWC propulsion on a motorized treadmill. Few similarities were found between the 2.7° to 3.6° and the 3.6° to 4.8° slope increments for shoulder flexion and adduction moments. Future studies incorporating interactions with various slopes and velocities could strengthen the results of the present study and provide additional evidence-based knowledge on wheelchair propulsion on a motorized treadmill.

Abbreviations

AIS:	American Spinal Injury Association Impairment Scale
EMG:	Electromyography
ER:	External rotation
LED:	Light-emitting diode
IMW:	Indicator of muscle work
IR:	Internal rotation
JCS:	Joint coordinate system
MUR:	Muscle utilization ratio
MVC:	Maximum voluntary contraction
MWC:	Manual wheelchair
U/L:	Upper limb
WUSPI:	Wheelchair User's Shoulder Pain Index.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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Research Article

A Participatory Approach to Develop the Power Mobility Screening Tool and the Power Mobility Clinical Driving Assessment Tool

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The electric powered wheelchair (EPW) is an indispensable assistive device that increases participation among individuals with disabilities. However, due to lack of standardized assessment tools, developing evidence based training protocols for EPW users to improve driving skills has been a challenge. In this study, we adopt the principles of participatory research and employ qualitative methods to develop the Power Mobility Screening Tool (PMST) and Power Mobility Clinical Driving Assessment (PMCDA). Qualitative data from professional experts and expert EPW users who participated in a focus group and a discussion forum were used to establish content validity of the PMCDA and the PMST. These tools collectively could assess a user's current level of bodily function and their current EPW driving capacity. Further multicenter studies are necessary to evaluate the psychometric properties of these tests and develop EPW driving training protocols based on these assessment tools.

1. Introduction

Independent mobility is one of the most important determinants of quality of life for individuals with disabilities [1, 2]. Electric powered wheelchairs (EPWs) are key assistive technology devices that increase mobility and comfort while promoting social integration among users, thereby improving their overall quality of life [3, 4]. Over the past few decades, the number of EPW users has been increasing [5, 6] due to advances in health care, increase in the aging population of baby boomers, and the number of veterans returning from conflict situations [7–9]. However, this increase in EPW usage has also been associated with an increase in wheelchair related accidents, injuries [6, 10], and equipment abandonment [11, 12]. To compound this critical issue, it has also been shown that over forty percent of those who receive EPWs continue to have problems with certain EPW driving skills [13]. There also exists a growing cohort of people who desire and deserve EPWs for mobility but who have not been able to acquire a device because of severe impairments in motor, sensory, or cognitive functions that have precluded them from passing a clinical assessment or because of inadequate resources to allow them to practice driving [14–16].

Specific impairments in body structures and function, especially cognition and sensory perception, have been linked to problems with driving EPWs. Cullen and colleagues linked self-perceived functional performance of driving an EPW with verbal recall, visual construction ability, and global cognition [17]. Mendoza et al. reported anecdotal evidence that accidents among EPW users in a nursing home increased if the user had executive dysfunction [18]. Routhier et al. suggested that a variety of psychological factors influence EPW use: cognitive function, motivation, analytical capacity,

and problem-solving [15]; and Batavia et al. noted that cognitive impairment affects EPW driving in users with traumatic brain injury [19]. Massengale et al. demonstrated that visual perception, visual function (specifically those with ocular motor function, stereo depth, field of vision, binocular vision, and far visual acuity), and cognition have a significant impact on EPW driving performance [20].

Currently, there are a few assessment tools available to rehabilitation professionals for evaluating EPW driving capacity, both in the user's natural environment and in the clinic. The Power Mobility Indoor Driving Assessment (PIDA) and the Power Mobility Community Driving Assessment (PCDA), developed by Dawson et al. and Letts et al., respectively, are the two most commonly used clinical tools that help identify general areas or tasks for which more training is needed (e.g., "parking under a table") or where modifications to the EPW or the environment are necessary [14, 20, 21]. Following the PIDA and PCDA, Kirby et al. published the Wheelchair Skills Test (WST) that has mainly been used to evaluate manual wheelchair mobility [22] but more recently has been adapted for the evaluation of EPW driving skills as well [23]. Massengale et al. developed the Power Mobility Road Test (PMRT) to study the effect of visual function, visual perception, personality traits, and cognitive skills that affect EPW use [20]. The PMRT consists of a set of 12 structured tasks and 4 unstructured tasks with moving obstacles. The performance scores from these tasks were then correlated with outcomes from evaluations of visual function, visual perception, cognition, and personality traits to study their influence on EPW driving skills.

Apart from these widely used clinical tools, two other tools have been developed for research settings. The Functional Evaluation Rating Scale (FERS) developed by Hasdai et al. uses a scoring system similar to the PIDA and has been adopted by other researchers to assess driving performance in simulation programs [24, 25]. In a three-step process, Routheir et al. established a framework for wheelchair driving assessment [15], developed theObstacle Course Assessment of Wheelchair User Performance (OCWA UP) [26], and established reliability of the assessment tool [27].

However, even with advancements in technology to measure driving skill, there still are no standardized tools to screen potential adult EPW drivers for specific impairments in motor, sensory, or cognitive function [15–17, 20, 21]. Such a tool, coupled with a tool to measure driving skill, could lead to the development of clinical training protocols with the ultimate goal of improving independence and safety of potential drivers [16].

To better understand the concepts of function, activity, and capacity as they relate to EPW driving, Mortenson and colleagues evaluated wheelchair related outcome measures on the basis of the World Health Organization's (WHO) international classification of functioning, disability, and health (ICF) [28–31]. Mortenson et al. concluded that all of the currently available EPW driving assessment tools have been focused on evaluating wheeled mobility *capacity* or *performance* or both, to assess *activity and participation* of a wheelchair user [30]. However, none of the tools assess how the driver's physical and cognitive functions affect driving; in other words, it is also crucial to assess specific impairments in *body structures and function*.

There are multiple ways to approach this problem. Adopting previously established techniques used in adaptive vehicle driving is one such approach [32–36]. Driving rehabilitation specialists employ a series of tests that help identify impairments in major functional domains (motor or sensory or cognitive) that affect driving ability [33, 37–40]. If a driver has an impairment in one of the functional domains, targeted training programs in that specific domain and teaching compensatory mechanisms to overcome that impairment have been shown to improve driving performance [41–44]. Such an evidence based approach toward driving assessment and training has led to the development of sound clinical practice guidelines which have been effective in training and counseling drivers, such as the elderly [45].

Secondly, learning strategies and techniques that have been employed in training children with cognitive disabilities could provide valuable insight into the development of newer assessment and training tools for potential adult EPW users. Tefft et al. reported that problem solving and spatial relations had a direct impact on the variance of EPW driving skills among children [46]. Following Tefft et al., Furumasu et al. published the pediatric powered wheelchair skills test, which used a five-point scale to quantify a child's driving ability [47]. In 2011, Nilsson et al. reported several strategies that have been applied to teach EPW driving skills for children with cognitive disabilities [48, 49]. These studies have demonstrated the strong association of cognitive, sensory, and motor assessments with better training strategies in children [48].

In this study, we adopted the principles used in adaptive vehicle driving to screen for impairments and pooled them with the neuropsychiatric measures that have been used to measure capacity for EPW driving skills among adults, along with strategies and principles that have worked well with children to develop two new tools. Specifically, we employed participatory research [50–52] and qualitative methods [53, 54] to develop the Power Mobility Screening Tool (PMST) comprising a list of simple tests to quantify motor, sensory, and cognitive functioning ("pre-road" screening tool) and the Power Mobility Clinical Driving Assessment (PMCDA) ("on-road" test) to assess EPW driving capacity. The specific aim of this study was to establish content validity of both the PMST and the PMCDA.

2. Materials and Methods

2.1. Participants. This study was reviewed and approved by the University of Pittsburgh's Institutional Review Board. A convenience sample of 21 participants in the United States were approached by word of mouth or phone calls or via email to participate in the surveys and focus group phase of the study. The inclusion criteria were being a professional expert (physician, occupational therapist, physical therapist, or a rehabilitation engineer) in the field of assistive technology with at least five years of professional experience with the wheelchair delivery process or an expert EPW user who has been using an EPW for a minimum of three years, and the

participant is between the ages of 18 and 80 years. There were no exclusion criteria.

A brief abstract explaining the purpose of the focus group and objective of the discussion forum was given to all the attendees as a part of the registration package of the 29th International Seating Symposium held in Nashville, TN, in 2013. Any attendee of the symposium who was interested in partaking was invited to participate in the discussion forum. There were no specific inclusion or exclusion criteria.

2.2. Research Protocol

2.2.1. Surveys. Two separate surveys were sent to the professional experts and the EPW expert users via email: the Tools and Tasks survey (Appendix A) and the Users' survey (Appendix B). The purpose of these surveys was to generate a list of items that could be included in the PMST and the PMCDA, and rank these items based on the level of importance. The professional experts were asked to complete the Tools and Tasks survey, while the expert EPW users were asked to complete the User's survey.

The Tools and Tasks survey consisted of two sections. Section one was a list of tests commonly used to evaluate motor, sensory, and cognitive function in adaptive vehicle driving [33, 36, 40, 55], and section two consisted of a list of driver tasks pooled from existing EPW driving assessment tools [14, 20–22, 26, 56]. Participants were asked to rank each of the screening tests in order of importance within the motor, sensory, and cognitive sections. They were instructed to use ranks ranging from 1 (most important) to 3 (least important) (Appendix A). Similarly, for section two, a rank of 1 (most important) to 5 (least important) was requested (Appendix A). A rank of "0" was given if the test or task should not be included. The participants were also given an option to add more tests or tasks.

The User's survey consisted of two questions. Question one asked the participants to list the top 5 skills that are important for a person to be a highly skilled driver in both indoor and outdoor environments, and question two asked them to list the top 5 skills that are important for a person to be a moderately skilled driver who drives only indoors. The users were also asked to rank these tasks in the order of their importance within each question.

The surveys were sent to the participants two weeks before the scheduled date of the focus group and a follow-up reminder email alert to return all the surveys was sent one week before the focus group.

2.2.2. Focus Group. After all the surveys were returned, a teleconference was set up for the focus group. Two researchers acted as moderators, and the entire focus group was audio recorded. The moderators presented the overall median rankings of the items and initiated a discussion using a structured set of questions [53, 54, 57] (Table 1). Following the focus group, the recording was transcribed and analyzed for common themes by each of the moderators individually. Then, the two moderators had a discussion to reach a consensus about predominant themes. Based on these themes and comments raised during the focus group, the first iteration of the PMST and PMCDA was established.

Table 1: Questions for the focus group.

Screening tool
What sections should it contain?
What tests should be included under each section and how many?
Can the tests be used in people with high-level motor impairment?
How should this tool be scored?
How long will it take to complete?
What supplies are needed?
Assessment tool
What sections should the tool have?
What tasks should be in each section?
How should it be scored?
How should each task be defined or delineated?
How long will it take to complete?
What supplies are needed?

2.2.3. Discussion Forum. Data from the surveys and the focus group were analyzed to develop the first iteration of the PMST and the PMCDA. Three months following the focus group, one of the moderators presented the first iteration of both tools in the discussion forum during the International Seating Symposium. A brief introduction of currently existing EPW driving tools was presented followed by the first iteration of the PMST and the PMCDA. The PMST and the PMCDA were further discussed, based on the structured set of questions listed in Table 1. Based on the comments put forth by the participants during the discussion forum, the second iteration of the PMST and PMCDA was developed (Tables 9 and 10).

3. Results

3.1. Surveys. Twenty-one experts were approached and invited to take the surveys, of which eight professional experts consented to participate. Of the ten expert EPW users approached, three consented to participate. All the eleven experts returned the surveys within two weeks (response rate of 100%). The mean duration of clinical experience of the professional experts was 13.75 (±6.9) years, and all of them had assistive technology professional (ATP) certifications. Tables 2 and 3 demonstrate the professional backgrounds of the eight experts who took the surveys and the 46 experts who participated in the discussion forum. Table 4 shows the demographic profile of the expert EPW users. It is important to note that the only expert EPW user, who participated in the discussion forum, was also a rehabilitation scientist.

Table 5 shows successive iterations of the list of tests to be included in the PMST. Tables 6 and 7 show successive iterations of the list of tasks to be included in the PMCDA according to experts. Table 8 shows the ranked list of tasks

Table 2: Professional background of the professional experts who took the surveys and participated in the focus group.

Professional background	Number of participants	Mean years of experience (years ± SD)	Min (years)	Max (years)
Physical therapists	4	13.75 (9.39)	5	26
Occupational therapists	3	14.66 (5.68)	10	21
Rehabilitation scientist	1	11	—	—
Total	8	13.75 (6.94)	5	26

Table 3: Demographics of the experts who participated in the discussion forum.

(a) Professional background

Professional background	Number of participants
Physical therapist	20 (44%)
Occupational therapists	10 (22%)
AT supplier	14 (31%)
Others (rehabilitation technicians and rehabilitation engineers)	2 (3%)
Total	46 (100%)

(b) Years of experience

Years of experience	Number of participants
0–2 years	1 (2%)
3–5 years	0
6–10 years	12 (26%)
More than 10 years	12 (26%)
More than 20 years	13 (28%)
More than 30 years	7 (16%)
More than 40 years	1 (2%)

suggested by the users. Although the users were asked to list the top 5 skills, all the users listed more. During the focus group, one of the users pointed out, "These tasks are essential for an individual who receives his or her first EPW."

3.2. Focus Group. All eight professional experts who took the surveys participated in the focus group. These experts defined essential criteria for the PMST and the PMCDA. The first criterion was that the tests should be easy to administer for therapists with any level of training (novice versus experienced) and with any professional background (occupation therapist versus physical therapist). As one of the physical therapists pointed out, "All physical therapists may not be trained to administer complex cognitive assessments... besides performing the mini mental status. So, we have to be clear that under my certification I can administer whatever test we choose to include, if this has to be a globally useful tool." Secondly, the tests should be inexpensive and should not require the purchase of any supplies that are not commonly available in clinical settings. The same therapist also noted, "I do not have access to an accessible bathroom all the time. So, if we define a task like approaching or parking by a sink, I might not be able to administer it to all my clients... all the time." Third, the scoring system should be clearly defined without any room for subjectivity. Experts agreed that a common problem with currently available tests is that the scoring systems are too complicated or subjective. One of the therapists indicated, "Either the 1 to 4 or 0 to 100 might provide a good system, but if it's not clearly defined, then the room for subjectivity is where it gets challenging." These criteria led to the common consensus that the list of five screening tests (Table 5) would be sufficient to assess a user's functional capacity to drive an EPW. Similarly, the list of ten indoor tasks (Table 6) and ten outdoor tasks (Table 7) should not only be sufficient to assess users' safety and EPW driving capacity but would also help therapists identify clearly what area would require more user training.

Analysis of the transcripts of the focus group led to the identification of important thematic concepts for the tools. The group suggested that separate sections are essential for assessing driving capacity in the indoor and outdoor environments, as driving under these two circumstances has different skill sets. Hence, it was recommended that the PMCDA be designed to have two sections with tasks ordered by increasing level of complexity. The group agreed that the number of tasks in the PMCDA is sufficient to assess the baseline driving capacity and safety of the EPW user. Further, they felt most testers would require few supplies to conduct testing with either of the tools. The experts ranked eight tasks as "0," indicating these tasks could be excluded from the assessment tool. However, during the development of the first iteration of the PMCDA, two of these tasks from the indoor section (drives backward or reverse 10 ft in a straight line and turns 90° while moving backward) and two of these tasks from the outdoor section (ascends 10° incline and descends 10° incline) were added to the list, since the users had ranked these skills highly and had indicated that these are essential skills necessary for a new EPW user.

Several themes emerged regarding the scoring system. The experts felt that the possible total scores on both tools must have a wide range in order to stratify drivers with variable functional capability and driving capacity. For example, a dichotomous pass or fail system should be avoided, since this system might not provide the sufficient variations in scores to include drivers with all skill levels. The possible total scores of 5–15 on the PMST and 23–69 on the PMCDA were felt to be sufficient for stratification. They also felt strongly that the scoring used for individual tasks should be clear and mutually exclusive and suggested that a score of "0" on the tools should be avoided. Based on these concerns, a three-point scoring system was proposed for individual tasks within both the PMST and PMCDA (Appendix C).

Table 4: Medical diagnosis and years of EPW usage of the users who participated in the study.

	Medical diagnosis	Number of years of EPW usage
Surveys and focus group	Cerebral palsy	17
	SCI	5
	Connective tissue disorders with multiple orthopedic abnormalities	4.5
Discussion forum	Cerebral palsy	21

Table 5: Successive iterations of the screening tests.

Ranked screening tests (from the survey)		List of screening tests in the first iteration of the PMCDA (after the focus group)	Specific changes suggested during the discussion forum
Professional experts' ranks	Screening tests		
1	(i) Range of motion of the head, neck, and trunk [61, 62] (ii) Others: (a) Knowledge of cause and effect (b) Motor planning/problem solving ability, for example, maneuvering out of a tight spot (c) Basic cognition: orientation to person, place, or situation		
1.5	(i) Confrontation testing [63, 64] (ii) Snellen's chart (for far vision) [65] (iii) Random Dot Stereoacuity test [66] (iv) Others: (a) Strength of the body part that will be controlling chair (b) Ability to use control interface, for example, switch, joystick, and so forth	Motor (i) Driver can functionally control an interface (joystick, head control, etc.) with appropriate body part to drive the chair (ii) Driver controls chair with sufficient endurance (ability to tolerate sitting and operating the control interface) during the screening Sensory (i) Driver can **visually identify** an object (e.g., therapy ball) 2 meters away with clinic in background, in left, center, and right visual fields Cognitive (i) Driver displays ability to understand cause and effect (action on the control interface will move the chair) (ii) Driver has ability to focus, concentrate, attend to task, and shift focus within the task	Sensory (i) Driver can **identify** an object (e.g., therapy ball) 2 meters away with clinic in background, in left, center, and right visual fields Cognitive (i) Driver has ability to focus, concentrate, attend to task, and shift focus within the task **during the entire period of assessment**
2	(i) Range of motion of the upper limbs [67, 68] (ii) Near vision acuity charts [69] (iii) Proteus maze [70] (iv) Continuous performance test [71] (v) Others: (a) Minimental exam		
2.5	(i) The NSUCO/Maples oculomotor test [72] (ii) Motor-free visual perception test (MVPT) [73] (iii) Digit span [74, 75]		
3	(i) Motor coordination [76, 77] (ii) Trail making A & B [78] (iii) Others: (a) Functional vision—visual scanning, visual conflict (b) Endurance with use of trial equipment with driving obstacles (c) Reliability of use of the control interface (nonfatigable, consistent)		

3.3. Discussion Forum. Among the 1300 attendees of ISS, 46 therapists, durable medical equipment suppliers, and rehabilitation technicians and one wheelchair user with cerebral palsy participated in the discussion forum. The discussion forum followed the same protocol as the focus group. Several salient issues were identified following the analysis of the transcription from the audio-recorded discussion forum. Overall, the group confirmed that all the tasks listed in the PMCDA are essential for the assessment of EPW driving capacity. In addition to the tasks listed in the first iteration,

Table 6: List of indoor driver tasks.

Ranked indoor driver tasks from the survey		List of indoor tasks in the first iteration of the PMCDA after the focus group	Specific changes suggested during the discussion forum
Professional experts' ranks	Indoor driver tasks		
1	(i) Drives forward (15 ft) (in a straight line) in narrow corridor without hitting walls (ii) Avoids one person coming towards participant in hallway	(i) Drives forward (15 ft) (in a straight line) in 36″ hallway (ii) Drives backward 10 ft in a straight line in 36″ hallway (iii) Passes through 36″ doorway (iv) Avoids therapy balls approaching from left and right (v) Turns 90° while moving forward (vi) Turns 90° while moving backward (vii) Turns 180° in place to the left (viii) Can safely maneuver in-between 2 chairs spaced 32 in apart (ix) Approaches an accessible sink (x) Negotiates over a 1 in door threshold or mock threshold (piece of wood)	These tasks were added to the list: (i) Turns 90° and enters a doorway (ii) Approaches a transfer surface (bed or chair) (iii) Stops on command (emergency stop)
2	(i) Turns 90° while moving forward (ii) Passes through doorways without hitting walls (36″doorways)		
3	Turns 180° in place—left		
4	Can safely maneuver in-between objects and tight spaces		
5	Approaches furniture without bumping into them		
0	(i) Drives backward (or reverse) 10 ft, in a straight line (ii) Turns 90° while moving backward (iii) Avoids "wet floor" sign (within a 5 ft wide corridor) (iv) Parking under table (v) Parking beside table		

Table 7: List of outdoor driver tasks.

Ranked outdoor driver tasks from the survey		List of outdoor tasks in the first iteration of the PMCDA after the focus group	Specific changes suggested during the discussion forum
Professional experts' ranks	Outdoor driver tasks		
1	(i) Avoids moving obstacles approaching from both sides—left and right (ii) Drives forward 30 ft in 30 s	(i) Drives forward 30 ft in 30 s (ii) Drives over an unpaved surface (iii) Ascends 5° incline (iv) Descends 5° incline (v) Ascends 10° incline (vi) Descends 10° incline (vii) Crosses a street (viii) Rolls 10 ft across 5° side-slope (ix) Ascends an ADA[1] curb cut (x) Descends an ADA[1] curb cut	No additional tasks suggested
2	Ascends 5° incline		
3	Descends 5° incline		
4	(i) Crossing street without lights (ii) Rolls 10 ft across 5° side-slope		
0	(i) Ascends 10° incline (ii) Descends 10° incline (iii) Is able to drive over 15 cm pot-hole		

[1] American Disabilities Act.

three other tasks were suggested for the indoor section of the PMCDA. First, "turning 90° and entering a doorway" was added since the group suggested this task is essential for safe driving and is a frequent occurrence in the user's natural environment. Secondly, "stopping the chair on command" was added, since participants proposed that this task was not only an assessment of the user's EPW driving capacity, but also a gauge of the EPW users' ability to respond to dynamic changes in their environment, which in turn reflects their ability to use the EPW safely. Lastly, "parking an EPW parallel to a transfer surface," which could either be a bed or a chair, was also added, since participants felt this is a vital task that every EPW user will have to perform at some point in time irrespective of his or her medical need for using an EPW. It was suggested that this last task could be performed during the mat assessment typically performed during a routine examination for an EPW [58].

One task that was discussed extensively during the discussion forum was the ability of an EPW user "to get on and off an elevator." Users also indicated this as one of the tasks that should be performed by an EPW user with moderate skill. However, this task was not added to the list for two reasons.

Table 8: Essential EPW driving tasks suggested by expert EPW users.

Indoor skills
(1) Carrying out skills in reverse direction (doorways, navigating around objects)
(a) Driving backwards in various environments
(2) Navigating around objects (couches, chairs, tables)
(a) Knowing where people are
(b) Knowing where tables and chairs are
(c) Navigating in crowded environments
(3) Negotiating tight doorways
(a) Navigating in narrow hallways and doorways (just wide enough for chair)
(4) Turning around in tight places (elevators)
(5) Parking next to transfer stations (bed, toilet)
(6) Speed control
(7) Paying attention to corners
(8) Paying attention to areas with stairs
(9) Opening and entering a door with an auto-closer
(10) Navigating over wet tile (hydroplaning)
Outdoor skills
(1) Looking everywhere before moving
(2) Always staying to one side of the hallway or sidewalk
(3) Combinations of skills (starting and stopping on ramps, etc.)
(a) Driving up and down steep grades
(b) Boarding public buses
(4) Turning around on cross slopes
(a) Driving straight on cross slopes
(5) Navigating ramps
(6) Driving on rough terrain (broken sidewalks, gravel, brick)
(a) Paying attention to bumps and edges
(b) Maintaining desired driving route on uneven ground such as cobblestone, brick, and gravel
(c) Climbing over obstacles (uneven curb cuts, sidewalks, small curbs)
(7) Starting and stopping quickly
(a) Speed control
(b) Ability to stop fast
Indoor skills for moderate drivers
(1) Navigating around objects
(2) Negotiating doorways
(a) Driving in ADA accessible hallways and doorways
(b) Opening, passing through, and closing a door
(c) Entering and exiting elevator and turning around inside as necessary
(3) Parking next to a transfer area (bed, toilet, mat table)
(a) Pulling under or up to side of table/counter
(b) Parking in desired space for transfer
(4) Turning around in open space
(5) Object avoidance
(a) Speed control
(b) Paying attention to corners
(c) Paying attention to areas with stairs
(d) Knowing where people are
(e) Knowing where tables and chairs are

Table 9: The power mobility screening tool (PMST).

Motor	
Driver can functionally control an interface (joystick, head control, etc.) with appropriate body part to drive the chair	1–3
Driver controls chair with sufficient endurance (ability to tolerate sitting and operating the interface)	1–3
Sensory	
Driver can identify an object (e.g., therapy ball) 2 meters away with clinic in background, in left, center, and right visual fields	1–3
Cognitive	
Driver displays ability to understand cause and effect (action on the control interface will move the chair)	1–3
Driver has ability to focus, concentrate, attend to task, and shift focus within the task during screening	1–3
Total	5–15

See Appendix C.1 for instructions on using the PMST.

Table 10: The power mobility clinical driving assessment tool (PMCDA).

	Score
Indoor	
Drives forward (15 ft) (in a straight line) in 36″ hallway	1–3
Drives backward 10 ft in a straight line in 36″ hallway	1–3
Passes through 36″ doorway	1–3
Avoids therapy balls approaching from left and right	1–3
Turns 90° while moving forward	1–3
Turns 90° and enters a doorway	1–3
Turns 90° while moving backward	1–3
Turns 180° in place to the left	1–3
Can safely maneuver in-between 2 chairs 32 in apart	1–3
Approaches an accessible sink	1–3
Approaches a transfer surface (bed or chair)	1–3
Negotiates over 1 in door/mock threshold (piece of wood)	1–3
Stops on command (emergency stop)	1–3
Outdoor	
Drives forward 30 ft in 30 s	1–3
Drives over an unpaved surface	1–3
Ascends 5° incline	1–3
Descends 5° incline	1–3
Ascends 10° incline	1–3
Descends 10° incline	1–3
Crosses a street	1–3
Rolls 10 ft across 5° side-slope	1–3
Ascends an ADA curb cut	1–3
Descends an ADA curb cut	1–3
Total	23–69

See Appendix C.2 for instructions on using the PMCDA.

First, it may not be feasible to administer the task in all clinics. And secondly, experts felt that two other tasks included in the list, namely, "can safely maneuver in-between 2 chairs spaced 32 inches apart" and "turns 180° in place to the left/right," assess basic skills also necessary for elevator use. However, the group agreed that, during training, the trainer should make this an essential task to be practiced and discussed with the driver.

The group noted that objective measures to quantify users' functional impairments were essential, and performing functional clinical tests rather than lengthy neuropsychiatric or motor measures was advisable and acceptable. However, the group suggested two changes in the PMST. First, under the sensory section, the group pointed out that the term "visually" may not be applicable to all EPW users and will have to be changed to accommodate individuals with all levels of sensory functioning. Second, under the cognitive assessment, it was suggested that estimated length of time should be changed to "the entire period of assessment." Finally, a change in the scoring system was also suggested.

The criterion definition of scoring level 2 should be changed to include all kinds of cues (visual, verbal, and auditory) that encompass users with any functional capabilities. Based on these comments and suggestions, the second iteration of the PMCDA and the PMST was developed (Appendix C).

4. Discussion

Over the years, there has been a growing need for testing and developing wheelchair-specific outcome measures that allow clinicians to justify their equipment recommendations and to show the effectiveness of specific interventions to the scientific community [30]. By adopting the principles of ICF, such outcome measures can delineate and define methods to assess body structures and function, tasks important for activities and participation [30], and the capacity and performance of the user [28, 30, 59]. Design of outcome measures for EPW driving should also follow few key principles that have been recognized as salient in the scientific literature. First, the goal of the assessment should be explicitly targeted towards enhancing mobility and independence of the user rather than preventing access to EPWs for potentially unsafe drivers [21]. In other words, the measure should be used with the goal of assessing safety and identifying areas where training can help a potential driver improve skills, rather than simply determining whether he or she is capable of driving at one point in time. Second, the measure should be scored in such a way that it can demonstrate progress with training [15, 26, 27]. Third, the measure should be able to identify key areas where training could improve skill [20], not only by identifying what tasks are difficult for a driver but also by identifying what body structures and functions are contributing to those specific difficult tasks. The experts and users in this study reinforced these principles, and the participatory approach that was adopted accommodated all three principles when developing the tools. The iterative approach, with inclusion of over 50 professional experts and expert EPW users, established good content validity for the PMST and the PMCDA.

Although we adopted concepts from adaptive vehicle driving literature to develop the content of the PMST, the tool that emerged is uniquely suited for EPW driving. Experts identified several concerns in administering many of the standardized neuropsychological tests commonly used for vehicle driving in a wheelchair clinic. First, the qualifications and training necessary to administer these tests might preclude use by many potential testers. Second, each of the tests would require the clinic to purchase a test kit, and if multiple tests were to be administered together, it would result in an expensive assessment process. Third, the process would become quite lengthy, which decreases the likelihood of a tester offering these tests in a busy wheelchair clinic. Most importantly, they excluded many of the tests because they did not feel that the tools were sensitive or specific enough to measure functional capability in terms of EPW driving ability. However, the experts did agree that objective measures are necessary in each of the three domains (motor, sensory, and cognitive) to quantify a user's functional impairments in these domains. Hence, rather than using standardized neuropsychological tests, the experts proposed the use of functional clinical tests (Table 4) for the purpose of screening. If any major clinical concerns would be identified during screening, then the experts recommended use of the PMST as a basis for referral for further testing by a specialist such as a neuropsychologist, an audiologist, or an ophthalmologist.

The content of the PMCDA includes similar driving tasks as those identified by a focus group conducted by Torkia et al. [60]. In that study, researchers identified four specific wheelchair mobility tasks/maneuvers that were difficult for EPW users, namely, controlling the EPW's joystick, avoiding obstacles, maneuvering backwards, and going through narrow doorways. In addition, this study also reported that, during outdoor mobility, EPW users face difficulty in four major areas: using streets and sidewalks, navigating through crowds, using adapted modes of transportation, and dealing with rain or snow conditions. Although our tool does not include measures of transportation or inclement weather for practical reasons, it is worthwhile to note the striking similarities in the other tasks identified in their study.

There are significant advantages to using the PMST and the PMCDA in combination as a tool kit to assess EPW driving capacity over the currently existing tools. Currently, there are no other validated methods of *screening* for cognitive, motor, and sensory issues related to EPW driving. This is the first time a screening tool with functional tasks has been validated dually with an assessment for EPW driving in adults. A validated screening tool for cognitive, motor, and sensory capacity may help to standardize the evaluation process if adopted across centers. This, in turn, could lead to the development of training interventions customized for each type of impairment that affects various driving skills. Development of such customized interventions is important because individuals with cognitive or sensory impairments may need extra training and should not be excluded from opportunities to learn to drive based on a sole screening or assessment. Rather, this combination of tools can help to identify areas that would need more training to make the user a better EPW driver. Another advantage to using the PMST and the PMCDA is that they are pure measures of driving capacity; that is, they include only tasks that are exclusively related to EPW driving, not other factors like wheelchair maintenance. Lastly, as reported by one of the participants during the discussion forum, because they have a clearly defined scoring system, the PMST/PMCDA combination tool kit is quick and easy to administer without ambiguity among scoring levels.

4.1. Study Limitations. Because the experts were identified through a convenience sample of colleagues and acquaintances in the field of assistive technology, they may have been following similar clinical practices as the investigative team, which may have made it easier to reach consensus on content validity. However, participants were recruited from several locations across the country and inclusion of a large number of participants from the discussion forum who were voluntarily attending the session increased the diversity of the input. Still, the tool was developed solely using input from American

and Canadian experts and is not validated for other cultures or languages. Offering the survey only via email limited the external validity because not all EPW users necessarily have computers. However, using email also provided the ability for some users to participate who might not otherwise been able to participate due to transportation barriers. The large number of participants in the discussion forum could have hindered some participants from expressing their views. However, we allowed ample time for individual questions and comments after the discussion forum ended, which provided the moderator with an opportunity to incorporate individual questions and concerns in the iterative revision of tools. In addition, both the professional experts and the expert users were included in one group for the focus group and the discussion forum. One benefit of having this structure was that participants were able to hear opinions that may be quite different from their own. On the other hand, diversity within a focus group can sometimes cause the group to stray from the topic or have trouble honing in on specific ideas. However, the latter was not a problem in this study as the group was closely moderated and sufficient content was produced to be useful for tool development. Finally, this study included only four expert EPW users, in comparison to the fifty professional experts in the study. However, tasks pooled from the past literature combined with the Users' survey were helpful in identifying key tasks for the PMCDA, which have also been identified by users in another focus group study [60].

4.2. Future Directions. Further testing is needed to evaluate feasibility of administering this tool in a busy clinical environment, and further psychometric testing is needed to establish inter- and intrarater reliability. The development of a strong clinical tool is an iterative process and, hence, future work will include a wider range of users with varying degrees of sensory or cognitive disabilities. In the future, we will develop a normative dataset that displays the functional capabilities of a wide range of EPW users, which would then help us develop training protocols targeted towards specific motor, sensory, or cognitive impairments that affect driving skill which are identified by the PMST. We intend to work with expert EPW users to identify strategies and techniques that could help teach newer users and marginally skilled users to drive better, based on their functional ability. Finally, we wish to study the effectiveness of such a training intervention on driving outcomes.

5. Conclusion

The scientific literature is sparse in measurements that can quantify a spectrum of driving skills among adult EPW users, and prior to this study, no screening tool for motor, sensory, and cognitive impairments that could impact EPW driving existed. This study used a participatory approach to establish content validity of a new screening tool for these impairments and an assessment tool to quantify EPW driving performance. Further work is necessary to establish the feasibility and reliability of these assessment instruments and to build and test training protocols for EPW driving. Multisite testing in large populations of EPW users is needed.

Appendices

A. The Tools and Tasks Survey

A.1. Screening Tools. Please choose 3 items in each section that are the most important components of a screening for power mobility. Rank them in order from 1 (most important) to 3 (least important) and leave the rest blank. You may choose a combination of existing items or write in your own, but please list only a total of 3 items in each section.

A.1.1. Motor: Choose 3 Here

(1) Range of motion of the upper limbs

(2) Range of motion of the head, neck, and trunk

(3) Motor coordination

(a) Purdue pegboard: measures two types of activities: gross movements of hands, fingers, and arms and "fingertip" dexterity in an assembly task. Involves sequential insertion of pegs and assembly of pegs, collars, and washers

(b) Grooved peg board/fine motor speed-manipulative dexterity test using holes with randomly positioned slots and pegs, which have a key along one side. Pegs must be rotated to match the hole before they can be inserted.

(4) Others: . . .

A.1.2. Sensory: Choose 3 Here

(1) Visual

(a) *Ocular movement*: the *NSUCO/Maples oculomotor test* is a standardized method of scoring standard eye movement testing.

(b) *Visual field* (*by confrontation testing*)

(c) *Visual acuity*

(i) *Snellen's chart* (for far vision)

(ii) *Near vision acuity charts*: charts that can assess vision within 1 m distance.

(d) *Depth perception (stereopsis)*: *Random Dot Stereoacuity test:* designed to rapidly test for amblyopia and strabismus in early and nonreaders and nonverbal children and adults.

(e) *Color vision*: *color vision testing made easy*: intended use is for screening color vision of young children beginning at age 3 and individuals with developmental delays.

(f) *Visual perception*

(i) *Motor-free visual perception test (MVPT-3)*: assesses an individual's visual perceptual ability—with no motor involvement needed to make a response

(ii) *Developmental test of visual perception adolescent and adult (DTVP-A)*: a comprehensive measure of visual perception that reliably differentiates visual-perceptual problems from visual-motor integration deficits.

(g) Others: . . .

(2) Auditory

(a) *Calibrated finger rub auditory screening test (CALFRAST)*: confrontational testing using fingers to make audible sound

(b) *Portable audiometer*

(c) Others: . . .

A.1.3. Cognitive: Choose 3 Here

(1) Cognition and memory skills

(a) *Trail making A & B*: specifically assesses working memory, visual processing, visuospatial skills, selective and divided attention, and psychomotor coordination.

(b) *Clock drawing test*: assesses a patient's long-term memory, short-term memory, visual perception, visuospatial skills, selective attention, abstract thinking, and executive skills. Preliminary research indicates an association between specific scoring elements of the clock drawing test and poor driving performance.

(2) Porteus maze: set of paper forms on which the subject is required to trace a path; tests problem solving

(3) Digit span (WSIR): tests speed of information processing, longest list of letters or numbers that a person can repeat back in correct order

(4) Continuous performance test by Connors (CPT): tests visual attention; task-oriented computerized assessment of attention disorders. Clients are presented with a repetitive, "boring," task and must maintain their focus

(5) Others: . . .

A.2. Driving Tasks. Please choose 5 items in each section that are the most important components of a driving skills assessment. Rank them in order from 1 (most important) to 5 (least important) and leave the rest blank. You may choose a combination of existing items or write in your own, but please list only a total of 5 items in each section.

Indoor

(1) Drives forward (15 ft) (in a straight line) in narrow corridor without hitting walls

(2) Drives backward (or reverse) 10 ft, in a straight line

(3) Turns 90° while moving forward

(a) Left

(b) Right

(4) Turns 90° while moving backward

(a) Left

(b) Right

(5) Turns 180° in place

(a) Left

(b) Right

(6) Passes through doorways without hitting walls (36″ doorways)

(7) Avoids "wet floor" sign (within a 5 ft wide corridor)

(8) Avoids one person coming towards participant in hallway

(9) Can safely maneuver in-between objects and tight spaces

(a) Drive between a couch and coffee table, in a living room setup

(b) Can enter an elevator

(c) Adjust within an elevator

(d) Exit the elevator

(10) Approaches furniture without bumping into them

(a) Parking under table

(b) Parking beside table

Others:

(11) . . .

(12) . . .

(13) . . .

(14) . . .

(15) . . .

Outdoor

(1) Drives forward 30 ft in 30 s

(2) Crossing street without lights

(3) Avoids moving obstacles approaching from both sides: left and right

(a) Avoids two or more moving obstacles (person coming towards participant) in sidewalk

(b) Avoid an unexpected ball

(4) Ascends 5° incline

(5) Descends 5° incline

(6) Ascends 10° incline

(7) Descends 10° incline

(8) Rolls 10 ft across 5° side-slope

(a) Left
(b) Right

(9) Is able to drive over 15 cm pot-hole

Others:

(10) ...
(11) ...
(12) ...
(13) ...
(14) ...

B. Users' Survey

Years of Experience Driving a Power Chair ...

Main Disability ...

Question 1. List the top 5 skills that you think are important for a person to be *a highly skilled driver in both indoor and outdoor environments*. Please rank them in order of importance from most important to least important.

Indoor Skills

(1)
(2)
(3)
(4)
(5)

Outdoor Skills

(1)
(2)
(3)
(4)
(5)

Question 2. List the top 5 skills that you think are important for a person to be *a moderately skilled driver who drives only indoors*. Please rank them in order of importance from most important to least important.

Indoor Skills

(1)
(2)
(3)
(4)
(5)

Please include any additional comments below.

C. Instructions for Using PMST & PMCDA

C.1. The Power Mobility Screening Tool (PMST). See Table 9.

Instructions

(i) Ask client to drive the EPW in an open space free from obstacles.
(ii) You may provide visual or auditory clues along with verbal instructions to complete tasks.
(iii) Tasks can be completed in any order and also as part of a routine physical examination or mat assessment.
(iv) Client may identify objects by any means (verbally, gestures, etc.) and may use visual aids.
(v) Control interface settings should be adjusted for safety and at discretion of the trainer and driver.

Scoring System for the Screening Tool

Score of 1: if the driver requires physical assistance, lacks the skill, or cannot complete the task, a score of 1 is given.

Score of 2: if the driver requires verbal or auditory hints or cues but no physical assistance or has partial skill (e.g., can identify an object in 2 of 3 visual fields or can partially move a joystick) a score of 2 is given.

Score of 3: if the driver completes the task without help or has adequate skill, then a score of 3 is given, even if additional time is needed for the task.

C.2. The Power Mobility Clinical Driving Assessment Tool (PMCDA). See Table 10.

Instructions

(i) You may provide visual or auditory clues along with verbal instructions to complete tasks.
(ii) Tasks can be completed in any order.
(iii) Control interface settings should be adjusted for safety and at discretion of the trainer and driver.

Scoring System for the Driving Assessment Tool

Score of 1: if the driver requires physical assistance or cannot complete the task, a score of 1 is given.

Score of 2: if the driver requires verbal or auditory hints or cues but no physical assistance, a score of 2 is given.

Score of 3: if the driver completes the task without help, then a score of 3 is given.

Abbreviations

EPW: Electric powered wheelchair
ICF: International Classification of Functioning, Disability, and Health
WHO: World Health Organization
PMST: Powered Mobility Screening Tool

PMCDA: Powered Mobility Clinical Driving Assessment Tool
PIDA: Power Mobility Indoor Driving Assessment
PCDA: Power Mobility Community Driving Assessment
WST: Wheelchair Skills Test
PMRT: Power Mobility Road Test
FERS: Functional Evaluation Rating Scale
OCAW UP: Obstacle Course Assessment of Wheelchair User Performance
ADA: American Disabilities Act.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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Research Article

Development of a Wheelchair Skills Home Program for Older Adults Using a Participatory Action Design Approach

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Restricted mobility is the most common impairment among older adults and a manual wheelchair is often prescribed to address these limitations. However, limited access to rehabilitation services results in older adults typically receiving little or no mobility training when they receive a wheelchair. As an alternative and novel approach, we developed a therapist-monitored wheelchair skills home training program delivered via a computer tablet. To optimize efficacy and adherence, principles of self-efficacy and adult learning theory were foundational in the program design. A participatory action design approach was used to engage older adult wheelchair users, care providers, and prescribing clinicians in an iterative design and development process. A series of prototypes were fabricated and revised, based on feedback from eight stakeholder focus groups, until a final version was ready for evaluation in a clinical trial. Stakeholder contributions affirmed and enhanced the foundational theoretical principles and provided validation of the final product for the target population.

1. Introduction

Canada has a rapidly aging population [1] and it is estimated that, over the next 50 years, the proportion of Canadians over 65 years of age will double to more than 25% [2]. With advancing age, the risk of a disabling health condition increases and personal mobility is the most prevalent area of impairment among older adults [3]. A manual wheelchair (MWC) is often prescribed to address compromised mobility and function. In 2001, an estimated 81,000 Canadian older adults were wheelchair users [4]. The intent of providing assistive devices such as a MWC is to improve functional independence and diminish caregiver burden [5]. In practice, however, elder MWC users experience substantial restrictions in the performance of their activities of daily living [3] particularly in comparison with elders who do not use a mobility device [4, 6]. As a result, nearly 60% of older Canadian MWC users are dependent upon formal or informal care providers for basic mobility [4].

The international classification of functioning, disability, and health (ICF) provides a conceptual framework for describing factors that contribute to or impede participation [7], including participation among wheelchair users. For the individual, compromised body function and structure related to pain, strength, and endurance may be influential. Relevant contextual factors may be environmental. These include the built environment, such as pseudo-accessible [8] and inaccessible locations, and challenging terrain; the social environment, such as social attitudes and level of personal assistance; and assistive technology devices including wheelchairs that are low-quality and inappropriate or do not fit the user. Contextual factors may also be personal, such as age and confidence (self-efficacy) with wheelchair use. At the activity level, wheelchair mobility skill and proficiency have also been identified as significant contributors to participation [9–12]. Wheelchair skills have been amenable to improvement through training, particularly when delivered in a structured format. For example, there is considerable

evidence supporting substantive benefits of the Wheelchair Skills Training Program [13] in a variety of populations and contexts [14–16].

Older adults typically receive little training when they obtain a wheelchair [17, 18], and whatever training they do receive tends to focus on functions related to hospital discharge (e.g., transferring from the wheelchair to bed or toilet). Insufficient training occurs for a variety of reasons, but primarily because therapists have limited time and must focus on pragmatic issues, and resources for follow-up services are restricted [19]. Coming in as an outpatient for multiple training sessions is costly—many visits, transportation, and inconvenience—and it does not provide training in the context of use (i.e., real-life obstacles). Providing training in the community would be desirable, but there are not resources to enable therapists to make multiple visits and provide training at home or in the community. This lack of comprehensive, context-appropriate training is particularly problematic because older adults require more time and training to acquire new motor skills due to age-related changes in motor, sensory, and cognitive function [20, 21].

To address this problem, we set out to develop a monitored home program for wheelchair skills training. Delivering rehabilitation training as a monitored or self-managed home program among older adults has been effective for a wide variety of outcomes including strengthening [22], physical activity [23], self-care [24, 25], and exercise [26, 27]. Home programs are advantageous because they allow privacy for the user, occur in a familiar and real-life context, can be conveniently integrated into the users schedule, and do not require the time, effort, and expense of travel to another location [23]. This approach would be economically viable, allow time for more practice, and facilitate training in the context of using authentic obstacles.

For rehabilitation home programs targeting motor skills in older adults, maximizing training frequency and practice in the natural context of use are essential elements. A 2010 Cochrane Review identified several factors related to adherence in exercise interventions. Programs that incorporated social cognitive theory (i.e., self-efficacy strategies), were clinician-monitored, and increasingly graded the complexity of training activities were more successful improving participants' adherence, frequency and duration of exercise [28]. Education is central to rehabilitation and training should utilize strategies from Andragogy (adult learning) [29, 30] as active ingredients to promote program adherence and successful skill acquisition with older adults [31].

With advances in affordability, size, portability, accessibility, and user-interface simplicity, computer-related devices are becoming increasingly useful for rehabilitation interventions. Computer and popular gaming systems (e.g., Nintendo Wii) have shown promising results in rehabilitation by casting therapy in a more engaging context. More recently, their use in rehabilitation among older adults has also been explored. For example, Aarhus et al. [32] provided physical activity training in a Danish nursing home using a commercial gaming system with a participation rate over 90% and found improvements in physical function, increased motivation and tolerance for activity, and trends towards improvement in fitness. Imam et al. [33] demonstrated improved mobility outcomes among individuals with lower limb amputation (age range: 45–59 years) using the Nintendo Wii to augment usual therapy and reported high rates of adherence and enjoyment with the program. Creating an interesting interface, such as the use of games, is positively associated with older adults' intention to use computer-related technologies [32].

Purpose. The purpose of this project was to develop a prototype Wheelchair Skills Training Program that could be delivered as a home program using a computer tablet, entitled Enhancing Participation in the Community by Improving Wheelchair Skills or EPIC Wheels. This involved the development of specific program content as well as a system of delivery. The intent was for content to include evidence-based skills relevant to novice older adult MWC users, incorporate Andragogical educational strategies, promote self-efficacy, and be delivered in an engaging and accessible format. The specific objectives were to

(1) engage older adult MWC users in the research and development process,

(2) incorporate stakeholder input through the design and evaluation phase,

(3) produce a prototype intervention program for proof of concept.

2. Materials and Methods

There is an emerging consensus in the field of assistive technology that consumer involvement *during* the process of intervention development is crucial [34–36]. This is particularly true with older adults to ensure that a technology "solution" itself does not induce more problems than it resolves [36]. An additional benefit of involving older adults is the "Design for All" tenet that assistive technology interventions that work well for the elderly are also likely to work better for consumers generally [36].

We employed participatory action design (PAD), which is an approach to innovation development that places high value in the on-going involvement of intended users during design and evaluation elements [37–39]. Using a PAD framework, stakeholder evaluation and feedback were incorporated into the development stages of program content and delivery through the use of focus groups (see Figure 1). Focus groups were used to capitalize on participant interaction to elicit needs and preferences, personal experiences, and exploratory solutions "outside of the box" [36] and have proved effective in other comparable participatory rehabilitation intervention studies [35–37, 40]. Including a qualitative approach ensured that learning strategies were relevant for older adults, practice activities were age-appropriate and achievable, potential for user motivation and adherence was maximized, and the product design considered the technological accessibility needs of an aging population. The study followed an iterative pattern where issues of importance are identified, prototypes

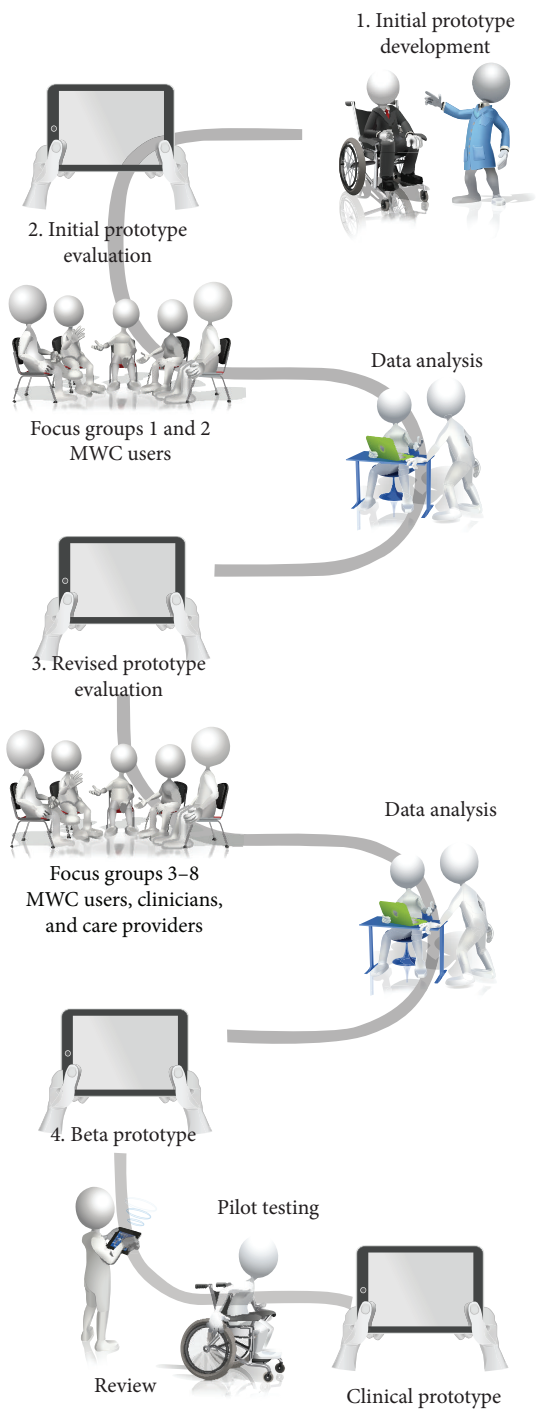


Figure 1: Phased study design.

are developed and refined, and the results are evaluated for utility (see Figure 1).

A total of eight focus groups were conducted in two cities: Winnipeg and Vancouver. These locations provided diversity in culture, weather, geography, and degree of wheelchair accessibility and would also serve as research sites for a subsequent clinical trial of the program. Finally, two older adult MWC users (one experienced user and one novice user) pilot tested the prototype using a research protocol intended for the subsequent clinical trial. Details of the Android software development have been published separately [41]. All participants provided consent and approval from the university affiliated Research Ethics Board at each site was obtained prior to conducting this study.

2.1. Participants. Three stakeholder groups in each city were included: experienced MWC users aged 60 and over, care providers of older adult MWC users, and clinicians who prescribed wheelchairs and/or provided wheelchair training for older adults. MWC users were the primary stakeholder group as we were most interested in their perception of the program content and delivery, since adherence to a home program is critical to effectiveness [28]. The user groups ($n = 10$) each participated in two focus groups (at different points in the program development), while care provider and clinician groups each attended one focus group; separate focus groups were conducted with each stakeholder group in each city. While the target population for the training program is *novice* users, we chose to use *experienced* users for several reasons. First, we anticipated that their availability and potential for attendance would be greater since they would have either acquired mobility skills or developed compensatory strategies over time. It was also more likely that whatever impairment precipitated their acquisition of a wheelchair would have stabilized sufficiently that they would be able to schedule and attend two focus groups. Second, novice users often experience a transitional period of emotional and social adjustment, and engagement in a research study might prove challenging [42, 43]. We reasoned that novice users would have a more limited experience and perspective to know what it was they *did not know* and the full scope of environmental situations that posed the greatest barriers to mobility and participation. Conversely, experienced users, while somewhat distanced from the "experience" of early adjustment to wheelchair use, would have a more comprehensive understanding of the scope of environmental barriers and could reflect on which barriers were most problematic and which mobility skills had been most important or influential in addressing participation restriction.

The MWC user and care provider participants were recruited using email and postal invitations, public advertisement, and word-of-mouth. MWC users were at least 60 years of age, were living in the community, had used a MWC as their primary means of mobility for at least one year, and have sufficient cognition and English language skills to engage in the focus group process. Care providers were individuals (e.g., spouse, relative, or caregiver) who assisted or accompanied a MWC user at least 60 years of age while using their wheelchair inside and outside of the home. For the clinician group, occupational and physical therapists at the largest rehabilitation hospital in each city who supervise or provide clinical services (e.g., prescribe a wheelchair or provide wheelchair mobility training) to individuals 60 years of age or older were invited to a lunch-hour focus group. Advertising posters and brochures were distributed to therapists at each site and local rehabilitation

managers distributed invitations to their staff via email. All participants provided informed consent prior to participating in this study.

A total of 10 MWC users participated in the focus groups. At the Vancouver site ($n = 6$), one individual was not able to attend the second focus group due to weather conditions. At the Winnipeg site ($n = 4$), two participants attended both focus groups while two attended only one focus group. The mean age of MWC users was 66.8 years (range 55–83 years) and had used a wheelchair for a mean of 31.9 years (range 4–60 years). Among the care providers, there were 2 participants at each site ($n = 4$) and all were female. At the Winnipeg site, Jamie was in her 30s and worked in an intentional community home where she was a caregiver for a variety of individuals with a disability, some of who were older adult wheelchair users. Felicia was in her 60s and assisted her husband who was in his 70s and used both a manual and power wheelchair. In Vancouver, Patricia assisted her husband and Bertha provided care for her daughter; in both cases, the care recipient had been a participant in the MWC user focus groups as well. A total of 20 clinicians participated in focus groups between the Winnipeg ($n = 9$) and Vancouver ($n = 11$) sites.

2.2. Data Collection and Analyses. The collection and analyses of the focus group data were central to the program development and revision process. One of the coinvestigators (EG), who had experience in focus group facilitation and expertise in the content area [44–46], cofacilitated all focus groups with a research assistant. All sessions followed a similar format: a brief introduction and audiovisual presentation of the study background, purpose, and design; interactive discussion using a semistructured guide designed for each respective stakeholder group; and opportunity to interact with the training program prototype and provide feedback. The discussion guide included background questions around previous wheelchair training experience and barriers to effective use, motivation for practicing and improving skills, and effective learning strategies and techniques. Discussion occurred prior to presentation of the tablet device and proposed content/features to avoid restricting the scope and breadth of participant feedback. Following the prototype presentation, discussion questions shifted to participants impressions of the program content and delivery; perceptions about appropriateness, engagement, and motivation; and suggestions for revision, addition, or elimination. Each session was audio-recorded and transcribed verbatim by the research assistant, and both facilitators kept field notes of their experience. A second research assistant verified the accuracy of the transcripts against the audio recordings before personal identifiers were removed. Portions of the user and care provider sessions were video-recorded so the investigators could observe participant interactions with the computer tablet.

A *directed content analysis* approach [47]was used with data from each focus group. The initial coding scheme was informed by concepts and themes from Andragogy and Social Cognitive theory, and Edward M. Giesbrecht parsed the content assigning codes to each discrete element, with emergent codes being integrated after each subsequent focus group. The research team met regularly to discuss analysis and coinvestigators WCM and RLW reviewed coding. Any discrepancies were discussed until consensus was reached. An audit trail of the research and analysis process, including all coding procedures, was documented by Edward M. Giesbrecht [48]. Participant collaborators also engaged in this process through debriefing and member checking at the second focus group.

Data analyses and intervention development took place concurrently as some participant feedback was self-evident and easily implemented (e.g., size of icons, number of menus) while other revisions required more in-depth analysis (e.g., relevance of activities and appropriateness of training approach). This iterative feedback/revision process began with an *initial prototype* followed by the development of a *revised* prototype after the first set of focus groups and a *beta* prototype after the second set, culminating in a *clinical prototype* following the final review and pilot-testing phase (see Figure 1).

3. Results

3.1. Initial Prototype Development. A variety of evidence-based resources were used to create the initial content outline, including the Wheelchair Skills Training Program, which is a comprehensive structured curriculum available online [13]. Initially four categories of content were created: safety, wheelchair components, body position, and mobility skills. The mobility skills were structured sequentially and grouped into natural categories, based on underlying prerequisite skills and increasing performance complexity or difficulty. A script was created with the intent of delivering content through a series of short video presentations. Training activities were developed for each curriculum component. To facilitate tablet presentation during focus groups, a "mock" program framework was created with an interactive menu. Several preliminary video segments were integrated (e.g., safety, demonstration of one skill, and sample training activity) for demonstration, but links and proposed features (e.g., the trainee-trainer voicemail function) were nonfunctional placeholders. A storyboard was used to outline the desired sequence and configuration of content. One of the authors (Ian M. Mitchell) oversaw development of the initial software and tablet user-interface.

The development team met regularly in person, via telephone, and by email to discuss design issues, curriculum content, and program delivery. Following each focus group data analysis phase, stakeholder feedback was presented to the team and further redevelopment work was undertaken. The study team used a consensus process to decide which revisions and additions proposed by focus group participants would be incorporated based on consistency with the conceptual framework. Where recommendations were technically and economically feasible, we tended to be inclusive given the fact that the subsequent feasibility trial would enable additional exploration of which features were most beneficial.

The following sections outline stakeholder response and subsequent revision in greater detail.

3.2. Initial Prototype Evaluation: MWC User Response. Participant responses fell into three major themes: *challenges to wheelchair use*, *optimizing strategies for learning skills*, and *critiques of the tablet device*. The input of the MWC user group provided confirmation for elements of the EPIC Wheels program content and strategies for delivering training and also resulted in several changes to the initial program prototype.

3.2.1. Content: Challenges to Wheelchair Use. The focus group discussion guide explicitly intended to elicit from experienced older MWC users the types of environments and activities that were most problematic and the skills that were most beneficial to enhancing participation. MWC users indicated that maneuvering in confined spaces indoors was difficult, particularly doorways, around furniture in small rooms, and negotiating tight corners. Skills such as tight, accurate turns and alternating forward and backwards movements were critical in these situations. Small elevation changes were also noted, such as doorway thresholds and sidewalk cracks or heaves, which can catch the front casters and initiate a forward tip. Soft or accommodating surfaces, such as grass, carpet, and gravel, were particularly difficult for older users with compromised strength. Participants reported that ramps and inclines required both effort and control, coordinating hand movements to prevent rollback during ascent and limiting speed during descent. Curbs and steps were identified as substantial barriers to ascend independently and daunting to descend due to the risk of a forward tip.

Participants also identified awareness of how wheelchair components operate as important to efficient use of the wheelchair. Specifically, operating the wheel locks (brakes) and positioning of the front casters were important knowledge-based components of wheelchair operation. In addition, participants highlighted the relevance of their position within the wheelchair and the impact on operation and responsiveness. For example, leaning forwards or backwards alters the weight distribution between the front and rear wheels, increasing or decreasing wheelchair stability.

3.2.2. Training: Optimizing Strategies for Learning Skills. The older adult MWC users spoke of the importance of a visual demonstration of each skill. Participants preferred "seeing" the task requirements before attempting performance. For example, getting over a doorway threshold could be broken down into positioning casters upon approach, shifting weight backwards, popping the casters over with a quick push, and forward weight shift while propelling the drive wheels over the threshold. Furthermore, demonstration by an older adult peer was deemed to be particularly helpful. Participants cautioned that seeing only "correct" performance was not sufficient. As Ted states: "*So do not always show the successful way . . . show us a way you could go wrong too,*" suggesting training should also include implications of incorrect performance particularly related to safety, including a demonstration. In addition to authentic demonstration models, participants advocated that training should occur in real environments using actual obstacles. In particular, training should occur in the home or community, where the obstacles encountered were truly representative of life situations rather than ones that might be constructed in a clinical setting.

In learning mobility skills, participants stated that success was important to bolster enthusiasm and confidence and that training activities should begin with simple and achievable skills before progressing to more difficult ones. The transition between activities should be graded and the initial speed of performance should be slower to ensure safety. Participants recommended an individualized approach focusing on skills relevant to the specific user, with the trainee having some control over which skills are practiced. Providing a rationale for using each specific skill was stressed. For example, the skill should be presented in the context of a particular situation and explain how acquisition of the skill will improve performance or reduce the risk of injury when performing a relevant activity.

Participants indicated that training activities needed to be engaging and interactive to promote adherence and overcome initial hesitation that might result from fear, low confidence, or apathy. The importance of the relationship between trainer and trainee was noted, identifying that personal contact, individualized evaluation, and feedback would contribute to greater motivation.

3.2.3. Interface: Critiques of the Tablet Device. Participants were impressed with the tablet device as a potential training device. In particular, the portability for use in a community context and the capacity for visual demonstration of individual skills and skill components were highlighted. Participants noted the tablet's built-in capacity for video recording trainee performance had great potential for learning. Concern was raised around the potential for the tablet to be lost, misplaced, or stolen given its small size; ironically, one participant returned to the meeting room shortly after the focus group had finished to locate and retrieve their cell phone.

During the demonstration, we indicated that the training content on the tablet would be delivered using the Internet. Participants expressed apprehension about this dependence on Internet connectivity and what might happen if trainees were without Internet access. Finally, there was considerable discussion around receptivity and capacity of older adults to use the tablet technology. In particular, some participants wondered whether older adults would have the cognitive and attention ability to learn to use the tablet in addition to learning wheelchair mobility skills. This discussion generally reflected participants' perceptions about other older adults and, in particular, those in their late 70s and 80s. All of the focus group participants were over 60 and felt this technology would not be particularly challenging or intimidating for them to use; however, some felt that this might not be the case for others older than they were:

> *"My mother just got an ipad and let me tell you I'm spending more time with my mother*

(laughs) . . . even her touching the screen to select things is a real challenge . . . she often gets totally discombobulated . . . but for someone like me . . . it's second nature" (Louise, 61, Spina Bifida)

"The tablets are neat, but . . . I guess I've got this sort of idea or intuitive sense that people are going to be older . . . closer to 70. . . even for me I'm familiar with that kind of stuff but . . . [for others] it just seems to be easier just to [use a DVD]" (Michelle, 63, Polio).

In particular, participants wondered whether some older adults would have difficulty navigating through multiple menus and icon options and become "lost."

Revisions to Initial Program Prototype. The participants' reporting on common challenges to wheelchair mobility provided confirmation for the content areas proposed in our initial prototype. The specific skills related to addressing environmental barriers, such as propelling straight, turning, and popping casters, were all contained in the training curriculum. The initial prototype had, by design, only a limited repertoire of video content as we anticipated substantial revision. In response to the challenges noted with negotiating small, crowded spaces, we expanded the content related to turning and maneuvering skills. For example, wheelchair casters have an off-center pivot, swinging into a trailing position when initially moving forwards and a leading position when reversing. During these transitions, the wheelchair has a tendency to veer towards one side. A training segment was added specifically addressing this response and how to best control caster swivel. Content was expanded to include explanation of the "mechanics" of turning a wheelchair and broken down into small progressive segments including stationary turns, stopping and turning, moving turns, spin turns, and backward turns. Manipulating body position within the wheelchair to improve stability, safety, and responsiveness of the wheelchair became a separate content area early in the program, as it is a prerequisite for many advanced mobility skills. Additional content was also added related to safety, based on participants' feedback. A separate section identifying equipment (i.e., antitippers, spotter's strap, and gloves) was added as well as educational information regarding tips and falls, spotting/supervision, feedback during training, and injury prevention.

For convenience and expediency, the initial prototype included video demonstrations featuring the first author. In the subsequent prototype, two individuals over 65 years of age (one male and one female) were recruited to model skill performance and training activities. As suggested in the focus groups, a number of skills were enacted with common *errors* to illustrate how and why unsuccessful performance might occur. These included both naturally occurring and contrived errors. Naturally, occurring errors emerged when models were initially unsuccessful attempting a skill and proved useful in demonstrating how to correct mistakes as the models adjusted their approach. Contrived errors were useful as a "3 bears" approach to learning (i.e., demonstrating what happens if you turn too soon, too late, and "just right") and addressed the recommendation to link potential consequences with skill performance.

To address concerns about network connectivity, the system was modified so that it could operate with only sporadic Internet access. In particular, video viewing and training could take place with or without being connected; however, brief connections were still occasionally required so that data and messages could be exchanged with the monitoring trainer.

3.3. Revised Prototype Evaluation. For the second round of focus groups, which involved MWC users, care providers, and clinicians, their responses were again categorized into three general themes: *challenges to wheelchair use*, *optimizing strategies for learning skills*, and *critiques of the tablet training device*. There was overlap between the stakeholder group feedback as well as unique contributions from the three perspectives.

3.3.1. Content: Challenges to Wheelchair Use. With the revised prototype, most of the situations and environments stakeholders identified (and the requisite skills for performance) were contained within the training program content. The clinicians highlighted the challenges of uneven, undulating, and irregular surfaces for older adult MWC users, which were particularly taxing on their endurance. This included working against gravity pushing both up- and downhill and lost momentum when stopping to overcome small gaps or changes in elevation. There was general consensus that performing a sustained wheelie was not an essential or even high priority skill for this user group, but the *transient wheelie* (i.e., popping the casters) was unquestionably a useful and productive strategy to learn.

The care provider groups identified functional upper extremity activities as problematic. This included reaching for objects on the floor and at height, such as operating a ticket kiosk in a public transit station. Using doors was also noted as difficult because it involved manipulating the wheelchair and the door simultaneously and can be compounded by mechanical closers. One MWC user group proposed inclusion of a section on carrying objects, a skill not previously identified specifically in the literature. Since propulsion is often a bilateral activity, transporting an object (such as a cup of hot coffee) is particularly problematic.

Both the MWC user and clinician groups noted the particular challenges of propelling on snow and ice; this was true at both sites, despite the substantial differences in climate between the cities (mean days of snowfall: Winnipeg 53, Vancouver 11; mean depth of snow between December and March: Winnipeg 15 cm, Vancouver 0 cm). Snow can be particularly soft and conforming to the casters, causing them to become buried and the wheelchair to "snowplow" or stop suddenly, causing a risk of forward tipping. In addition, low friction reduces traction at the rear wheels, resulting in one or both wheels spinning.

3.3.2. Training: Optimizing Strategies for Learning Skills. The clinician groups identified that in training older adults,

memory for new learning is often a challenge and needs to be addressed through increased repetition and breaking skills into smaller steps. While they spoke positively about the content of the training videos, both the MWC user and clinician groups indicated the importance of using lay terminology and avoiding excessive technical jargon. At the same time, the MWC users suggested value in using accurate terminology for the wheelchair components to ensure consistency and clarity throughout the learning process. Consistent with the MWC users in round one, the clinicians identified value in describing the benefit of each skill for trainees but also suggested this was important for family and care providers to secure their support in the training process. The care providers wondered whether there might be a benefit to trainees being able to navigate through individual videos to rewind or fast-forward depending upon their learning needs and desires. They affirmed the use of games and interactive activities to engage the trainee in practice, such as the "roller coaster" activity that requires trainees to lean backwards, forwards, and sideways in their wheelchair as the roller coaster car ascends, descends, and turns along the tracks. Care providers also highlighted the need for flexibility to select individual skills and activities, rather than having to follow a prescribed sequence.

The clinician groups suggested that the training program should include not only skills for independent mobility targeting the user but also skills and techniques for care providers to assist older MWC user when independent mobility is not feasible. This was particularly true for skills that might not be reasonably achieved independently, such as managing steps or curbs safely.

3.3.3. Interface: Critiques of the Tablet Device. The clinicians commented that the instruction and demonstration videos were not all of one uniform size and suggested greater consistency and, more importantly, maximizing the size of the video image. Care providers commented that the tablet surface has a significant glare which compromised viewing, particularly when positioned on an angle. The buttons were described as being adequate but somewhat small, and the text was hard to read for some. Likewise, the volume was described as adequate but could potentially be problematic for trainees with compromised hearing. Both the user and care provider groups wondered how the tablet might be positioned and supported during training activities and the risk of it falling to the floor and being damaged.

Revisions to Second Program Prototype. In response to the stakeholder feedback, several additional content areas were introduced. Within the training section related to soft surfaces, we added instruction and video footage on propelling over snow and ice. We also incorporated content specific to care provider (assisted) mobility skills such as getting up and down curbs, steps, and ramps and using the tipping bar to get over small obstacles. Managing doors (with and without closers) and strategies for carrying objects were incorporated as distinct sections.

Several changes were made to the tablet display and user-interface. Video clips were configured to display in the same size configuration. Navigation buttons (e.g., play, pause, and stop) were relocated from below the video image (horizontally) to the right of the image (vertically) to maximize image height and permit a widescreen display. The vertical orientation also permitted an increase in the size of the buttons for easier targeting [49], along with decreasing the amount of text and increasing font size to address visual acuity changes with aging.

We proposed a training schedule of 30 minutes per day (1-2 sessions per day for 15–30 minutes each) at least 5 days per week totaling a minimum of 150 minutes per week. These guidelines are based on the National Blueprint consensus document on promoting physical activity for adults over 50 years, which advocates that lifestyle- or endurance-related activity of moderate intensity should be undertaken for at least 30 minutes (in bouts of at least 10 minutes) 5–7 days per week [50]. All 3 stakeholder groups affirmed this schedule as reasonable and appropriate for the target population.

To address the potential issues with users becoming "lost" during program navigation, we developed 2 strategies—pretraining and reference material. The EPIC Wheels program incorporates two 1 : 1 training sessions with an experienced trainer. In practice, these sessions might occur shortly after an older adult obtains their wheelchair. As part of the initial evaluation and training session, we included a 30-minute interactive orientation to the tablet for the user and care provider. Trainees also receive a handbook that provides instructions for tablet navigation, including screenshots for visual assistance. For simplicity, menus were configured to have 3–8 options related by content area, limiting clutter, and distraction without requiring an excessive number of embedded submenus [51]. We also addressed potential audio issues by including headphones, as augmented audio output increases usability for older adult users of touchscreen technology [51].

The first author (Edward M. Giesbrecht) and a rehabilitation engineer created a lap-mounted support to enable viewing and practice without risk of loss of or damage to the tablet while trainees sit in their wheelchair. A nylon strap and buckle were integrated into a rigid platform with a neoprene foam base, upon which a commercial tablet holder (Cyber Acoustics IS-4000 Universal Tablet Stand, Vancouver, WA) was mounted using hook and loop fasteners (see Figure 2). The tablet could be used in chair or easily removed and placed on a table or other surfaces if desired. A training "kit" was created using common household objects (e.g., boxes, balls, balloons, etc.) at a total cost of less than $20 and could fit in a grocery bag. A kit would be provided to trainees to support all of the tablet-based training activities.

3.4. Beta Prototype: Review and Pilot Testing. Following revision, we met individually with one of the MWC users and one clinician for a final review of the beta prototype. Both reviewers provided confirmation of the scope and presentation of the training content and usability of the user interface, and no substantive revisions were required. In particular, the MWC user was pleased with the tablet holder, indicating it was easy to don and doff in the wheelchair and

Figure 2: Tablet mounting platform for wheelchair use.

provided a good viewing location with adequate adjustability. Subsequently, we conducted pilot testing of the EPIC Wheels program in preparation for a randomized controlled trial. A primary intent was to evaluate the robustness and feasibility of the EPIC Wheels home program and supporting technology. We selected two older adults with diverse wheelchair backgrounds who had no previous involvement in the study. One participant was very experienced, having used a MWC for over 30 years following a spinal cord injury, including participation as a wheelchair athlete in earlier years. The second participant had recently transitioned to MWC use (<6 months) following an above-knee amputation. Participants with diverse wheelchair experience were intentionally selected to obtain perspective from individuals new to MWC use (to determine the acceptability and potential benefit of the EPIC Wheels program) and proficient users (to ensure comprehensiveness and accuracy of the program content). Because the feedback from the reviewers and pilot participants was obtained during tablet use, it was not audio-recorded and transcribed as with the focus groups. Observations and concerns were recorded by the first author and consolidated with the previously obtained data.

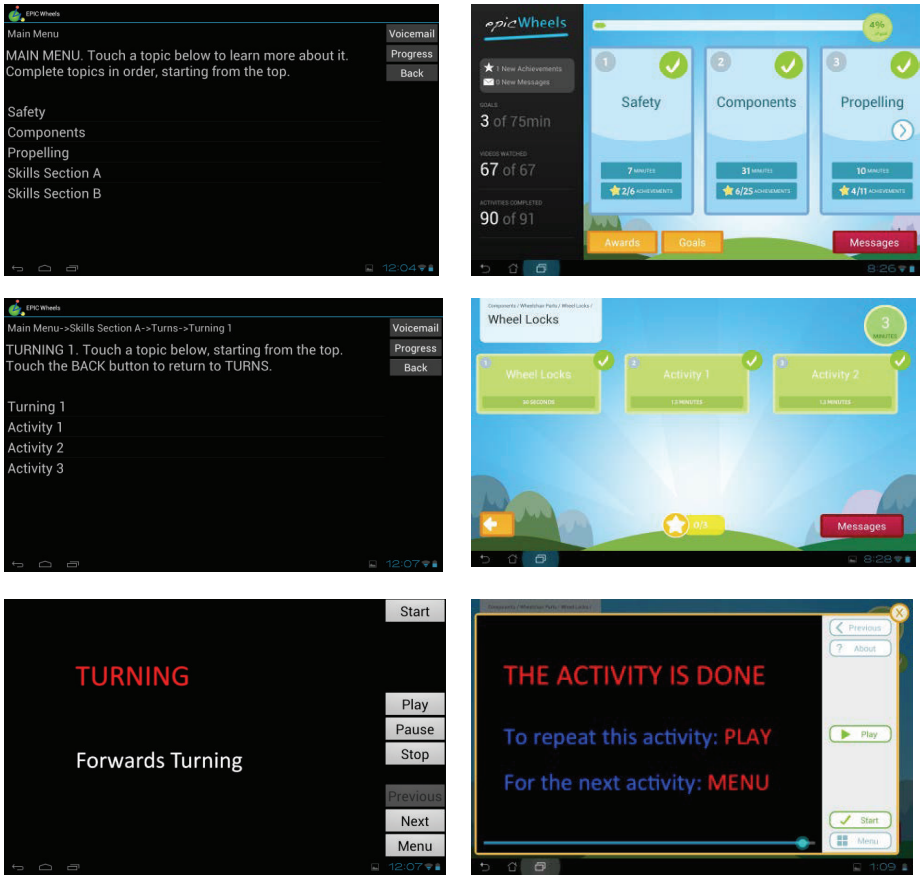
The EPIC Wheels intervention was 4 weeks in length. Participants attended a 1 : 1 session with their trainer (an occupational therapist with wheelchair skills expertise) at the beginning of weeks 1 and 3 and trained at home the remaining days using the tablet device, with the trainer making follow-up calls at the end of the first and third weeks. Participants can use the voicemail feature as a built-in function of the EPIC Wheels program to exchange messages with their trainer, if desired; trainers can respond via their website interface. Because many potential trainees will not have broadband Internet access at home and the Android tablets used in EPIC Wheels are Wi-Fi only, trainees were also given a mobile cellular hotspot device. The tablet can connect through this device to the Internet to upload training data and exchange voicemail messages with the trainer.

After the study period, participants completed a program evaluation questionnaire and provided informal feedback on their experience. As anticipated, the experienced participant indicated he had not learned any new skills but had already been fully proficient with all desired mobility skills for many years. He indicated he would have appreciated such a program when he first obtained his wheelchair and felt there were no important skills or components missing from the program. He commented that the training activities were engaging and fun, although he modified some to increase the challenge because of his level of proficiency. The novice user also found the program comprehensive and identified a substantial benefit, not only due to his capacity for wheelchair mobility but also due to his ability to participate in meaningful activities of daily living. In particular, he identified improvement in turning (related to mobility in tight spaces) and learning to "pop the casters" enabled him to traverse small obstacles like doorway thresholds and propel over soft surfaces like carpet and snow. The lack of large open spaces and hard floor surfaces in his condominium presented some challenges with training activities. The experienced user did not use the voicemail feature, while the novice user had several voicemail exchanges with the trainer. Intermittent connectivity issues with the hotspot device resulted in some data uploading delays and this participant reverted to telephone calls rather than voicemail.

3.5. Clinical Prototype. Several additional changes were made to the clinical prototype, which would be used in a subsequent randomized controlled trial [52].

(1) *Upgrade to the User-Interface Software Program*. The training program was given a more bright and appealing appearance, similar to a commercial software application. The EPIC Wheels program automatically loads upon powering up or waking up the tablet. To increase ease of use and minimize distraction, there are no other applications or features visible. The home screen provides information on the number of minutes spent viewing instructional videos, minutes spent on training activities, and a graphic with weekly progress compared against the goal of 150 minutes (see Figure 3). Videos are accessed through five submenus arranged by content, with blue icons indicating a further embedded menu and green icons indicating that a video will play. A legend at the top of the screen indicates current location within the program menus. All videos display a slider bar with time played/remaining as well as a menu with buttons (previous/next video, play/pause, and exit/back). A stopwatch-like timer with a start/stop button allows trainees to record the amount of time spent on self-training activities. Once a training video or activity is accessed a "check mark" appears on the corresponding button, while a "star" is awarded after completion. To increase motivation, a series of "awards" are provided after completing an increasing number of training activities; trainees can view these by clicking on the Awards icon.

(2) *Upgrade to the Trainer Web-Based Monitoring Software*. As participants are enrolled, an account is created on the trainer's monitoring website. Approximately every 24 hours the tablet attempts to connect with the server via the Internet to upload tablet usage data, providing the trainer with updated information on the number of minutes spent on various training components with each tablet session as well as the number of visits and time spent on each training



(a) Beta prototype (b) Clinical prototype

Figure 3: EPIC Wheels home screen and sample submenu.

activity. The data can be viewed at the website or downloaded in tabular form for further analysis.

(3) *Self-Contained Training Program*. While tablet functionality remains intact, the EPIC Wheels program is being operated as a stand-alone program with other applications hidden using a custom launcher program. All training content is included on the tablet and can be operated independent of Internet connectivity. A mobile WiFi device (AirCard 763S mobile hotspot, Sierra Wireless Inc, Richmond, BC) automatically connects the tablet to the Internet when it is in range (up to 34 metres indoors). The tablet can then update any voicemail messages between trainee and trainer as well as perform its daily upload of tablet training data. However, even if the tablet fails to connect through the hotspot to the Internet, it will continue to operate and record data independently until such time as the connection is reestablished.

(4) *Protection and Safety*. To protect trainee information, the program requires a password for access to protect trainee information and all data is encrypted before storage and uploaded to the secure server. A screen protector on the tablet reduces glare and protects the viewing surface from damage. The software requires trainees to complete the safety content section before permitting access to the remaining training content, and for higher risk content (e.g., popping casters) a pop-up window requires trainees to acknowledge compliance or click on a link to review the safety section. The BORG perceived rate of exertion scale was introduced in the safety section and trainees instructed to limit their effort to "somewhat hard" to prevent overexertion. In addition, content specific to care providers is provided including strategies to enhance effective training, safe spotting, and demonstration of techniques for assisting the wheelchair user during challenging or high-risk activities (e.g., high curbs and steps).

4. Discussion

We were successful in achieving the three objectives of this project. Our older adult MWC user partners were engaged throughout the design and implementation process and all stakeholder groups provided substantial contributions to the development of a clinical prototype that is currently being evaluated in a RCT. The PAD framework proved to be a valuable approach to creating the EPIC Wheels program. The iterative consultation process provided critical input into the evolving content and user interface. Incorporating a number of stakeholder groups provided validation for relevance and

appropriateness of the included content. The MWC users confirmed the scope of skills included was comprehensive and contributed to inclusion of additional material such as the task of carrying an item while propelling a wheelchair. Care providers negotiated that training content around some high-level skills (e.g., wheelies and ascending steps) should be restructured with assisted, rather than independent, strategies. The clinician groups confirmed skills that were most enabling and often neglected among older adults, such as transient wheelies, and provided input on teaching strategies. Focus groups were particularly useful as they facilitated interaction and discussion among participants. The resulting dialogue was often animated and engaging, and there was not always agreement or consensus. While this made analysis more challenging, the outcome was a richer and more comprehensive product with greater potential for application to a broad audience.

The critiques and recommendations by stakeholders proved to be consistent with, and confirmatory of, the theoretical bases with which EPIC Wheels was created. Four key components of self-efficacy, as proposed in Social Cognitive Theory [53], were evident. The sequencing of skills from basic to advanced and the inclusion of multiple training activities for each skill graded from simple to complex maximize opportunity for successful skill performance or *mastery experience* which has the strongest influence on self-efficacy [54]. Early success experiences induce confidence that more difficult skills are attainable and enhance perseverance among trainees. Progress is monitored by trainers, who encourage skill advancement following successful performance but before proficiency, creating a "just right" challenge as proposed in the occupational therapy literature [55, 56]. The recommendation to include age-appropriate demonstrators of both sexes corresponds to *vicarious experience* or the observation of a comparable peer achieving success in a given skill, which is the penultimate factor influencing self-efficacy [54]. Knowles [29] also promotes the value of modeling to provide a rationale for older adults to pursue a specific skill, as it has been associated with improvement in skill performance [57]. A third component is the encouragement of meaningful others, or *verbal persuasion*. Stakeholders advocating for regular monitoring and follow-up by the trainer and for inclusion of spotting, training, and feedback strategies specifically for care providers in the EPIC Wheels program were particularly relevant in this regard. Finally, appropriate management and interpretation of one's *physiological state* is important to wheelchair confidence. The inclusion of games and other engaging training activities increase motivational investment while distracting trainees from the demands of performing mobility skills. While some older adults may be unfamiliar with computer games, we anticipate their inclusion will increase training time as recent studies show promising results in this regard, even among the very old [32, 33]. We also included information on self-monitoring physical expenditure, including information on the BORG Perceived Rate of Exertion scale [58] and parameters for not exceeding the recommended level of "somewhat hard" during training, based on best practice guidelines [59].

Stakeholders also provided input that aligned with Andragogical principles. Adult learners, particularly older adults, prefer an autonomous and self-directed approach that is goal-oriented and respectful [31]. The EPIC Wheels program allows trainees to control the time and location of training activity and provides continuous updates on the number of components completed and total time spent in practice. Flexible navigation ensures trainees can control which specific skills they want to work on, advancing when they feel ready and revisiting material if desired. Trainers assist in prioritizing skills most relevant to trainee goals and activities of interest. Providing a rationale for each skill in relation to specific occupations of interest, inclusion of typical daily activities and commonplace equipment for practice, and demonstration of incorrect performance with the resultant hazards offer a practical and life-experience approach to learning consistent with Andragogical principles.

A key benefit of the PAD approach was optimizing the tablet interface. Despite the numerous benefits that a tablet offers, such as touch screen access, interactivity, portability, and Internet connectivity, acceptance and adherence by older adults are critical to the success of this home program. By bringing evolving prototypes back to the target users, as well as other stakeholders, we were able to ensure usability by older adults. Although older adults are less likely to use technologies such as computers and cell phones than young people, computer use is continuing to grow. Recent studies in the United States found 84% of those over 60 years had experience with computers [60] and 40% of those over 65 years are regular computer and Internet users [61]. Use of a computer tablet involves some new learning, and age-related declines in memory and fluid intelligence may restrict uptake. These issues can be addressed through self-paced training structured for success experiences to build confidence and adapting the interface design for familiarity and ease of use with minimal memory requirements [57, 62].

At the conclusion of the PAD process the EPIC Wheels program and training tablet demonstrated robust and consistent performance and are currently being evaluated in a randomized controlled trial with novice older adult MWC users [52]. The training program is downloaded onto a tablet with an individualized identity and password for each trainee and a corresponding identity is created on the trainer's website, located on a secure server. The wheelchair user can perform training independently or with supervision by a care provider, particularly when more advanced or higher risk skills are being learned.

Future development will focus on several improvements. Communication between trainee and trainer is currently conducted via voicemail, but the capacity for recorded video communication is already in place. Expediting video data transfer and integration of a video player applet are under development and the potential for real-time video communication is also being explored. The software content and user interface are self-contained and preloaded on the trainee tablet. A content management software program will provide the potential for trainers to customize a trainee's program, adding and removing content as desired. In the future, this would allow a trainer to release new content over time via the

Internet. Using built-in or external sensors could expand the scope of interactive training activities used and collection of performance-based data.

Some limitations with the EPIC Wheels program should be noted. The training content specifically targets MWC users who propel with both upper extremities, including those who also use one or both lower extremities. Other propulsion strategies, such as one arm and one leg used by individuals with hemiplegia, and mobility equipment (e.g., power wheelchairs, scooters) are common and require a different set of skills. Such content will need to be created to address these users groups. The software developers were proficient with the Android platform and EPIC Wheels is currently available only on these devices; creating a version compatible with the Apple iPad would facilitate broader appeal and availability. Finally, while the tablet tracks all program interactions and uploads detailed activity information to the trainer website, there is no way to verify that trainees physically engage in training beyond viewing the program content. In future, synchronizing training activities with input from a data logger or tablet accelerometer may address this issue.

5. Conclusion

A participatory action design process proved valuable in the development and refinement of a tablet-based wheelchair skills home training program. The involvement of older adult wheelchair users, as well as care providers and clinician stakeholders, was critical to achieving a product that was both comprehensive and acceptable to the target users. The contributions of these research partners confirmed the underlying theoretical principles of self-efficacy and adult learning theory upon which the program was developed. The clinical prototype that emerged is currently under evaluation in a randomized controlled trial and further enhancements to the current program are anticipated in the near future.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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Research Article

Effect of Wheelchair Frame Material on Users' Mechanical Work and Transmitted Vibration

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Wheelchair propulsion exposes the user to a high risk of shoulder injury and to whole-body vibration that exceeds recommendations of ISO 2631-1:1997. Reducing the mechanical work required to travel a given distance (WN-WPM, weight-normalized work-per-meter) can help reduce the risk of shoulder injury, while reducing the vibration transmissibility (VT) of the wheelchair frame can reduce whole-body vibration. New materials such as titanium and carbon are used in today's wheelchairs and are advertised to improve both parameters, but current knowledge on this matter is limited. In this study, WN-WPM and VT were measured simultaneously and compared between six folding wheelchairs (1 titanium, 1 carbon, and 4 aluminium). Ten able-bodied users propelled the six wheelchairs on three ground surfaces. Although no significant difference of WN-WPM was found between wheelchairs ($P < 0.1$), significant differences of VT were found ($P < 0.05$). The carbon wheelchair had the lowest VT. Contrarily to current belief, the titanium wheelchair VT was similar to aluminium wheelchairs. A negative correlation between VT and WN-WPM was found, which means that reducing VT may be at the expense of increasing WN-WPM. Based on our results, use of carbon in wheelchair construction seems promising to reduce VT without increasing WN-WPM.

1. Introduction

Whereas wheelchairs contribute greatly to the physical activity, mobility, and autonomy of their users, wheelchair propulsion is also physiologically detrimental. About half of manual wheelchair users will develop shoulder injuries due to the high mechanical load at the shoulders during propulsion [1–4]. It is believed that improving propulsion efficiency can help in preserving the shoulder function [5–7]. One way to achieve this goal is to propel a more efficient wheelchair, that is, wheelchair that minimizes the required mechanical work to travel a given distance (work-per-meter (WPM)). On the other hand, the regular use of a wheelchair exposes the users to whole-body vibrations and shocks that exceed the recommendations of ISO 2631-1:1997, which may be detrimental for both their comfort and safety [8–10]. This means that, currently, wheelchair design faces a dual challenge in reducing both WPM and vibration transmissibility (VT).

Frame material is believed to have an impact on WPM and VT. Originally, most wheelchairs were made of steel, but alternative materials such as aluminium, titanium, and carbon are now becoming more and more popular, mostly because they allow building lighter and stiffer wheelchairs [11]. However, little is known about the effect of frame material on WPM and VT. The impact of wheelchair weight on WPM is still debated. Lighter wheelchairs are believed to decrease WPM because rolling resistance is proportional to weight. However, de Groot et al. [12] did not observe changes in any kinetic parameter or oxygen uptake when propelling a wheelchair with 5 kg or 10 kg extra. On the other hand, Cowan et al. [13] did not measure power output or WPM but did observe higher peak forces and lower self-selected velocities with 9.1 kg extra. Titanium wheelchairs are believed to transmit less vibration than aluminium wheelchairs [6, 14], but evidence contradicts this claim [15, 16]. Carbon wheelchairs are also thought to transmit less vibration than aluminium

wheelchairs, but no study currently exists to prove this claim. Therefore, there is a need for a better understanding of the frame material impact on WPM and VT.

The choice between a folding and a rigid frame also has an impact on WPM and VT, and this choice is really up to the user's preferences. A survey of 549 wheelchair users associated folding wheelchairs with increased shoulder pain compared to rigid wheelchairs when propelling for more than 10 minutes or when propelling up a ramp [7]. In terms of VT, although Garcia-Mendez et al. [9] found no difference between 9 folding wheelchairs and 20 rigid wheelchairs in everyday conditions, Kwarciak et al. [15] found that, for a curb-descending task with 16 wheelchairs, the vibration transmitted by folding wheelchairs was higher than by rigid wheelchairs when acceleration was frequency-weighted as recommended by ISO 2631-1:1997. Therefore, to limit the scope of this work, only folding wheelchairs were considered.

Other wheelchair components also have an impact on VT. Although the impact of rear suspension on VT is mitigated [9, 15, 18–20], caster suspension does reduce lower body vibration [19]. However, as rear suspension is only available on rigid frames and caster suspensions are add-on devices that are not intrinsic to the wheelchair, we excluded suspension from this study. Composite wheels are also very popular devices that are advertised to improve WPM and VT [14]. However, as both claims were not verified [21, 22] and as these wheels are also add-ons that are not intrinsic to the wheelchair, we excluded composite wheels from this study and concentrated only on the effect of frame material.

The purpose of this work is twofold. The first is to measure the effect of frame material on folding wheelchairs' WPM and VT. Following the common belief that carbon and titanium wheelchairs have better vibration properties than aluminium wheelchair [6, 14], we hypothesized that carbon and titanium wheelchairs have lower vibration transmissibility (VT) than aluminium wheelchairs. The second is to investigate a possible compromise between WPM and VT. In fact, if a wheelchair is designed to deform to absorb vibration and shocks, some propulsion energy may be lost in this deformation; thus, reducing VT could increase WPM. We made the second hypothesis that a negative correlation exists between WPM and VT. To verify these hypotheses, six commercially available ultralight folding wheelchairs made of titanium, carbon, and aluminium were compared in terms of both WPM and VT. 10 able-bodied subjects propelled the six wheelchairs on three ground surfaces. WPM was measured using two instrumented rear wheels, and VT was measured using five accelerometers placed into the seat cushion and on the frame.

2. Methodology

ISO 2631-1:1997 standard on mechanical vibration and shock [8] defines the methods to assess the user's vibration exposure in seated position, which is defined by these parameters.

a_w : root-mean-square (RMS) value of the frequency-weighted acceleration (base analysis): this figure is related to the continuous vibration transmitted to the user and is expressed in m/s^2.

VDV: vibration dose value (complementary analysis): this figure is related to the shock-induced vibration transmitted to the user and is expressed in m/s$^{1.75}$.

Principal requirements of the norm are as follows.

(1) Measurements are made in situations approaching real-life conditions.

(2) Measurements are made at the interface between the user and the source of vibration.

(3) Measurements are made orthogonally to the ground.

(4) Measurements bandwidth must cover the 1 to 80 Hz spectrum.

This norm was followed at different extents in wheelchair propulsion studies [9, 10, 23]. A variant of this norm was also used in DiGiovine et al. [24] and Garcia-Mendez et al. [25] to measure the vibration transmissibility of seat cushions. They made the ratio between the vibration measured on the subject (output) and the vibration measured below the seat (input). In our case, this can be adapted to measure the wheelchair frame VT, by comparing the vibration measured at the seat-user interface (output) to the vibration measured at the four wheels hubs (inputs).

It is important to note that ISO 2631-1:1997 is based on studies performed with able-bodied persons, subjected to vibration in a passive way. As such, it cannot be completely generalized to wheelchair propulsion, and it should be interpreted carefully when used on the SCI population. A discussion on the application of ISO 2631-1:1997 to wheelchair propulsion is given in the appendices.

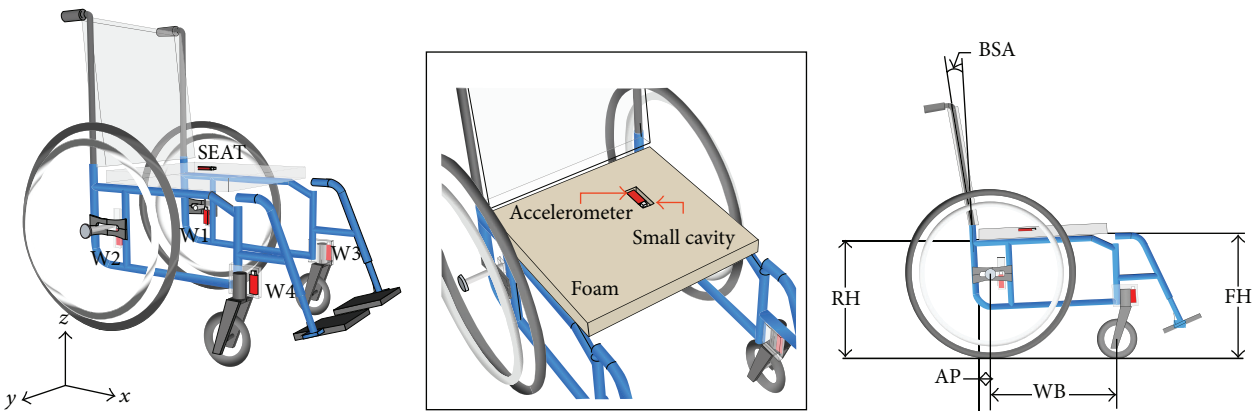
ASTM F1951 standard specification for determination of accessibility of surface systems under and around playground equipment [26] uses the work-per-meter (WPM) method described in Chesney and Axelson [17] to compare the mechanical work required to propel the same wheelchair on different ground surfaces, based on instrumented wheels data. The wheelchair must be propelled from stop on a 2 m surface and stop by itself exactly at the end of the two meters. This requirement is to ensure that all the energy produced as work by the user is exhausted at the end of the run. This however requires propelling at a very low average speed of 0.3 m/s and thus violates the first requisite of ISO 2631-1:1997 that measurements must be representative of real-life conditions.

A similar method was used in Cooper et al. [23] to measure the average mechanical work required to traverse different surfaces, but at a steady state average velocity of 1 m/s over a 7.6 m long surface. However, they did not specify if the final velocity was always equal to the initial velocity, which is very important. In fact, from conservation of energy, we have

$$E_0 + W_{\text{nc}} = E_1, \tag{1}$$

where E_0 and E_1 are the initial and final total mechanical energy and W_{nc} is the work done by nonconservative forces. W_{nc} can be expressed as

$$W_{\text{nc}} = W_{\text{prop}} + W_{\text{res}}, \tag{2}$$



(a) Overview (b) Details of the seat accelerometer location (c) Wheelchair adjustments

Figure 1: Accelerometers placement on the wheelchairs.

where W_{prop} is the propulsion work generated at the wheels by the user and W_{res} is the resistive work dissipated by friction and vibration. To measure W_{res} using instrumented wheels, one must be sure that $E_1 = E_0$, so that $|W_{\text{res}}| = |W_{\text{prop}}|$. Propulsion work is obtained by

$$W_{\text{prop}} = \int_{\theta_{R0}}^{\theta_{R1}} M_R d\theta_R + \int_{\theta_{L0}}^{\theta_{L1}} M_L d\theta_L, \tag{3}$$

where M_R and M_L are the right and left moments applied on the wheels by the user, $d\theta_R$ and $d\theta_L$ are the right and left wheel angle variations, and θ_{R0}, θ_{R1}, θ_{L0}, and θ_{L1} are the initial and final angular position of right and left rear wheels. WPM is obtained by dividing W_{prop} by the travelled distance L.

In this work, WPM was calculated using a realistic steady state average velocity of 1 m/s on a 7 m run. After each recording, we kept only a 5 m interval within these 7 m where the final velocity was equal to the initial velocity. Contrarily to the ASTM F1951 standard, multiple users performed the same task. Therefore, to take the differences of body and wheelchair weights into account, WPM was weight-normalized by the total mass of the users and wheelchairs, so that

$$\text{WN-WPM} = \frac{\int_{\theta_{R0}}^{\theta_{R1}} M_R d\theta_R + \int_{\theta_{L0}}^{\theta_{L1}} M_L d\theta_L}{Lm}, \tag{4}$$

where WN-WPM stands for weight-normalized work-per-meter, L is the distance (5 m), and m is the mass.

2.1. Material. A total of six folding wheelchairs were tested and compared. To measure the impact of frame material on WN-WPM and VT, three similar wheelchairs featuring a single cross-brace folding mechanism were compared: one made of titanium (Ti), one made of carbon (C), and one made of aluminium (Al1). To consider the effect of folding design on WN-WPM and VT, three additional aluminium wheelchairs (Al2, Al3, and Al4) were tested, each featuring a different folding mechanism. Table 1 lists these wheelchairs along with their weight, material, and folding design.

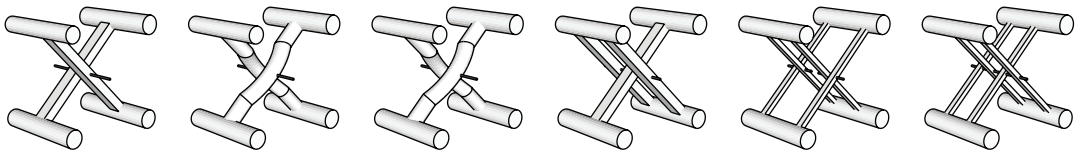
All wheelchairs used the same wheels. Solid tires were used on all wheels, so that tire pressure did not need to be monitored. Wheelchairs were weighed without rear wheels and seat cushion using an AMTI-OR6 force platform with a resolution of 150 g. Weight distribution on the wheels was not verified using a force platform, but wheelchairs were adjusted equally according to Figure 1(c), with an anteroposterior seat position (AP) of 4.4 ± 0.3 cm, a rear seat height (RH) of 41.0 ± 0.6 cm, a front seat height (FH) of 44.9 ± 2.0 cm, a wheel base (WB) of 48.4 ± 2.8 cm, a wheelchair width (including handrims) of 66.2 ± 4.0 cm, a rear wheels diameter of 60 cm, and a backseat angle (BSA) of 1.7 ± 2.6 degrees. Although backrest supports were different from one wheelchair to another (the stock backrest supports were used), the same midrange foam seat cushion with a width of 3 cm was used on every wheelchair.

Five triaxial piezoelectric accelerometers with a bandwidth of 240 Hz (VR001, Midé Technology) were installed on the wheelchairs as shown in Figure 1 and were sampled at 3200 Hz. Accelerometers W1 to W4 were installed on small aluminium plates fixed on the frame, so that they recorded the vibration induced on the frame by the wheels. Accelerometer SEAT was installed into a small cavity above the seat cushion, just below the user's left ischion. This accelerometer recorded the vibration transmitted to the user. A SIT-BAR indenter was not used because it would have modified the pressure distribution at the seat-user interface, which is not allowed by ISO 2631-1:1997. All accelerometers were installed orthogonally to the ground and wheelchair.

Rear wheels were instrumented wheels (SmartWheel, Out-Front Corp.), weighing 4.73 kg each. Forces and moments applied by the user on the rear wheels were sampled at 240 Hz, along with the angular position of the rear wheels.

2.2. Methods of Experiment. The experimental protocol was approved by the Research Ethics Committee (CÉR) of the École de Technologie Supérieure (ÉTS). 10 able-bodied subjects were recruited for this experiment, which took place at ÉTS. Subjects distribution was 6 men and 4 women, with an average weight of 73.4 ± 13.3 kg, an average height of 173.9 ± 8.1 cm, and an average body-mass index (BMI) of

Table 1: Properties of the tested wheelchairs.

Wheelchair	Ti	C	Al1	Al2	Al3	Al4
Model	2GX	Helio C1	Helio A7	Quickie 2	Litestream XF	Catalyst 5
Company	TiLite	Motion Composites	Motion Composites	Sunrise Medical	Pride Mobility	Ki Mobility
Weight (w/o rear wheels and seat cushion)	10.25 kg	8.57 kg	9.54 kg	10.37 kg	12.48 kg	9.44 kg
Material	Titanium	Carbon	Aluminium	Aluminium	Aluminium	Aluminium
Folding design	 One single cross-brace	One single cross-brace	One single cross-brace	One tri-cross-brace	Two single cross-braces	One dual cross-brace

$24.2 \pm 3.3\,\mathrm{kg/m^2}$. After giving their informed consent, the subjects were instructed to practise propelling a random wheelchair at 1 m/s. This velocity was first controlled with a chronometer, and subjects were then asked to try to keep this velocity during all trials.

After the period of familiarization, subjects were asked to sit on one of the six instrumented wheelchairs and to propel successively on three different ground surfaces:

(1) smooth vinyl floor;

(2) textured rubber mat with a diamond-shaped pattern (1 mm thick diamonds, 2200 diamonds per square meter);

(3) obstacle: smooth vinyl floor with one bump (rectangular section, 5 mm thick, 30 mm wide).

Testing conditions are shown in Figure 2. For each condition, subjects started 2 meters away from the surface, accelerated until they reached their steady state velocity, and propelled on the 7 meters of the surface. Each trial was performed three times. After propelling on each surface, data were transferred to a computer and the accelerometers and instrumented wheels were installed on the next wheelchair. Subjects were then asked to repeat the same steps with the new wheelchair.

Every subject propelled all six wheelchairs on the three ground surfaces. Wheelchairs and ground surfaces order were randomized to avoid a bias due to the fatigue of the subjects. At the end, a total of 540 trials were analyzed (10 subjects × 6 wheelchairs × 3 surfaces × 3 trials).

2.3. Processing of Vibration Data. All data processing was performed with Matlab (The Mathworks, Inc., Natick, MA). For each trial, the following steady state data were kept for analysis.

Smooth Vinyl Floor. From the start of the third push to the end of the last push.

Textured Rubber Mat. From the start of the third push, as long as the wheelchair was on the textured mat.

Smooth Vinyl Floor with Bump. Two seconds before and after the front wheels roll over the bump.

Parameters a_w and VDV were then calculated on the three axes of each accelerometer:

$$a_{wi(\mathrm{ACC})} = \left(\frac{1}{T}\int_0^T \left(w_i(t) * a_{i(\mathrm{ACC})}(t)\right)^2 dt\right)^{1/2},$$
$$\mathrm{VDV}_{i(\mathrm{ACC})} = \left(\int_0^T \left(w_i(t) * a_{i(\mathrm{ACC})}(t)\right)^4 dt\right)^{1/4}, \quad (5)$$

where i is the axis (x, y, z), ACC is the accelerometer identifier (SEAT, W1, W2, W3, or W4), $a_{i(\mathrm{ACC})}(t)$ is the acceleration measured on axis i in $\mathrm{m/s^2}$, $w_i(t)$ is the impulse response of the frequency weighting transfer functions given by ISO 2631-1:1997, T is the total recording time, and $*$ is the convolution operator.

ISO 2631-1:1997 states that vibration should be reported on the axis with the highest vibration. However, a special case is also accepted when vibration is comparable on two axes, in which case both vibration values can be combined into a total vibration. In our data, the average vibration at the seat for every trial was highest on the z-axis but comparable to the x-axis. Vibration on the y-axis was about 80% lower than x and z. Literature does not agree on the choice between the vertical (z) vibration or the total (x-z) vibration for wheelchair vibration assessment [9, 10, 23, 24]. Therefore, both were calculated. The choice between one or the other is discussed in the appendices. Total (x-z) vibration was calculated as follows:

$$a_{wt(\mathrm{ACC})} = \sqrt{k_x^2 a_{wx(\mathrm{ACC})}^2 + k_z^2 a_{wz(\mathrm{ACC})}},$$
$$\mathrm{VDV}_{t(\mathrm{ACC})} = \sqrt[4]{k_x^4 \mathrm{VDV}_{x(\mathrm{ACC})}^4 + k_z^4 \mathrm{VDV}_{z(\mathrm{ACC})}^4}, \quad (6)$$

where $k_x = 1.4$ and $k_z = 1$ [8].

In total, eight vibration parameters were assessed. The vibration transmitted to the user by the seat was defined by the following four parameters:

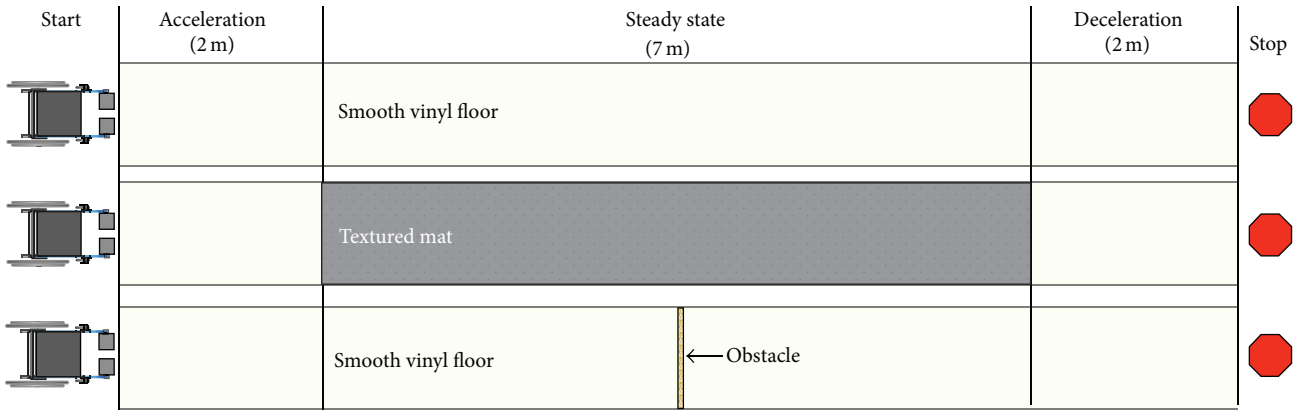


Figure 2: Testing conditions.

$a_{wz(\mathrm{SEAT})}$: vertical (z) continuous vibration (m/s^2),

$\mathrm{VDV}_{z(\mathrm{SEAT})}$: vertical ($z$) shock-induced vibration (m/s$^{1.75}$),

$a_{wt(\mathrm{SEAT})}$: total (x-z) continuous vibration (m/s^2),

$\mathrm{VDV}_{t(\mathrm{SEAT})}$: total ($x$-$z$) shock-induced vibration (m/s$^{1.75}$).

The vibration transmissibility (VT) of the frame (%) was defined by the following four parameters:

VT_{awz}: vertical (z) continuous vibration transmissibility,

$\mathrm{VT}_{\mathrm{VDV}_z}$: vertical ($z$) shock-induced vibration transmissibility,

VT_{awt}: total (x-z) continuous vibration transmissibility,

$\mathrm{VT}_{\mathrm{VDV}_t}$: total ($x$-$z$) shock-induced vibration transmissibility,

where

$$\mathrm{VT}_{aw\{z,t\}} = \frac{a_{w\{z,t\}(\mathrm{SEAT})}}{(1/4)\sum_{i=1}^{4} a_{w\{z,t\}(\mathrm{W}i)}} \times 100\%,$$
$$\mathrm{VT}_{\mathrm{VDV}\{z,t\}} = \frac{\mathrm{VDV}_{\{z,t\}(\mathrm{SEAT})}}{1/4\sum_{i=1}^{4} \mathrm{VDV}_{\{z,t\}(\mathrm{W}i)}} \times 100\%. \quad (7)$$

For each subject, wheelchair, and surface, the mean of each parameter was taken over the three trials.

2.4. Processing of Mechanical Work Data. The mechanical work was assessed on the smooth vinyl floor and on the textured rubber mat. Mechanical work analysis used the same steady state data as vibration analysis. For each trial, we kept a 5 m subset where the final wheelchair velocity was equal to the initial wheelchair velocity. Weight-normalized work-per-meter (WN-WPM) was then computed from (4). For each subject, wheelchair, and surface, the mean of the WN-WPM was taken over the three trials.

2.5. Statistical Analysis. Lilliefors tests were performed to assess the normality of WN-WPM values and every 8 vibration parameters values. Tests were performed independently for the six wheelchairs. We found that normality cannot be rejected ($P < 0.05$) for the majority of our samples; therefore, a parametric statistical analysis was selected. For each parameter, an analysis of variance (one-way ANOVA) was performed over the six wheelchairs. When the ANOVA resulted in a P value below 0.05, a Tukey-Kramer post hoc test was performed to determine which wheelchair(s) stood out from the others. Statistical analysis was performed under Matlab using `anova1` and `multcompare` methods.

3. Results

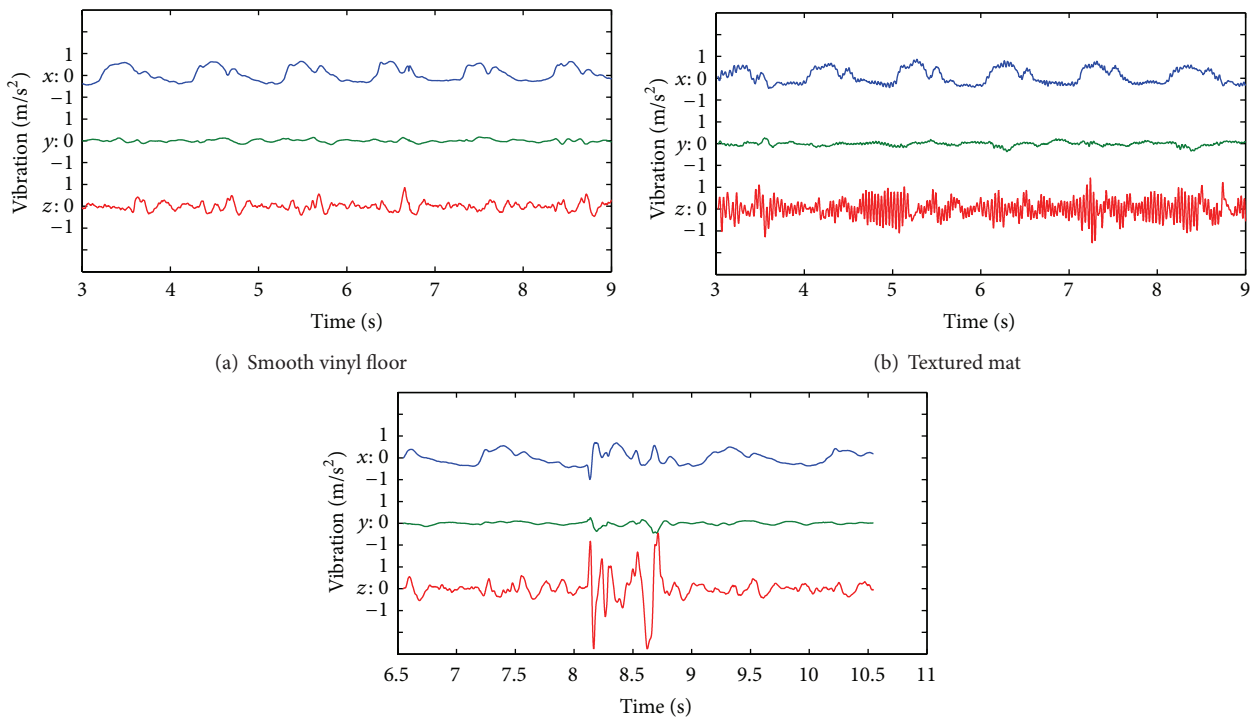
3.1. Vibration. Figure 3 shows samples of the steady state frequency-weighted vibration recorded on the three axes for the three surfaces. The shape of the signals was similar for all subjects and wheelchairs. On these samples, the main source of anteroposterior (x) vibration appears to be the self-induced acceleration and deceleration of the wheelchair due to propulsion. We also observe that mediolateral (y) vibration is negligible compared to the other axes. Finally, the main effect of vibration due to the textured mat or obstacle appears to be on the vertical axis (z).

Table 2 shows the average values for the crest factor, MTVV_z/a_{wz}, and $\mathrm{VDV}_z/a_{wz}T^{1/4}$, which are defined in ISO 2631-1:1997. If, for a given condition, these values are equal to or greater than 9, 1.5, and 1.75, respectively, a complementary vibration analysis is required (VDV). Table 2 shows that this complementary analysis is indeed justified.

Table 3 shows the average vibration transmitted to the subjects when rolling on the three surfaces. No significant difference between wheelchairs was observed in any situation.

Table 4 shows the average vibration transmissibility (VT) of each wheelchair when rolling on the three surfaces. Contrarily to Table 3, significant differences between wheelchairs were observed. In terms of continuous vibrations (T_{aw}),

(1) on the smooth vinyl floor, C and Al1 had a lower vertical (z) VT than Ti and Al3;



(a) Smooth vinyl floor

(b) Textured mat

(c) Obstacle

Figure 3: Samples of the triaxial steady state frequency-weighted acceleration recorded at the seat for the three surfaces.

Table 2: Requirements for complementary analysis.

	Smooth vinyl floor	Textured mat	Obstacle
Crest factor on vertical axis (max. 9)	5.14	3.38	6.10
MTVV_z/a_{wz} (max. 1.5)	**1.55**	**1.48**	**1.48**
$\mathrm{VDV}_z/a_{wz}T^{1/4}$ (max. 1.75)	1.49	1.33	**2.01**

Bold values indicate that a complementary analysis (VDV calculation) is required.

(2) on the textured mat, C, Al1, Al2, and Al4 had a lower vertical (z) VT than Al3;

(3) on the textured mat, C, Al2, and Al4 had a lower total (x-z) VT than Al3;

(4) on the obstacle, a significant difference of total (x-z) continuous VT was observed between wheelchairs; however, the post hoc test did not allow discriminating one wheelchair from the others.

In terms of shock-induced vibrations (T_{VDV}),

(1) on the smooth vinyl floor, C and Al1 had a lower vertical (z) VT than Ti and Al3;

(2) on the smooth vinyl floor, Al2 had a lower vertical (z) VT than Al3;

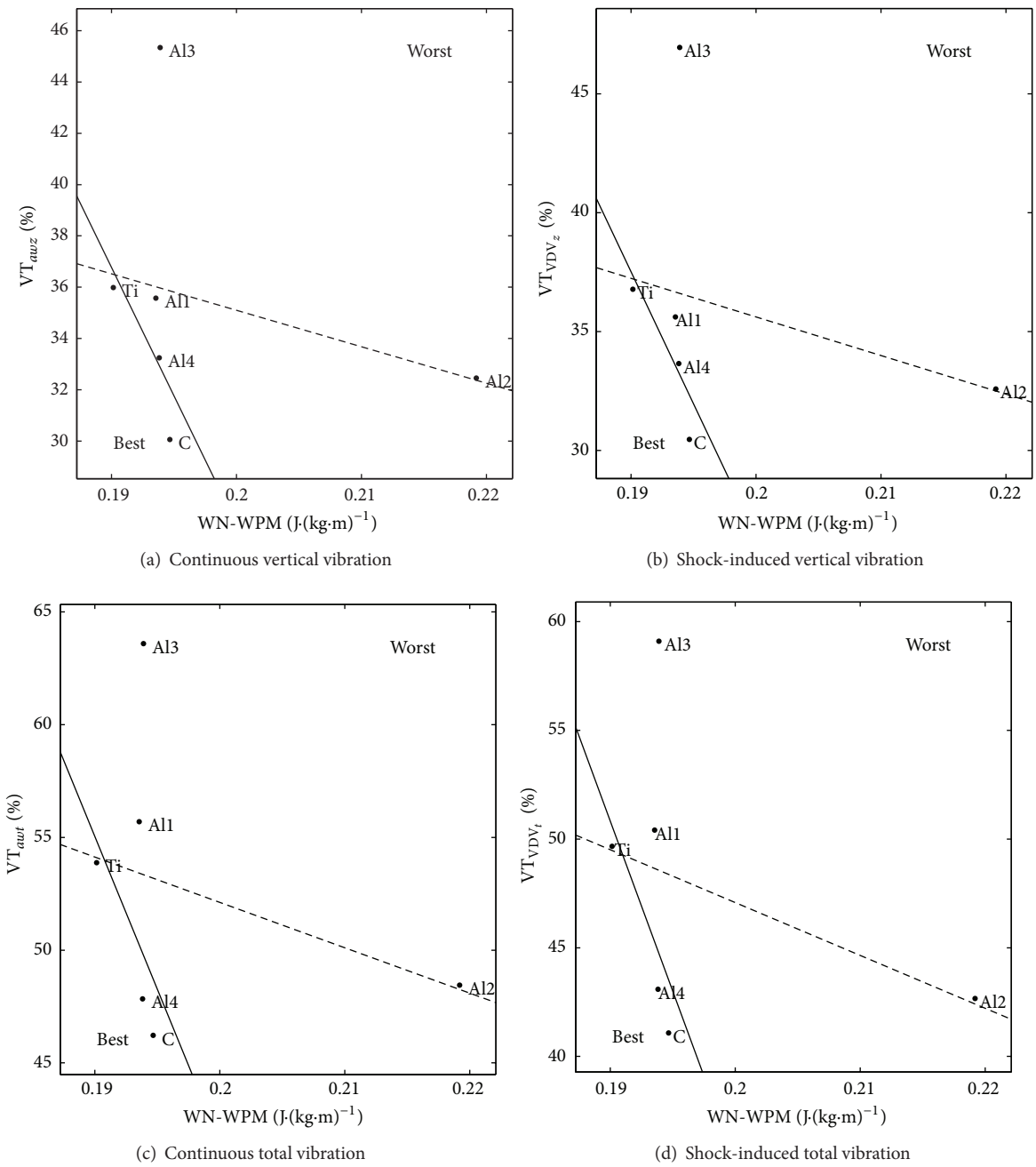
(3) on the textured mat, all wheelchairs had a lower vertical (z) VT than Al3;

(4) on the textured mat, C, Al2, and Al4 had a lower total (x-z) VT than Al3;

(5) on the obstacle, a significant difference of total (x-z) continuous VT was observed between wheelchairs; however, the post hoc test did not allow discriminating one wheelchair from the others.

3.2. Mechanical Work. Table 5 shows the average WN-WPM values for all wheelchairs on the smooth floor and on the textured mat. No significant difference was observed between wheelchairs ($P > 0.1$ for the smooth floor and $0.05 < P < 0.01$ for the textured mat).

3.3. Relation between WN-WPM and VT. Figure 4 shows the average of all trials for each wheelchair, where the x-axis is the weight-normalized work-per-meter and the y-axis is the vibration transmissibility. A wheelchair situated at the left requires less mechanical work to travel the same distance, and a wheelchair situated at the bottom has lower vibration transmissibility. Therefore, the best wheelchair in terms of both parameters is the nearest to the origin. We selected the propulsion on the textured mat instead of smooth vinyl,



(a) Continuous vertical vibration

(b) Shock-induced vertical vibration

(c) Continuous total vibration

(d) Shock-induced total vibration

Figure 4: Vibration transmissibility (VT) versus weight-normalized work-per-meter (WN-WPM) when propelling on a textured mat.

because this combination maximized the differences between wheelchairs on both VT and WN-WPM.

A first-order regression between VT and WN-WPM is shown as a dashed line in Figure 4. A small negative correlation was observed between VT and WN-WPM, with a coefficient of correlation ranging from −0.29 to −0.42 in the four cases of Figure 4. We found that the group composed of Al2 and Al3 stood out compared to Ti, C, Al1, and Al4 and thus may be considered as outliers. When Al2 and Al3 were removed, the coefficient of correlation increased to a range of −0.68 to −0.83. The first-order regression with Al2 and Al3 removed is shown as a solid line in Figure 4. These results suggest that VT is indeed negatively correlated to WN-WPM.

4. Discussion

4.1. Vibration. In Table 3, we observed that, for all parameters, the vibration was higher on the textured mat and on the obstacle compared to the smooth vinyl floor. This was expected as the smooth vinyl floor was not expected to provide significant vibration or shocks. This observation also matches samples in Figure 3. Although vibration at the

Table 3: Vibration transferred to the user.

WC	$a_{wz(\text{SEAT})}$ (m/s^2)	VDV$_z$ (m/s$^{1.75}$)	$a_{wt(\text{SEAT})}$ (m/s^2)	VDV$_t$ (m/s$^{1.75}$)
		Smooth vinyl floor		
Ti	0.25 ± 0.07	0.56 ± 0.14	0.63 ± 0.13	1.19 ± 0.31
C	0.25 ± 0.11	0.58 ± 0.27	0.63 ± 0.12	1.15 ± 0.26
Al1	0.22 ± 0.09	0.54 ± 0.21	0.60 ± 0.13	1.15 ± 0.30
Al2	0.25 ± 0.08	0.58 ± 0.18	0.61 ± 0.13	1.15 ± 0.29
Al3	0.24 ± 0.07	0.61 ± 0.21	0.57 ± 0.16	1.09 ± 0.38
Al4	0.23 ± 0.12	0.53 ± 0.25	0.60 ± 0.16	1.12 ± 0.29
Av.	**0.24 ± 0.09**	**0.57 ± 0.22**	**0.61 ± 0.14**	**1.14 ± 0.31**
		Textured rubber mat		
Ti	0.44 ± 0.08	0.92 ± 0.17	0.76 ± 0.17	1.28 ± 0.30
C	0.44 ± 0.10	0.91 ± 0.20	0.74 ± 0.12	1.24 ± 0.24
Al1	0.40 ± 0.11	0.84 ± 0.21	0.72 ± 0.12	1.21 ± 0.23
Al2	0.48 ± 0.11	0.98 ± 0.19	0.79 ± 0.13	1.31 ± 0.21
Al3	0.46 ± 0.10	1.00 ± 0.22	0.77 ± 0.19	1.31 ± 0.34
Al4	0.50 ± 0.12	1.01 ± 0.24	0.79 ± 0.14	1.32 ± 0.26
Av.	**0.46 ± 0.11**	**0.94 ± 0.22**	**0.76 ± 0.15**	**1.28 ± 0.27**
		Obstacle		
Ti	0.78 ± 0.11	1.43 ± 0.22	1.02 ± 0.11	1.55 ± 0.18
C	0.86 ± 0.13	1.50 ± 0.15	1.11 ± 0.19	1.60 ± 0.17
Al1	0.79 ± 0.17	1.41 ± 0.26	1.04 ± 0.16	1.52 ± 0.23
Al2	0.79 ± 0.10	1.52 ± 0.21	1.04 ± 0.15	1.61 ± 0.21
Al3	0.87 ± 0.16	1.57 ± 0.35	1.12 ± 0.20	1.66 ± 0.34
Al4	0.86 ± 0.16	1.56 ± 0.35	1.08 ± 0.15	1.64 ± 0.32
Av.	**0.83 ± 0.15**	**1.50 ± 0.28**	**1.07 ± 0.17**	**1.60 ± 0.26**

seat differed between surfaces, we observed no significant differences between wheelchairs in any of the four vibration parameters shown in Table 3. This can be explained partly by the overall good quality of the chosen wheelchairs and by the variation of velocity between trials (see Section 4.4).

We however observed significant differences of vibration transmissibility between wheelchairs (Table 4). In these cases, we believe that the variability of the measurements at the seat was compensated by the same variability at the inputs. It is however impossible to tell how much VT has a real impact on the health risk of the user.

We also observed in Table 4 that VT was often higher than 100% on the smooth vinyl floor, which means that measured vibration at the seat was greater than input vibrations. We believe that, in this particular case with low induced vibration, the seat compression and decompression due to wheelchair propulsion, which is itself a low-frequency vibration, may be nonnegligible compared to input vibrations.

Figure 5 compares the average total (x-z) vibration between the three tested surfaces and the real-life conditions measured by Garcia-Mendez et al. [9]. Continuous vibration is similar to real-life conditions, but shock-induced vibration is much lower in our study. One explication is that shock-induced vibration (vibration dose value (VDV)) is a cumulative measure that always grows during a measurement period [27]. Therefore, as their measurement period was an entire day while ours was 4 seconds, this explains why they measured higher VDVs. Additionally, the placement of the accelerometer was different between both studies. Garcia-Mendez et al. [9] placed the accelerometer below the seat; therefore, the vibration absorption by the seat was not taken into account in their measure.

Figure 6 compares the vertical (z) continuous vibration between the three tested surfaces and nine concrete and brick surfaces tested by Wolf et al. [10]. Shock-induced vibration values were not presented in Wolf et al. [10], but continuous vibration values are comparable between both studies. As for Garcia-Mendez et al. [9], Wolf et al. [10] placed their accelerometer below the seat while we placed ours above the seat. Therefore, our data should normally be slightly lower than theirs due to the vibration absorption of the seat. This is however impossible to verify because traversed surfaces were different.

4.2. Mechanical Work. A high P value was obtained for both surfaces. This means that whereas Ti minimized the mechanical work (0.188 J·(kg·m)$^{-1}$ on the textured surface) compared to Al2 (0.219 J·(kg·m)$^{-1}$ on the textured surface), such differences are statistically nonsignificant and strong conclusions on WN-WPM could not be drawn.

The mechanical work to traverse different surfaces was measured by Cooper et al. [23], but as the total distance was not specified, their data cannot be compared with ours. Chesney and Axelson [17] also measured WPM for different

Table 4: Vibration transmissibility of the frame (%).

WC	T_{awz}	T_{VDV_z}	T_{awt}	T_{VDV_t}
		Smooth vinyl floor		
Ti	172.7 ± 9.0	163.9 ± 13.0	103.3 ± 11.5	99.3 ± 10.7
C	127.9 ± 26.7	126.4 ± 35.0	104.8 ± 8.3	101.4 ± 9.0
Al1	121.7 ± 18.2	115.3 ± 16.7	103.6 ± 8.3	100.3 ± 10.4
Al2	143.7 ± 15.4	138.7 ± 16.1	100.4 ± 8.8	96.9 ± 10.4
Al3	171.2 ± 26.7	175.2 ± 35.2	105.3 ± 11.0	101.7 ± 11.1
Al4	148.8 ± 19.8	140.8 ± 24.0	101.3 ± 10.8	98.0 ± 11.5
Av.	**146.4 ± 27.9****	**142.1 ± 32.3****	**103.1 ± 9.9**	**99.6 ± 10.7**
		Textured rubber mat		
Ti	36.0 ± 5.4	36.8 ± 5.7	53.9 ± 6.9	49.7 ± 7.9
C	30.1 ± 6.1	30.5 ± 6.2	46.2 ± 7.6	41.1 ± 8.0
Al1	35.6 ± 7.3	35.6 ± 6.9	55.7 ± 5.8	50.4 ± 6.7
Al2	32.5 ± 5.7	32.6 ± 5.2	48.5 ± 5.2	42.7 ± 5.2
Al3	45.3 ± 8.0	46.9 ± 9.8	63.6 ± 11.3	59.1 ± 13.6
Al4	33.2 ± 7.3	33.7 ± 7.1	47.8 ± 5.6	43.1 ± 5.3
Av.	**35.5 ± 8.3****	**36.1 ± 8.8****	**52.6 ± 9.6****	**47.7 ± 10.4****
		Obstacle		
Ti	60.3 ± 9.9	47.1 ± 8.8	68.3 ± 7.3	49.8 ± 7.4
C	54.7 ± 5.7	42.4 ± 6.2	62.2 ± 4.1	44.5 ± 5.1
Al1	53.7 ± 8.3	43.4 ± 8.1	62.3 ± 5.3	46.1 ± 6.5
Al2	61.0 ± 7.1	48.3 ± 6.4	67.7 ± 5.3	50.2 ± 5.7
Al3	62.3 ± 8.3	49.8 ± 8.4	69.5 ± 4.8	51.9 ± 6.9
Al4	61.9 ± 6.8	47.5 ± 5.3	68.3 ± 4.0	49.4 ± 4.6
Av.	**59.0 ± 8.5**	**46.4 ± 7.8**	**66.4 ± 6.0***	**48.7 ± 6.6**

$^{*}P < 0.05$, $^{**}P < 0.01$.

Table 5: Weight-normalized work-per-meter (J·(kg·m)$^{-1}$).

WC	Smooth floor	Textured mat
Ti	0.155 ± 0.020	0.188 ± 0.020
C	0.162 ± 0.024	0.196 ± 0.025
Al1	0.156 ± 0.023	0.193 ± 0.021
Al2	0.158 ± 0.049	0.219 ± 0.024
Al3	0.155 ± 0.014	0.196 ± 0.019
Al4	0.153 ± 0.027	0.196 ± 0.028
Av.	**0.157 ± 0.003**	**0.198 ± 0.011**

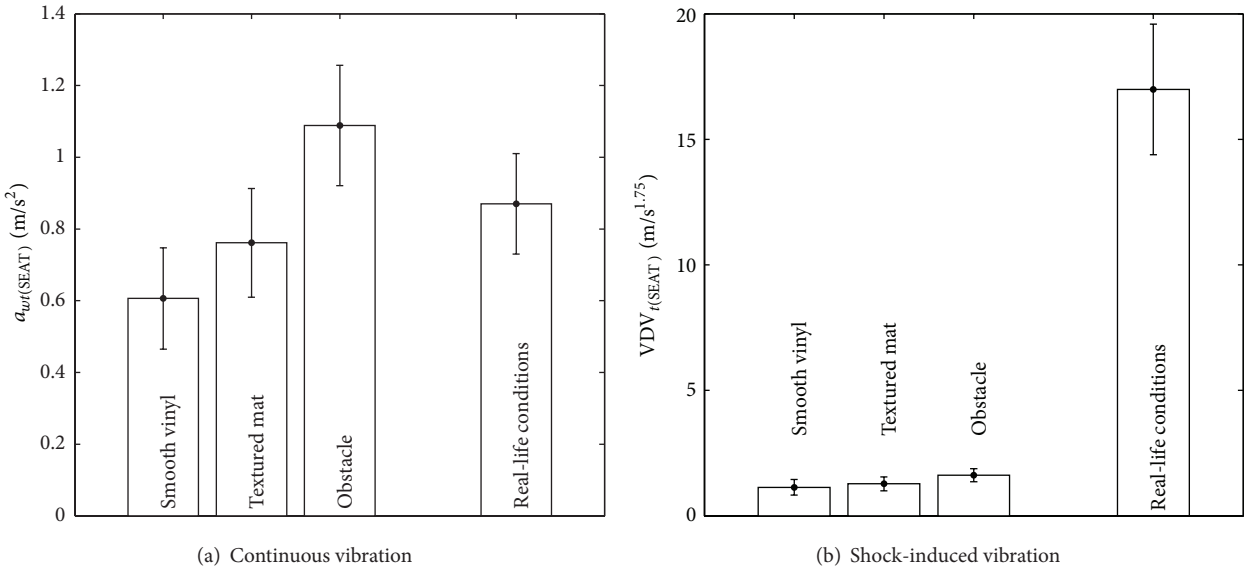
surfaces. Some of these surfaces are compared with ours in Figure 7. We observe that the work was always higher in their study; however, most of their comparable surfaces were on a 2% grade ramp, which necessarily needs more work to traverse because part of the work generated by the user is stored as potential energy gains.

4.3. Relation between WN-WPM and VT. In Figure 4, we observed that Al2 and Al3 stood out both in terms of WN-WPM and VT from the other wheelchairs. As all wheelchairs were equally configured, their different behaviour could be due to a combination of weight difference (they were the two heaviest tested wheelchairs) or due to frame design difference. When these wheelchairs were removed, the correlation between VT and WN-WPM rose to a range of −0.68 to −0.83. This quite high negative correlation supports our hypothesis that a wheelchair that transmits less vibration requires more mechanical work to traverse the same distance. This result could not be compared with actual literature because vibration and mechanical work were never assessed simultaneously and compared between wheelchairs before.

The titanium wheelchair (Ti) was not found to absorb vibration better than aluminium wheelchairs, which contradicts our hypothesis and current belief but concords with Kwarciak et al. [15] and Cochran [16]. However, it was the wheelchair that demanded the least mechanical work. As only one titanium wheelchair was tested, more wheelchairs should be tested before generalizing this observation to titanium wheelchairs in general.

This is the first time a wheelchair made of carbon fibre was tested for its vibration transmissibility. The carbon wheelchair (C) had the lowest VT, and there was no added work related to this improvement. A comparison between C and Al1, which feature very similar geometry, supports the role of frame material in this lower vibration transmissibility. Whereas only one carbon wheelchair was tested, this encouraging result means that carbon wheelchairs should be studied more thoroughly for the aspects of vibration transmissibility and mechanical work.

We observed in all cases of Figure 4 that the span of Al1, Al2, Al3, and Al4 (same material, different folding design) over WN-WPM and VT was always at least equal to the span



(a) Continuous vibration

(b) Shock-induced vibration

Figure 5: Comparing our total (x-z) vibration measurements to on-the-field data from Garcia-Mendez et al. [9].

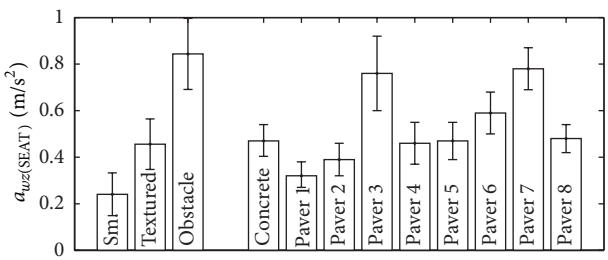


Figure 6: Comparing our vertical continuous vibration measurements to concrete/brick surfaces from Wolf et al. [10].

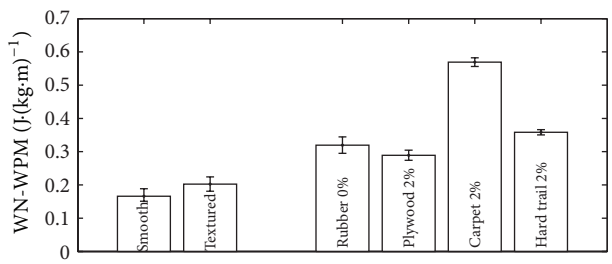


Figure 7: Comparing our weight-normalized work-per-meter measurements to selected surfaces from Chesney and Axelson [17]. Percentage indicates the ascending ramp grade.

of Ti, C, and Al1 (same folding design, different materials). As all wheelchairs were similar in dimensions, adjustments, and weight, this suggests that the folding design may be as important as the frame material when optimizing mechanical work and vibration transmissibility. Additional research on different frame design should be envisaged to confirm this observation.

4.4. Study Limitations. The following limitations were identified in this study.

4.4.1. Speed Control. Although wheeling velocity was controlled at 1 m/s during the familiarization, it was not controlled subsequently. Therefore, it varied slightly between trials, with an overall average and standard deviation of 1.00 ± 0.31 m/s. As wheeling velocity does have an effect on vibration [28], this variation may have altered the reproducibility between trials. Wolf et al. [29] controlled time to complete a trial at ±0.5%, while others did not control velocities at all [24, 25, 30]. For future work, we believe that velocity should be controlled during all trials.

4.4.2. Placement of the Seat Accelerometer. We initially chose to place the seat accelerometer in a cavity above the seat so that it was directly coupled with the user. We believe this is the best placement to measure the vibration transmitted to the user. However, when the outcome measure is the frame transmissibility, it would be more advisable to place the accelerometer below the seat, as did Garcia-Mendez et al. [9] and Wolf et al. [10], so that seat cushion absorption is not measured. For future work on vibration transmission, we advise placing an additional accelerometer below the seat. By using two seat accelerometers (one above, one below), it will be possible to isolate the frame and seat vibration transmissibility.

4.4.3. Mechanical Work. The mechanical work was measured only in steady state. Whereas the start-up work (mechanical work required to initiate the movement) was not measured, we believe this value would be of great interest. In fact, higher propulsion moments are required on start-ups than on steady state because of the additional inertial forces caused by the weight of the subject and wheelchair. Therefore, lighter wheelchairs may require less work to initiate movement than heavier ones.

4.4.4. Population. This work was done with able-bodied subjects who had not driven a wheelchair before. As the centre

of mass and wheeling technique differ between wheelchair users and nonusers, future studies should also be done with wheelchair users.

5. Conclusion

We compared the vibration transmissibility (VT) and the weight-normalized work-per-meter (WN-WPM) of six folding wheelchairs propelled by ten able-bodied users on three ground surfaces. We found significant differences in VT ($P < 0.05$) between wheelchairs, but not in WN-WPM ($P < 0.1$). With both parameters considered at the same time, Ti, Al1, Al4, and C performed better than Al2 and Al3. A negative correlation between vibration transmissibility and mechanical work was found, which supports our hypothesis that a wheelchair that transmits less vibration requires more mechanical work. More wheelchairs should be tested to confirm this correlation. Based on our results, use of carbon in wheelchair construction seems promising to reduce VT without increasing WN-WPM. On the other hand, the titanium wheelchair did not have a lower VT than aluminium wheelchairs, which is in contrast with current belief. As only one carbon wheelchair and one titanium wheelchair were tested, more research on wheelchairs made of these materials should be considered to confirm these observations. For future studies, we recommend giving special attention to wheeling velocity control, placing accelerometers above and below the seat to isolate frame and seat vibration transmissibility, and including start-up work as an additional analysis of mechanical work.

Appendices

These appendices put into perspective some aspects of ISO 2631-1:1997 applied to the wheelchair propulsion. It must be emphasized that the norm was developed for the able-bodied population, and its application to wheelchair propulsion is still debated.

A. Frequency Weighting

The recommended frequency weighting is originally based on discomfort contours obtained from persons subjected to accelerations in different axes and different frequencies [31]. This subjective method based on comfort may seem a serious issue for its application to the SCI population. However, the resulting frequencies are coherent with the resonance frequencies of the seated human body [27, 32]. Therefore, literature on wheelchair vibration tends to agree with this frequency weighting [9, 10, 15, 24]. Some authors prefer to completely avoid it [21].

B. Axis Selection and Weighting

As introduced in Section 2.3, literature does not agree on the axis to use to assess whole-body vibration when propelling a wheelchair. Wolf et al. [10] considered the vertical (z) axis only, Garcia-Mendez et al. [9] considered the total (x-z) vibration, and Cooper et al. [23] considered the total (x-y-z) vibration. A direct application of ISO 2631-1:1997 gives reason to both first and second authors since no quantitative measure is available to tell if two axes are comparable or not. The standard advises using the total (x-y-z) vibration only to assess comfort.

Having said that, there is a fundamental difference between x and z vibration in wheelchair propulsion: the latter is induced to the user by ground irregularities, whereas the former is mostly due to the propulsion motion and is voluntarily induced by the user [30]. Figure 3(b) offers a good visual representation of these vibration disparities between both axes. As ISO 2631-1:1997 is based on passive vibration and not on user induced vibration, we believe that assessing the vibration only the vertical (z) axis is more advisable than combining z with other axes.

C. Health Considerations

Appendix B of ISO 2631-1:1997 indicates vibration thresholds that quantify the risk of developing health issues due to whole-body vibrations. This can be a starting point to evaluate the maximal allowable vibration for wheelchair propulsion [9], but no validation of the health risks to the SCI population is available at the moment. Therefore, in our work, we chose to avoid comparing the vibration exposure of our subjects to these thresholds and used the standard only to compare wheelchairs among them.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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Research Article

Detailed Shoulder MRI Findings in Manual Wheelchair Users with Shoulder Pain

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Shoulder pain and pathology are common in manual wheelchair (MWC) users with paraplegia, and the biomechanical mechanism of injury is largely unknown. Establishing patterns of MRI characteristics in MWC users would help advance understanding of the mechanical etiology of rotator cuff disease, thus improving the logic for prescribed interventions. The purpose of this study was to report detailed shoulder MRI findings in a sample of 10 MWC users with anterolateral shoulder pain. The imaging assessments were performed using our standardized MRI Assessment of the Shoulder (MAS) guide. The tendon most commonly torn was the supraspinatus at the insertion site in the anterior portion in either the intrasubstance or articular region. Additionally, widespread tendinopathy, CA ligament thickening, subacromial bursitis, labral tears, and AC joint degenerative arthrosis and edema were common. Further reporting of detailed shoulder imaging findings is needed to confirm patterns of tears in MWC users regarding probable tendon tear zone, region, and portion. This investigation was a small sample observational study and did not yield data that can define patterns of pathology. However, synthesis of detailed findings from multiple studies could define patterns of pathological MRI findings allowing for associations of imaging findings to risk factors including specific activities.

1. Introduction

Shoulder pain in manual wheelchair (MWC) users with paraplegia is common. The prevalence is reported in the literature to range from 30% to 75%, and shoulder pain is associated with increasing years of WC use [1–3]. Further, 63% of long-term MWC users will have a rotator cuff tear compared to 15% of a matched able-bodied group [4]. The exact biomechanical mechanism and its contribution to the increased risk of shoulder pain development in this upper extremity dependent population are largely unknown. It is theorized that mechanical impingement is responsible for the reported rotator cuff disease seen with imaging and concomitant pain [5, 6]. However, current concepts in rotator cuff disease are evolving. Subtypes of impingement have been identified including subacromial or external impingement, internal impingement, and subcoracoid impingement, each with a different mechanical etiology and treatment implication [7]. Establishing patterns of magnetic resonance imaging (MRI) characteristics in different patient populations would help advance understanding of the mechanical etiology of rotator cuff disease, thus improving the logic for prescribed interventions.

Previous investigations of MRI findings in MWC users have revealed a high prevalence of rotator cuff tears, coracoacromial (CA) ligament thickening and edema, and acromioclavicular (AC) degenerative joint disease (DJD) [4, 8–10]. Common to all previous studies is the reporting of the presence of tears and pathology in the AC joint and CA ligament; however, results were not expanded to include

the specific location of the tear or degree of tendinopathy present. Further, the morphology of the acromion and pathology related to the bursa, labrum, or glenohumeral (GH) joints have not been described in MWC users.

Defining patterns of shoulder pathology on MRI in the MWC user population has not been possible thus far due to the paucity of studies and the limited reporting of findings. Further, the sample sizes are often small and have heterogeneity of confounding factors such as exposure to overhead sports. In order to accommodate for small heterogeneous sample sizes, consideration should be given to expanding imaging findings that are reported by including details such as tear size and location, severity of tendinopathy, joint degeneration, and condition of the labrum whenever MRI findings are reported. The synthesis of detailed imaging findings from multiple sources along with the reporting of additional risk factors will aid in the establishment of patterns of MRI characteristics in the MWC user population. The purpose of this study was to report detailed shoulder MRI findings in a sample of 10 MWC users with anterolateral shoulder pain that were participating in a shoulder pain rehabilitation intervention.

2. Methods

2.1. Study Participants. Participants who use manual wheelchairs ($n = 10$, 9 males, 1 female) were recruited as a sample of convenience from area clinics and organizations that provide treatment and services for individuals with disabilities. Participants in the current study were originally recruited for two larger studies investigating (1) shoulder rehabilitation and (2) daily exposure to shoulder elevation and upper extremity loading. A subsample of participants was included in the current MRI study based on participant availability and willingness to undergo an MRI. As such, there was no a priori power calculation performed for the MRI substudy. The Mayo Clinic Institutional Review Board approved the study protocol and informed consent was obtained from all participants before initiating test procedures.

Participants were eligible for the study if they were between 18 and 65 years of age, had a spinal cord injury, used a manual wheelchair as their primary means of mobility for a minimum of one year, were able to perform transfers and sit independently, and had shoulder pain with a date of onset no sooner than 2 weeks from the date of consent. Participants were excluded if they had cognitive impairments that limited their ability to independently follow instructions, if they had significant traumatic injury to the shoulder in which preinjury status was not attained, or if they were unwilling to have an MRI of one shoulder. The aforementioned criteria were screened over the phone. A licensed physical therapist (Meegan G. Van Straaten) performed a physical examination for additional screening. If shoulder pain was deemed to be of nonshoulder etiology (i.e., cervical spine), or if there was a presence of adhesive capsulitis (loss of greater than 25% of range of motion) or gross instability, participants were excluded from the study. Participants were required to have pain as indicated by one of the following: positive Neer, Hawkins-Kennedy or Empty Can sign, pain with humeral elevation, or complaints of anterior lateral shoulder pain during activity [11–13]. Self-reported shoulder pain was measured with the Wheelchair User's Shoulder Pain Index (WUSPI) [14]. Prior to the MRI, patients were required to meet Mayo Clinic requirements for MRI compatibility including specific screening for implanted metal within the body. Detailed demographic data was recorded including information on job type and recreational activities for the purposes of identifying participation in overhead work or competitive sports.

2.2. Magnetic Resonance Imaging of the Shoulder. The most painful shoulder was imaged using a standard clinical non-contrast imaging protocol. All participants were imaged on a GE (Milwaukee, WI) Signa HD MRI scanner with magnet field strength of 3 Tesla using a dedicated, commercially available, shoulder coil. The participants were positioned supine with the arm and wrist in a neutral position by the participant's side. Axial, oblique sagittal, and oblique coronal proton density (TR 3200–3500; TE 32-33) and T2 FSE FS (TR 4000; TE 50) image sequences were obtained with a matrix acquisition range of 256–384 × 256, 2 NEX, and 14 cm FOV.

2.3. Assessment of MR Images. The images were assessed by a board-certified, fellowship-trained, musculoskeletal radiologist (Naveen S. Murthy) using a standardized MRI Assessment of the Shoulder (MAS) guide developed by the study authors (see Supplementary Material available online at http://dx.doi.org/10.1155/2014/769649). The reliability, validity, and sensitivity of the MAS guide were not assessed in this study. The radiologist was blinded to the identity and symptoms of the participants throughout the study. Tears of the supraspinatus, infraspinatus, subscapularis, and teres minor muscles were classified as partial, full, or complete, located in one of three anatomical zones: insertion, tendon, or critical zone (myotendinous junction). The region of the tear was classified as intrasubstance, bursal, or articular. For the supraspinatus, infraspinatus, and teres minor, the portion of the tendon was defined as anterior, middle, or posterior. The subscapularis portions were defined as superior, middle, or inferior. A long head of the biceps tear was categorized as partial, split, or complete with locations defined as extra-articular, intra-articular, or bicep anchor. Tendinopathy was classified as mild, moderate, or severe [15]. AC joint degenerative arthrosis was categorized as mild, moderate, or severe, and subacromial spurs and subchondral edema/cystic changes were recorded as present (+) or absent (−). The acromion was classified according to type (1, 2, 3) and whether a lateral downslope or convex undersurface was present. CA ligament thickening was categorized as present (+) or absent (−). The presence (+) or absence (−) of bursitis and location were noted as subacromial or subcoracoid. The presence (+) or absence (−) of labral irregularities, tears, and paralabral ganglion cysts were noted. Labral tear location was designated using quadrant terminology (i.e., anterior/superior; posterior/superior; anterior/inferior; posterior/inferior). Degenerative arthrosis of the GH joint was

Table 1: Participant demographics, athletic involvement, and WUSPI scores.

Subject	Age	Injury level	Years of WC use	Current athletics	Previous athletics	Overhead occupation	WUSPI	Imaged side and dominancy
1	26	T9	9.4	Overhead sports, throwing			78.9	R, D
2	25	T12	21.5	Overhead sports, throwing, weight-lifting			62	R, D
3	48	T4	16.1	Endurance, ski/skate			4.9	R, D
4	45	T12	3.7	Outdoorsman, endurance, ski/skate		Yes—high force	45.1	R, D
5	32	C 6/7	6.5	Overhead sports			21.7	L, ND
6	35	T12	33.2	Overhead sports			1.2	R, D
7	33	T10	18.8	Endurance, ski/skate, overhead sports			23.9	R, D
8	32	T5	8.9	Outdoorsman, weight-lifting	Contact sport		19	R, ND
9	57	T10	3.6	Outdoorsman	Contact sport	Yes—high force	17.1	R, D
10	59	T12	23.1	Overhead sports, endurance, outdoorsman		Yes—low force	57.4	R, D

WUSPI = Wheelchair User's Pain Index.
R = right, L = left, D = dominant, ND = nondominant.

designated as mild, moderate, or severe. The subchondral edema/cystic changes of the GH joint were noted and chondromalacia was noted as mild, moderate, or severe.

2.4. Descriptive Data. This investigation was an observational study. Imaging findings are presented for each participant and the number of participants with each specific finding is reported.

3. Results

Our participants ranged in age from 24 to 58 years (mean 38 years) with levels of spinal cord injury from C 6-7 to T12 (Table 1). One of the 10 participants was a woman, and 7 currently practiced or competed in organized sports (such as wheelchair basketball) at least once a week. The participants were MWC users for an average of 14.5 ± 9.7 years and their WUSPI pain scores were 33.1 ± 26.1 (Table 1).

There were a total of 10 rotator cuff tears (partial thickness = 6, full thickness = 1, and complete = 3), across the supraspinatus (n = 5), infraspinatus (n = 2), and subscapularis (n = 3). Additionally, there were four tears of the long head of the biceps tendon (complete = 1 and split = 3) (Table 2). In the supraspinatus, five of 10 participants had partial tears at the insertion in either the intrasubstance or articular regions. Three tears were in the anterior portion and one was in the middle portion. One participant had a complete insertion zone tear, and nine of 10 participants had tendinopathy (mild = 3, moderate = 4, and severe = 2). Two participants had tears in the infraspinatus tendon at the insertion zone, one partial (intrasubstance, posterior portion) and one complete. Mild and moderate infraspinatus tendinopathy was observed in nine participants (mild = 5, moderate = 4). In the subscapularis, three participants had tears: one partial tear at the insertion (intrasubstance, superior portion), one full thickness tear in the tendon (superior portion), and one complete tear at the insertion. Tendinopathy in the subscapularis was present in all 10 participants (mild = 4, moderate = 5, and severe = 1). In the long head of the biceps, three of the 10 participants had intra-articular split tears and one participant had a complete tear at the insertion. Nine participants had tendinopathy in the long head of the biceps (mild = 4, moderate = 4, and severe = 1).

All participants had AC joint degenerative arthrosis (mild = 4, moderate = 4, severe = 2) (Table 3). Five participants had AC joint spurs and 8 out of 10 participants had AC joint edema. Five participants had a type 1 acromion and four participants had a type 2 acromion. Five out of 10 participants had a downsloped acromion and one participant had a convex undersurface. CA ligament thickening was present in seven of 10 participants. Subacromial bursitis was observed in nine of the 10 participants. Three participants had labral irregularities and six had labral tears, predominantly in the posterior/superior and posterior/inferior quadrants. Paralabral ganglion cysts were present in four participants. Four participants had degenerative arthrosis of the GH joint (mild = 3, moderate = 1) and chondromalacia (mild = 3, moderate = 1). One participant had GH joint edema.

4. Discussion

The purpose of this study was to document expanded MRI findings of shoulder pathology in MWC users with anterior-lateral shoulder symptoms. Noncontrast clinical shoulder MR images were performed on a sample of 10 MWC users with SCI (average years of MWC use 14 years). The MRI scans demonstrated tendon tears in 70% of examined shoulders, and 90% of the rotator cuff tears occurred at the tendon insertion site. The condition of the long head of the biceps and the labrum, the presence and severity of tendinopathy,

Table 2: Detailed MRI findings of the rotator cuff and long head of the biceps tendons.

Subject	Supraspinatus			Infraspinatus			Subscapularis			Biceps		
	Tear	Location	Tendop	Tear	Location	Tendop	Tear	Location	Tendop	Tear	Location	Tendop
1			Mod			Mild			Mod			Mod
2			Mild			Mild			Mod			Mild
3			Mod			Mild			Mod			Mod
4	P	Insert, Artic, Ant	Severe			Mod			Mild			Severe
5			Mild	P	Insert, Intra, Ant	Mod			mild			Mild
6			Mild			Mod			Mild	S	Intra	Mod
7	P	Insert, Artic, Ant	Mod			Mild			Mild			Mild
8	P	Insert, Intra, Ant	Mod			Mild	P	Insert, Intra, Sup	Mod	S	Intra	Mild
9	P	Insert, Intra, Ant	Severe			Mod	F	Tendon, Sup	Mod	S	Intra, Anchor	Mod
10	C	Insert		C	Insert		C	Insert	Severe	C	Insert	

Tears of the supraspinatus, infraspinatus, subscapularis, and teres minor muscles were classified as partial (P), full (F), or complete (C) located in one of three anatomical zones: insertion (Insert), tendon, or critical (Crit) zone. The region of the tear was classified as intrasubstance (Intra), bursal (Burs), or articular (Artic). For the supraspinatus, infraspinatus, and teres minor, the portion of the tendon was defined as anterior (Ant), middle (Mid), or posterior (Post). The subscapularis portions were defined as superior (Sup), middle, or inferior. A long head of the biceps tear was categorized as partial, split (S), or complete (C) with locations defined as extra-articular, intra-articular (Intra), or bicep anchor (Anchor). Tendinopathy was classified as mild, moderate (Mod), or severe.

Table 3: Detailed MRI findings of the acromioclavicular (AC) joint, acromion, coracoacromial (CA) ligament, bursa, labrum, and glenohumeral (GH) joint.

Subject	AC Joint			Acromion			CA Ligament	Bursitis		Labrum				GH Joint		
	Arthros	Spurs	Edema	Type	Downslope	Convex	Thickening	Yes/No	Location	Irreg	Tear	Location	Ganglion	Degen Arthrosis	Edema	Chond
1	Mod	−	+	2	−	−	−	+	Subacr	−	+	Postsup	+		−	
2	Mild	−	−	1	+	−	+	+	Subacr	−	−		−		−	
3	Mod	+	+	1	+	−	+	+	Subacr	−	−		−	Mild	−	Mild
4	Severe	+	+	1	+	+	+	+	Subacr	+	+	Postsup	−	Mild	−	Mild
5	Mild	−	+	1	+	−	+	+	Subacr	−	−		−		−	
6	Mild	−	−	1	−	−	−	+	Subacr	−	+	Postinf	+		−	
7	Mod	+	+	1	+	−	+	+	Subacr	−	+	Antsup, Postsup	−		−	
8	Mild	−	+	2	−	−	−	+	Subacr	−	−		−		−	
9	Mod	+	+	2	−	−	+	+	Subacr	+	+	Postsup	+	Mod	+	Mod
10	Severe	+	+	2	−	−	+	−		+	+	Postinf	+	Mild	−	Mild

AC joint degenerative arthrosis was categorized as mild, moderate (Mod), or severe, and subacromial spurs and subchondral edema/cystic changes were recorded as present (+) or absent (−). The acromion was classified according to type (1, 2, 3) and whether a lateral downslope or convex undersurface was present (+). CA ligament thickening was categorized as present (+) or absent (−). The presence (+) or absence (−) of bursitis and location was noted as subacromial or acronym (subacr) subcoracoid. The presence (+) or absence (−) of labral irregularities, tears, and paralabral ganglion cysts was noted. Labral tear location was designated with anterior/superior (Antsup), posterior/superior (Postsup), anterior/inferior, or posterior/inferior (Postinf). Degenerative arthrosis of the GH joint was designated as mild, moderate (Mod), or severe. The presence (+) or absence (−) of subchondral edema/cystic changes was noted and chondromalacia was noted as mild, moderate (Mod), or severe.

and the presence and location of bursitis were among the additional factors reported. Although this sample was not large enough to define MRI characteristics specific to MWC users, we have provided the MAS guide for other investigators to follow in evaluating and reporting detailed MRI findings for the purpose of data synthesis.

Our findings support the high prevalence of shoulder CA ligament thickening, AC joint edema, and AC joint degenerative arthrosis (Table 3) that has been previously reported in MWC users with SCI [8–10]. Additionally, 50% of our participants had tears of the supraspinatus tendon at its insertion, 60% had tears in at least one of the rotator cuff tendons, and 70% had tears when including the long head of the biceps tendon. These findings support the observation that 63% of MWC users after three decades of use will have rotator cuff tears predominantly in the supraspinatus [4]. Our study is the first to report detailed location of tears and findings of widespread tendinopathy ranging from mild to severe in this population. Additionally, subacromial bursitis and labral tears were common and have not been previously reported.

Our finding that supraspinatus tears were more common than tears in the infraspinatus, subscapularis or long head of the biceps tendon was similar to shoulder imaging findings from able-bodied adult and pediatric populations [16, 17]. This is interesting as it implies that while wheelchair use increases the frequency of rotator cuff symptoms and tearing, it may not be a different location or tear type from nonwheelchair users. Studies on able-bodied populations report more articular-side partial-thickness tears than bursal or interstitial tears [16]. Articular side tears are reported to be 2-3 times more common than bursal-sided tears, and intrasubstance tears are the least common [17, 18]. Our small sample differed in that we found more intrasubstance tears than articular-side tears (4 compared with 2) and no bursal-side tears were observed. The majority of our partial-thickness tears were at the insertion which is similar to the findings reported from a recent study of 201 able-bodied pediatric participants [16]. Tendon tears at the insertion site have been attributed to younger populations, and in the nonwheelchair adult population, the critical zone has been found to be the most common site of tears [19–21]. Our sample had an average age at injury of 24 years which may provide some explanation for the presence of insertion tears that are more common in younger and pediatric populations.

Eight of our 10 subjects are currently or were previously exposed to overhead sports or occupations. Of those seven, five had labral tears which is associated with overhead sports. Tears of the posterior supraspinatus or anterior infraspinatus, as a result of internal impingement during the cocking phase of a throw, have been documented in overhead athletes [22]. Two of our subjects who were exposed to overhead sports or occupations had a partial tear of the supraspinatus, both in the anterior portion of the tendon. More data is needed to determine whether shoulder pathology in overhead wheelchair athletes differs from overhead able bodied athletes, and whether shoulder pathology in overhead wheelchair athletes differs from MWC users who do not participate in overhead sports.

Mechanistic interpretation of these findings is limited due to the small sample size and heterogeneity with regard to factors such as age at first wheelchair use, years in chair, and exposure to overhead activity. Our sample was comprised of individuals who participate or have participated in sports, some of which are overhead. Without longitudinal imaging or comparison groups of more sedentary MWC users, we cannot conclude that the pathological findings are related to specific job or recreational activities. While the participants are described as having shoulder pain, two of the 10 have WUSPI scores below five. Since it has been documented that asymptomatic individuals can have positive MRI findings, it would have been beneficial to compare MRI results between participants' painful and nonpainful shoulders. One musculoskeletal radiologist reviewed all the images, and although he was masked to the symptoms and participant characteristics, the study would be more robust with multiple radiological reviewers. At the same time, however, this type of single-radiologist review is common in the clinical setting and is the most frequent way that imaging is used to diagnose patients. Cross-sectional studies such as ours cannot determine cause or etiology; therefore, longitudinal data is needed to truly advance the understanding of pathology development in this population. Future investigations should determine the reliability and validity of the MAS guide.

5. Conclusion

In our detailed reporting of shoulder MRI assessment on 10 MWC users with spinal cord injury, we observed that the tendon most commonly torn was the supraspinatus at the insertion site in the anterior portion in either the intrasubstance or articular region. Additionally, widespread tendinopathy, CA ligament thickening, subacromial bursitis, labral tears, and AC joint degenerative arthrosis and edema were common in our sample. Further reporting of detailed shoulder imaging findings is needed to confirm patterns of tears in MWC users regarding probable tendon tear zone, region, and portion. This investigation was a small sample observational study and did not yield data that can define patterns of pathology. However, synthesis of detailed findings from multiple studies could define patterns of pathological MRI findings allowing for associations of imaging findings to risk factors including activities.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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Research Article

The Relationship between Independent Transfer Skills and Upper Limb Kinetics in Wheelchair Users

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Transfers are one of the most physically demanding wheelchair activities. The purpose of this study was to determine if using proper transfer skills as measured by the Transfer Assessment Instrument (TAI) is associated with reduced loading on the upper extremities. Twenty-three wheelchair users performed transfers to a level-height bench while a series of forces plates, load cells, and a motion capture system recorded the biomechanics of their natural transferring techniques. Their transfer skills were simultaneously evaluated by two study clinicians using the TAI. Logistic regression and multiple linear regression models were used to determine the relationships between TAI scores and the kinetic variables on both arms across all joints. The results showed that the TAI measured transfer skills were closely associated with the magnitude and timing of joint moments ($P < .02$, model R^2 values ranged from 0.27 to 0.79). Proper completion of the skills which targeted the trailing arm was associated with lower average resultant moments and rates of rise of resultant moments at the trailing shoulder and/or elbow. Some skills involving the leading side had the effect of increasing the magnitude or rate loading on the leading side. Knowledge of the kinetic outcomes associated with each skill may help users to achieve the best load-relieving effects for their upper extremities.

1. Introduction

In 2010 there were about 1.6 million people using wheelchairs for mobility [1]; with that number expanding each year [2]. Wheelchair users must use their upper extremities for almost all activities of daily living (ADLs) such as getting in and out of bed, transferring to a shower or toilet, and transferring in and out of a car [3]. A full-time wheelchair user will perform on average 14 to 18 transfers per day [4]. Transfers are a key element of living an active and productive life and play a vital role in maintaining independence of wheeled mobility device users. If wheelchair users cannot transfer freely, their quality of life and community participation will be severely affected [5].

Transfers are one of the most strenuous wheelchair activities performed [6] and nearly half of wheelchair users do not learn how to use proper transfer techniques during rehab [3]. Incorrect transfer skills may predispose wheelchair users to developing upper limb pain and overuse related injuries, such as rotator cuff tears, elbow pain, and carpal tunnel syndrome [7–11]. The onset of pain can lead to social isolation [5], dependence on others for assistance with ADLs, and increased medical expenditures [7]. Only about half of wheelchair users seek treatment for pain [6, 12, 13] and many feel that their symptoms were not improved after treatment [6, 13, 14]. Therefore, it seems that prevention may be crucial to reducing upper limb pain and overuse injuries. Learning to transfer in a way that reduces forces and awkward joint motions is an important strategy for preserving upper limb function [14].

During transfers, the shoulders often assumed a position of flexion, abduction, and internal rotation [4, 15]. This

position brings the glenohumeral head in closer alignment to the undersurface of the acromion and has been identified as a critical risk factor for impinging subacromial soft tissue [16]. Previous studies also indicate that the loading on the upper extremity joints during transfers is greater than any other wheelchair related activity [17]. Transfers have been associated with high peak posterior force and shoulder flexion and adductor moments at the shoulders [17–19]. Large posterior forces at the shoulder are thought to contribute to the development of shoulder posterior instability, capsulitis, and tendinitis [20]. The combination of shoulder posterior and superior forces increases the risk of shoulder impingement syndrome [21]. Furthermore, the elbow has been shown to sustain high superior forces during transfers which may cause nerve compression and result in secondary elbow injuries [19]. Extremes of wrist extension during transfers have also been reported which combined with the weight-bearing loads during transfers may exacerbate wrist injuries such as carpal tunnel syndrome [22, 23]. Using transfer techniques that reduce upper limb joint forces and moments may help prevent injuries [24–27].

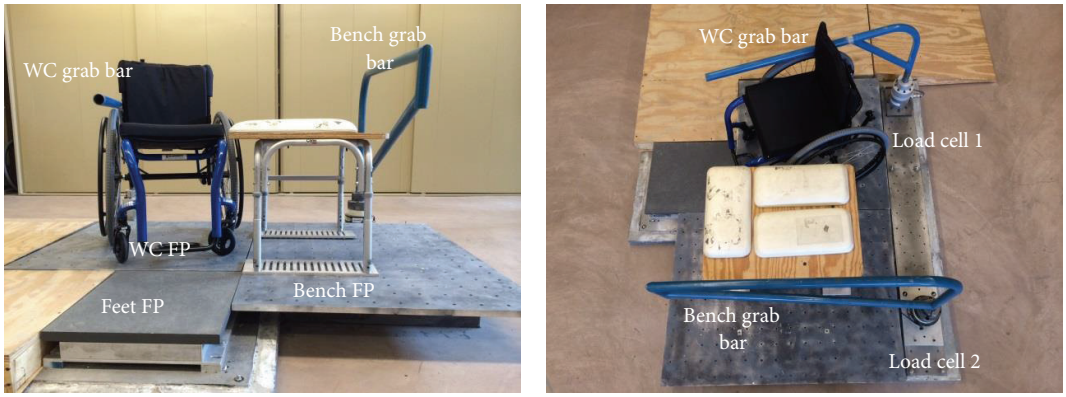
The current standard for evaluating transfer technique is observation by the therapist and a qualitative assessment. Transfer technique evaluations are not scientifically oriented and uniform across rehabilitation facilities [3, 28]. Results are impacted by the experience of the therapist and their idea of what constitutes a proper transfer, leading to less precise evaluations and a great degree of variability in transfer skills. The Transfer Assessment Instrument (TAI) is the first tool to standardize the way clinicians evaluate transfer technique and to help identify specific skills to target during transfer training. The items on the TAI were based on clinical practice guidelines [11], current knowledge in the literature [18], and best clinical practices related to transfers. The TAI has acceptable to high inter- and intrarater reliability (intraclass correlation coefficient (ICC) values ranging from 0.72 to 0.88) and good face, content, and construct validity [29–31]. However, no study has associated a clinical assessment of transfer skills to biomechanical changes. The purpose of this study is to examine the relationship between transfer skills as measured with the TAI and upper limb joint loading and to determine if using proper transfer skills as defined by the TAI results in better biomechanical factors that prevent the upper limbs from getting secondary injuries. We hypothesize that better transfer skills (higher scores on the TAI) will correlate with lower magnitudes and rates of rise of forces and moments at the shoulders, elbows, and wrists. Knowledge on the relationship between TAI skills and joint biomechanics will lead to more effective transfer assessments and help to focus training on skills that protect the upper limbs for long term use.

2. Methods

2.1. Participants. The study was approved by the Department of Veterans Affairs Institutional Review Board. All testing occurred at the Human Engineering Research Laboratories in Pittsburgh, PA. The subjects participating in the study were required to be over 18 years old and one year after injury or diagnosis, use a wheelchair for the majority of mobility (40 hours/per week), and be unable to stand up without support. Individuals with pressure sores within the past year and history of angina or seizures were excluded.

2.2. Testing Protocol. After written informed consent was obtained, subjects completed a general demographic questionnaire. Anthropometric measures were collected, such as upper arm length and circumference, to determine the center of mass and moment of inertia for each segment [32]. Subjects were asked to position themselves next to a bench, which was at a height level to their own wheelchair seats, on a custom-built transfer station (Figure 1) [33]. The transfer station contains three force plates (Bertec Corporation, Columbus, OH) which were underneath the wheelchair, level bench, and the subject's feet, respectively. Two 6-component load cells (Model MC5 from AMTI, Watertown, MA; Model Omega 160 from ATI, Apex, NC) were attached to two steel beams used to simulate an armrest and grab bar (Figure 1). Subjects were asked to naturally position and secure their wheelchairs in the 3∗3 square foot (91.44 cm by 91.44 cm) aluminum platform that covered the wheelchair force plate. They were also asked to choose where they wanted to position and secure the bench on the other 3∗4 square foot aluminum platform (91.44 cm by 121.92 cm) that covered the bench force plate (Figure 1). The position of the wheelchair grab bar was also adjusted based on the subjects' preference. Reflective markers (Figure 2) were placed on subjects' heads, trunks, and upper extremities to build local coordinate systems [34] for each segment. Marker trajectories were collected at 100 Hz using a ten-camera three-dimensional motion capture system (Vicon, Centennial, CO.) Kinetic data from all the force plates and load cells were collected at 1000 Hz.

Subjects were asked to perform up to five trials of level-height bench transfers. In each trial, subjects needed to perform transfers to and from their own wheelchairs in a natural way. Movement from one surface to the other (e.g., wheelchair to bench) was considered as one transfer. They were provided an opportunity to adjust their wheelchair position and familiarize themselves with the setup prior to data collection. Subjects had time to rest in between trials and additional rest was provided as needed. They were asked to use their own approaches to transferring so their transfer movement pattern and techniques would be as natural as possible. Subjects were asked to place their trailing arm (right arm) on the wheelchair grab bar (Figure 1) when they transferred to the bench on their left side so the reaction forces at the hand could be recorded. On the bench side, subjects were free to place their hand on either the bench or the grab bar. During each trial, up to two study clinicians independently observed and scored each subject's transfer skills using the TAI. All of the participants in the study were evaluated by the same two clinicians. Both were physical therapists who were trained to use the TAI before the study started. The TAI was completed after watching participants perform three to five transfers from the wheelchair to the bench. After independently scoring each subject, the clinicians compared their findings. Any discrepancies in the



(a) (b)

Figure 1: Front (a) and top (b) views of the transfer station. WC: wheelchair; FP: force plate.

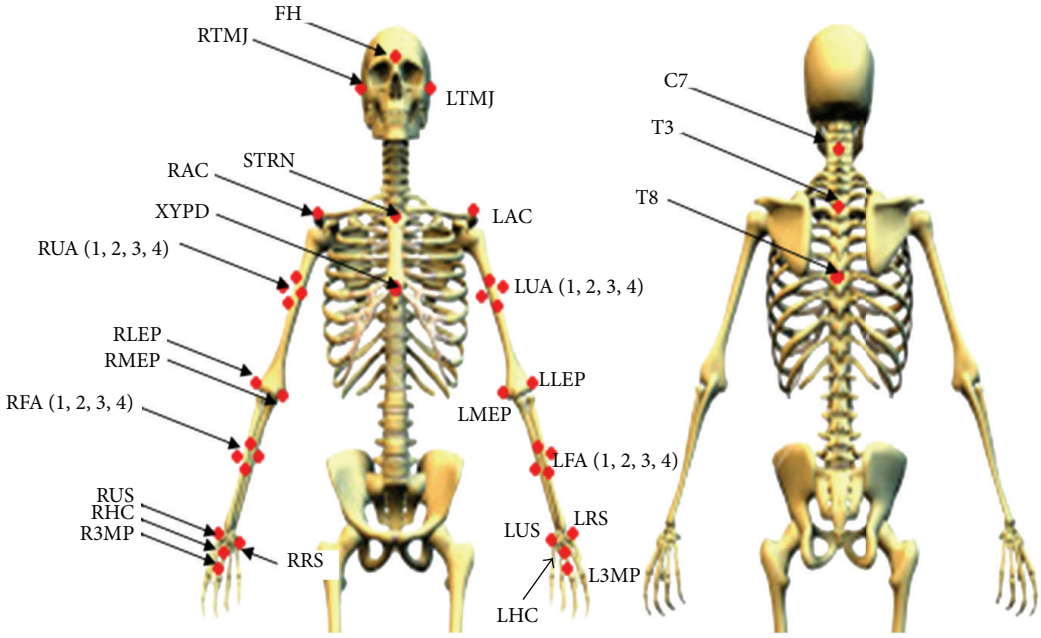


Figure 2: The marker set used in the current study. FH: forehead; RTMJ: right temporomandibular joint; LTMJ: left temporomandibular joint; STRN: sternum; RAC: right acromioclavicular joint; LAC: left acromioclavicular joint; XYPD: xiphoid; RUA: right upper arm; LUA: left upper arm; RLEP: right lateral epicondyle; LLEP: left lateral epicondyle; RMEP: right medial epicondyle; LMEP: left medial epicondyle; RFA: right forearm; LFA: left forearm; RUS: right ulnar styloid; LUS: left ulnar styloid; RRS: right radial styloid; LRS: left radial styloid; RHC: right hand center; LHC: left hand center; R3MCP: right 3rd metacarpophalangeal joint; L3MCP: left 3rd metacarpophalangeal joint; C7: 7th cervical spinous process; T3: 3rd thoracic spinous process; T8: 8th thoracic spinous process.

scoring were discussed and a score reflecting the consensus decision was recorded.

2.3. Data Analysis. The biomechanical variables were computed using MATLAB (Mathworks, Inc., Natick, MA, USA). A zero-lag low-pass 4th order Butterworth filter with cutoff frequency of 7 and 5 Hz was used to filter the kinetic and kinematic data, respectively [19]. Only the lift phase of the transfer from the wheelchair to the bench was analyzed in this study. A transfer was determined to begin when a vertical reaction force was detected by the load cell on the wheelchair side grab bar (Figure 1) and ended before a landing spike was detected by the force plate underneath the bench [35]. The end of the lift phase and the beginning of the descent phase are defined by the highest elevated point of the trunk which is indicated by the peak of the C7 and T3 marker trajectories [35]. Hanavan's model was used to calculate center of mass and moment of inertia using the subjects' segment lengths and circumferences [32]. Three-component forces and moments measured by the load cells and the force plates (Figure 1), the marker data of the trunk and upper extremities, and the inertial properties of each body segment were inputs into an inverse dynamic model [36]. Each segment was assumed as a rigid body and linked together by

ball and socket joints. The 3rd metacarpophalangeal joint was assumed as the point of force application. The output of the inverse dynamic model included upper extremity net joint forces and moments.

The key kinetic variables included average and maximum resultant forces and moments, and maximum rate of rise of resultant force and moment at the shoulders, elbows, and wrists on both sides. Since shoulder pain is more commonly associated with transfers [7], we analyzed the maximum superior and posterior shoulder forces and extension, abduction, and internal rotation shoulder moments. These variables were selected because they have been linked to shoulder pain, median nerve function, and other upper extremity injuries [18, 21, 24, 25, 37–39]. The resultant force on each joint is indicative of the total joint loading. The maximum rate of rise of resultant force is the peak instantaneous loading rate and impact force on each joint. The resultant moment on each joint represents the rotational demands associated with the muscle forces around the joint and the external forces. The maximum rate of rise of resultant moment indicates the peak rate of moment production on each joint. The superior and posterior shoulder forces were defined as the components of resultant shoulder force acting along the vertical upward and posterior axes of shoulder. Each kinetic variable was normalized by body mass (in kilogram) [17, 18, 40]. Descriptive statistics (means and standard deviations (SD)) were calculated for each variable. Kinetic variables were averaged over a minimum of three and a maximum of five trials.

The TAI contains two parts—parts 1 and 2. Both parts are scored and averaged to produce a third, final score. Only part 1 item scores and part 1 summary scores were used because the part 1 items evaluate whether the individual used specific transfer skills. Part 2 was not analyzed in this study as it encompasses some of the same transfer skills that are measured in part 1. Part 1 is comprised of 15 items which are scored "yes" (1 point) when the subject performs the specified skill correctly and "no" (0 points) when the subject performs the skill incorrectly or not applicable "(N/A)" which means the item does not apply. The part 1 summary score is the summation of each item's score multiplied by 10, and then divided by the number of applicable items, ranging from 0 to 10 [29]. TAI items that had a 50% response rate or higher in a N/A category or greater than an 80% response rate in the same non-NA category (e.g., yes or no) were not considered for further analysis on the individual item scores. Point-biserial correlations were conducted between the remaining items. Among the items that were highly correlated ($r > 0.80$), one was selected for the logistical modeling analysis (see below).

All of the kinetic data and TAI part 1 summary scores (e.g., continuous variables) were examined for normality using the Shapiro-Wilk test. Point-biserial correlation tests between each TAI item score (e.g., dichotomous variable) and the kinetic variables and Spearman's correlation tests between part 1 summary scores and kinetic variables were conducted to identify relationships with at least a medium effect size ($r \geq .30$ or $\leq -.30$ [41]). In order to verify specific kinetic effects of each transfer skill, logistic regression was used to model the association between individual TAI item scores (dichotomous outcome variable) and kinetic variables (predictors). Multiple linear regression was used to model the association between the TAI part 1 summary scores (continuous outcome variable) and kinetic variables (predictors). Separate models were created for the left and right sides. For the logistic regression model, histograms and Q-Q plots were used to check the assumption of no outliers. The assumption of multicollinearity for the kinetic variables (predictors) was tested using the variance inflation factors (VIFs) [42]. The assumption of linear relationships between continuous predictors and the log of the outcome variable was tested by Box-Cox transformation [43]. For the multiple linear regression models, histograms and Q-Q plots were used to check the assumption of no outliers on both predictors and outcome variables. The scatter plot of the standardized residuals against the predicted value was used to test the assumption of linearity. Shapiro-Wilk test was used to check the normality of the error term of the regression model. The assumption of multicollinearity for the predictors was also tested using the VIFs [42]. The assumptions of homoscedasticity and independence for multiple linear regression was checked using the Breusch-Pagan test [44] and Durbin-Watson test [45], respectively.

Backward elimination was used to determine the subset of predictors (kinetic variables) for each TAI outcome variable. The level of significance was set at $P < .05$. All the statistical analyses were performed in SPSS 21 (SPSS Inc., Chicago, IL).

Table 1: Participants' demographic information.

Subjects, $n = 23$	Mean ± standard deviation (range)
Age (years)	38.30 ± 11.07 (21–55)
Height (meters)	1.67 ± 0.23 (.99–1.85)
Weight (kilograms)	67.14 ± 19.18 (29.96–98.15)
Average duration of using a wheelchair (years)	13.15 ± 8.13 (1–27.25)

3. Results

3.1. Participants. Twenty men and three women volunteered to participate in the study. Table 1 shows summary demographic information. Eighteen subjects had a spinal cord injury (SCI); 14 subjects reported a complete SCI and four subjects an incomplete SCI (three with American Spinal Injury Association (ASIA) Grade B, one with ASIA Grade C). Three subjects had quadriplegia (C4 to C6), 9 had high paraplegia (T2 to T7), and six had low paraplegia (T8 to L3) [46]. The remaining five participants had bilateral tibial and fibular fractures with nerve damage ($n = 1$), double above knee amputation ($n = 1$), muscular dystrophy ($n = 1$), osteogenesis imperfecta ($n = 1$), and myelopathy ($n = 1$).

3.2. TAI Variables. Since the TAI part 1 summary scores and final scores were highly correlated ($r = .97$), the TAI part 1 summary scores were used for the multiple regression model. The part 1 summary scores ranged from 3.08 to 10.00 with an average (±SD) of 7.30 (±1.76). Table 2 shows the items in the

Table 2: The items in part 1 of the TAI.

Items in part 1 of the TAI
(1) The subject's wheelchair is within 3 inches of the object to which he is transferring on to.
(2) The angle between the subject's wheelchair and the surface to which he is transferring is approximately 20–45 degrees.
(3) The subject attempts to position his chair to perform the transfer forward of the rear wheel (i.e., subject does not transfer over the rear wheel).
(4) If possible, the subject removes his armrest or attempts to take it out of the way.
(5) The subject performs a level or downhill transfer, whenever possible.
(6) The subject places his feet in a stable position (on the floor if possible) before the transfer.
(7) The subject scoots to the front edge of the wheelchair seat before he transfers (i.e., moves his buttocks to the front 2/3rds of the seat).
(8) Hands are in a stable position prior to the start of the transfer.
(9) A handgrip is utilized correctly by the leading arm (when the handgrip is in the individual's base of support).
(10) A handgrip is utilized correctly by the trailing arm (when the handgrip is in the individual's base of support).
(11) Flight is well controlled.
(12) Head-hip relationship is used.
(13) The lead arm is correctly positioned. (The arm should not be extremely internally rotated and should be abducted 30–45 deg.)
(14) The landing phase of the transfer is smooth and well controlled (i.e., hands are not flying off the support surface and the subject is sitting safely on the target surface).
(15) If an assistant is helping, the assistant supports the subject's arms during the transfer.

part 1 of the TAI. Items 1, 2, 6, 7, 9, and 12 met the inclusion criteria for the logistic models (yes response rate ranges from 39% to 78%, $n = 23$). Items 4, 5, and 15 were not modeled because of the high number of N/A responses. Items 8, 10, 11, 13, and 14 were not modeled because they had too high of a "yes" response rate (e.g., greater than 80% of subjects). Items 3 and 7 scores had the same exact responses for both ($r = 1$). Item 7 scores were modeled because it can be applied to both manual and power wheelchair users, whereas item 3 only applies to manual wheelchair users.

3.3. Kinetic Variables. Means and standard deviations of the selected kinetic variables are shown in Table 3.

3.4. Correlation Test Results. The TAI part 1 summary and item scores were statistically associated and at least moderately correlated ($r \geq .3$ or $\leq -.3$) with one or more of the kinetic variables [41] (Table 4).

3.5. Logistic Regression Models for Item Scores. Lower average resultant shoulder force and higher maximum rate of rise of resultant shoulder moment on the leading (left) side were associated with a "yes" score on item 1 (Table 5). Subjects with lower maximum internal rotation shoulder moments on the leading (left) side had an increased likelihood of a "yes" score for item 2. Lower average resultant shoulder moment on the trailing (right) side and lower maximum rate of rise of resultant shoulder moment on the leading (left) side corresponded with a "yes" score on item 6.

On the trailing (right) side, subjects with lower average resultant moment and maximum rate of rise of resultant moment at the elbow were more likely to have a "yes" score on item 7. On the leading (left) side, a higher maximum shoulder extension moment was associated with a "yes" score on item 7.

On the trailing (right) side, a "yes" score on item 9 corresponded with lower average resultant shoulder moment and lower maximum rate of rise of resultant elbow moment.

Table 3: The mean (±standard deviation (SD)) of the kinetic variables normalized by body mass (kg).

Variables	Trailing (right) side Mean (±SD)	Leading (left) side Mean (±SD)
Shoulder		
AveRF (N/Kg)	2.98 (±0.75)	2.52 (±0.54)
MaxRF (N/Kg)	4.54 (±1.10)	4.24 (±0.97)
MaxRFRate (N/sec∗Kg)	15.95 (±6.09)	13.14 (±5.72)
AveRM (N∗m/Kg)	0.53 (±0.26)	0.60 (±0.17)
MaxRM (N∗m/Kg)	0.87 (±0.38)	1.06 (±0.25)
MaxRMRate (N∗m/sec∗Kg)	3.36 (±1.95)	3.96 (±1.38)
MaxSupF (N/Kg)	1.58 (±0.70)	2.18 (±1.14)
MaxPosF (N/Kg)	3.22 (±1.17)	3.23 (±0.95)
MaxIRM (N∗m/Kg)	0.10 (±0.11)	0.10 (±0.15)
MaxAbdM (N∗m/Kg)	0.43 (±0.21)	0.42 (±0.26)
MaxExtenM (N∗m/Kg)	0.41 (±0.30)	0.70 (±0.32)
Elbow		
AveRF (N/Kg)	2.76 (±0.71)	2.37 (±0.59)
MaxRF (N/Kg)	4.35 (±1.07)	4.20 (±1.03)
MaxRFRate (N/sec∗Kg)	16.06 (±6.00)	4.66 (±2.91)
AveRM (N∗m/Kg)	0.38 (±0.16)	0.21 (±0.10)
MaxRM (N∗m/Kg)	0.62 (±0.23)	0.39 (±0.15)
MaxRMRate (N∗m/sec∗Kg)	2.43 (±1.18)	1.85 (±0.89)
Wrist		
AveRF (N/Kg)	2.69 (±0.70)	2.34 (±0.61)
MaxRF (N/Kg)	4.29 (±1.05)	4.19 (±1.06)
MaxRFRate (N/sec∗Kg)	16.21 (±6.08)	13.17 (±5.74)
AveRM (N∗m/Kg)	0.22 (±0.06)	0.15 (±0.08)
MaxRM (N∗m/Kg)	0.35 (±0.09)	0.26 (±0.14)
MaxRMRate (N∗m/sec∗Kg)	1.34 (±0.57)	0.86 (±0.46)

Ave: average; Max: maximum; RF: resultant force; RFRate: rate of rise of resultant force; RM: resultant moment; RMRate: rate of rise of resultant moment; SupF: superior force; PosF: posterior force; IRM: internal rotation moment; AbdM: abduction moment; ExtenM: extension moment.

Table 4: Point-biserial correlation coefficients between TAI items and kinetic variables and Spearman's correlation coefficients between the part 1 summary scores and kinetic variables. The table shows the relationships that were statistically significant and had a medium effect size or larger: $r \geq .3$ or $\leq -.3$.

Correlations	Trailing (right) side							Leading (left) side						
	1	2	6	7	9	12	Part 1	1	2	6	7	9	12	Part 1
Shoulder														
AveRF				−.43*			−.35	−.30	−.31					
MaxRF									−.36					
MaxRFRate			−.31	−.32	−.54*					−.37				
AveRM	.30		−.52$		−.44*							−.34		
MaxRM	.31		−.47*		−.49*									
MaxRMRate	.37		−.51*		−.52*			.37		−.55*		−.39	−.46*	−.39
MaxSupF														
MaxPosF														
MaxIRM			.37						−.56*			.42*		
MaxAbdM						.33			−.32					
MaxExtenM						.31			.35	.30	.43*			.49*
Elbow														
AveRF	.30			−.44*	−.33		−.42							
MaxRF	.31								−.33					
MaxRFRate				−.33	−.57$					−.33				−.32
AveRM				−.61$	−.49*		−.39				−.34			
MaxRM				−.59$	−.54$		−.38				−.32			−.32
MaxRMRate				−.64$	−.62$		−.40			−.35	−.43*	−.30	−.36	−.52*
Wrist														
AveRF				−.44*	−.34		−.41							
MaxRF					−.30				−.32					
MaxRFRate			−.30	−.32	−.55$									
AveRM				−.50$			−.49*	−.31	−.35			.62$		
MaxRM				−.38			−.33	−.33	−.36			.64$		
MaxRMRate			−.35	−.31	−.46*							.36		

* $P < 0.05$; $ $P < 0.01$; Ave: average; Max: maximum; RF: resultant force; RFRate: rate of rise of resultant force; RM: resultant moment; RMRate: rate of rise of resultant moment; SupF: superior force; PosF: posterior fore; IRM: internal rotation moment; AbdM: abduction moment; ExtenM: extension moment.

On the leading (left) side, a "yes" score on item 9 was associated with lower maximum rate of rise of resultant shoulder moment, higher maximum internal rotation shoulder moment, lower maximum rate of rise of resultant elbow moment, and higher maximum rate of rise of resultant wrist moment. Subjects with a lower rate of rise of resultant shoulder moment on the leading (left) side were more likely to score a "yes" on item 12.

3.6. Multiple Regression Model for Part 1 Score. Lower average resultant trailing (right) elbow moment, lower maximal rate of rise of resultant leading (left) elbow moment, and higher maximal leading (left) shoulder extension moment were associated with proper completion of a greater number of transfer skills overall (higher TAI part 1 score) (Table 6).

4. Discussion

This is the first study to examine the association between proper and improper transfer skills and the resulting forces and moments imparted on the upper limb joints during the transfer process. Specific transfer skills, identified using the TAI, were found to be associated with kinetic variables related to injury risks on the upper extremities [18, 21, 24, 25, 37–39]. Our study sample included a diverse sample of wheelchair users who had a wide range of transfer skills (e.g., part 1 summary scores that ranged from 3.08 to 10.00). Despite differences across studies in measurement techniques and subject characteristics, our kinetic variables were in line with those values reported for level transfers in other studies. For example, the studies from Gagnon and Desroches et al. measured upper limb joint forces and moments during transfers among individuals with SCI and indicated that maximum wrist resultant moment ranged from 0.14 Nm/Kg to 0.48 Nm/Kg and shoulder posterior force on both sides were 2.64 N/kg and 3.14 N/kg, respectively [17, 40].

From the regression model results (Tables 5 and 6), it appears that transfer skills identified by the TAI are closely associated with the magnitude and timing of joint moments. During transfers, the wheelchair user's trunk and his/her arms can be thought as a tripod [47] which forms

Table 5: Logistic regression model results for each TAI item. Odds ratio (Exp(B)) is shown for the predictors that significantly contributed to predicting the TAI item scores. The Negelkerke R^2 value for each model is reported.

Item	Variables	B	χ^2	Sig.	Exp(B)	Model results
Item 1: the subject's wheelchair is within 3 inches of the object to which he is transferring on to.	Leading (left) shoulder AveRF	−2.45	3.55	.06		$\chi^2(2, N = 23) = 8.72$, $P = .01$, $R^2 = .42$
	Leading (left) shoulder MaxRMRate	1.32	3.39	.07		
Item 2: the angle between the subject's wheelchair and the surface to which he is transferring to is approximately 20–45 degrees.	Leading (left) shoulder MaxIRM*	−16.53	4.29	.04	.00	$\chi^2(1, N = 23) = 9.09$, $P < .01$, $R^2 = .46$
Item 6: the subject places his feet in a stable position (on the floor if possible) before the transfer.	Leading (left) shoulder MaxRMRate	−1.34	3.67	.06		$\chi^2(1, N = 23) = 7.86$, $P < .01$, $R^2 = .42$
	Trailing (right) shoulder AveRM*	−5.73	4.19	.04	.00	$\chi^2(1, N = 23) = 6.76$, $P < .01$, $R^2 = .37$
Item 7: the subject scoots to the front edge of the wheelchair seat before he transfers (i.e., moves his buttocks to the front 2/3rds of the seat).	Leading (left) shoulder MaxExtenM	3.91	3.54	.06		$\chi^2(1, N = 23) = 4.70$, $P = .03$, $R^2 = .27$
	Trailing (right) elbow AveRM	−13.34	2.96	.09		$\chi^2(2, N = 23) = 14.78$, $P < .01$, $R^2 = .69$
	Trailing (right) elbow MaxRMRate	−3.70	2.82	.09		
Item 9: a handgrip is utilized correctly by the leading arm (when the handgrip is in the individual's base of support).	Leading (left) shoulder MaxRMRate	−1.39	1.62	.20		$\chi^2(4, N = 23) = 18.29$, $P < .01$, $R^2 = .74$
	Leading (left) shoulder MaxIRM	22.10	1.85	.17		
	Leading (left) elbow MaxRMRate	−4.74	2.21	.14		
	Leading (left) wrist MaxRMRate	7.51	2.83	.09		
	Trailing (right) shoulder AveRM	−9.91	1.43	.23		$\chi^2(2, N = 23) = 19.92$, $P < .01$, $R^2 = .79$
	Trailing (right) elbow MaxRMRate*	−10.38	4.07	.04	.00	
Item 12: Head-hip relationship is used.	Leading (left) shoulder MaxRMRate	−.81	3.82	.05		$\chi^2(1, N = 23) = 5.13$, $P = .02$, $R^2 = .27$

Note: * the predictor significantly contributed to the regression model. B: unstandardized regression coefficients; Sig.: significance; Exp(B): odds ratio; Ave: average; Max: maximum; RMRate: rate of rise of resultant moment; RF: resultant force; IRM: internal rotation moment; RM: resultant moment; ExtenM: extension moment.

Table 6: Multiple linear regression analysis summary for predicting part 1 score.

Variable	B	SEB	β	sr^2	Sig.	Regression model
Trailing (right) elbow AveRM*	−5.86	2.02	−.53	.29	<.01	$F(1,21) = 8.40$, $P < .01$, $R^2 = .29$
Leading (left) shoulder MaxExtenM	1.94	.85	.35	.12	.03	$F(2,20) = 12.54$, $P < .01$, $R^2 = .56$
Leading (left) elbow MaxRMRate*	−1.13	.30	−.57	.31	<.01	

Note: *the predictor significantly contributed to the regression model. B: unstandardized regression coefficients; SEB: standard error of the unstandardized regression coefficients; β: standardized regression coefficients; sr^2: squared semipartial correlations; Sig.: significance; Ave: average; Max: maximum; RM: resultant moment; ExtenM: extension moment; RMRate: rate of rise of resultant moment.

a closed kinetic chain [48]. The skills used in transfers (e.g., positioning of the wheelchair, using correct handgrips, etc.) cause alterations in the moment arms or the distances separating the hands and trunk center of mass and changes in upper limb joint angles [49] that act along with the external forces to produce the resulting moments. Certain transfer skills helped to reduce the moments imparted on both upper limbs, while other skills had the effects of increasing the magnitudes or rates of loading on the leading (left) arm. Proper completion of the skills related to the trailing (right) arm (part 1 summary score and Items 6, 7 and 9) had the effect of lowering the trailing (right) shoulder and/or elbow peak resultant moment or rate of resultant moment loading. This is significant considering that the trailing arm tends to support a higher percentage of the body weight during sitting-pivot transfers [50, 51].

The six transfer skills as measured by the TAI were modeled because at least 20% of our subject sample scored a "no" for incorrect performance of a particular skill. Four of the six applicable TAI items (transfer skills) dealt with the setup of the wheelchair and body prior to making the transfer. Positioning the wheelchair within 3 inches of the target surface, as measured by item 1, was associated with a reduction in the average resultant shoulder force ($B = -2.45$, $P = .06$) and an increased rate of rise of shoulder resultant moment ($B = 1.32$, $P = .07$) (Table 5) on the leading (left) side. The increase in rate of rise may be associated with a shorter time needed to make the transfer when the body is in a position that is closer to the target surface. A proper angle (20 to 45 degrees) between the wheelchair and transfer surface (item 2) was associated with lower peak internal rotation shoulder moment on the leading (left) side ($B = -16.53$, $P = .04$) (Table 5). Angling the wheelchair next to the target as opposed to parallel parking provides a space that can be used to pivot the trunk and lower body over to the target surface. Angling the wheelchair also allows for the user to clear the rear wheel more easily. The pivoting actions of the trunk and lower body and clearing the pathway to the target surface may have helped to reduce the rotational demands on the leading shoulder.

Proper positioning of the feet (Item 6) can provide wheelchair users with greater dynamic postural control during transfers [18]. About 30% of the body weight during sitting pivot transfers is supported by the feet and legs [51]. Subjects who scored well on this item had lower resultant moments on the trailing (right) shoulder ($B = -5.73$, $P = .04$) and less maximum rate of rise of resultant moment at the leading (left) shoulder ($B = -1.34$, $P = .06$) (Table 5). "Scooting forward" to the front edge of the wheelchair seat before transfers (Item 7) was associated with less trailing (right) elbow moment and its rate of rise ($B = -13.34$ and -3.70, $P = .09$ and $.09$) (Table 5). Scooting forward brings wheelchair users and their trailing hand positions closer to the target surface which would decrease the lever arm that the applied force is acting through. Our regression model however also indicated that this skill increases leading (left) shoulder extension moment ($B = 3.91$, $P = .06$) (Table 5). The increasing shoulder extension moment may have resulted from a shift in loading from the trailing arm to the leading arm. As mentioned, the trailing arm bears more force in a transfer. Getting closer to the surface allows for placing both hands closer to the trunk center of mass which helps to balance the loading more equally across both arms [52]. For persons who position themselves correctly this will mean seeing less loading on the trailing arm and possibly more loading on the leading arm. In any case higher shoulder extension moment has been shown to increase the risk of pathology, such as ligament edema [24]. Close positioning and appropriate angling wheelchair and foot placement may help to mediate the increased shoulder moments experienced on the leading side.

Item 9 evaluates whether wheelchair users use a correct handgrip on the leading arm within their base of support when performing transfers. Clinical practice guidelines encourage wheelchair users to use handgrips instead of flat hands or fists when performing transfers [11]. Using flat hands during transfers will cause extreme wrist extension which is one factor identified in the etiology of carpal tunnel syndrome, while a closed-finger fist will result in excessive pressure on the metacarpal joints [11, 53]. The use of handgrips can prevent extreme wrist angles, provide more stability, and help apply forces during transfers [11]. During transfers, the handgrip choices are limited by the type of transfer surface and the handgrip option available. For the bench transfer evaluated in this study, subjects could either drape their leading fingers over the edge of the bench with the palm resting on the surface, place a flat palm or fist anywhere on top of the bench, or use the adjacent grab bar. If they used a flat palm, used a closed-finger fist, and/or placed their leading hand outside of where the clinicians felt would be their base of support, the subjects were scored a "no" on this item. Our results from the regression models showed that using a correct leading handgrip (item 9) can lower shoulder resultant moment ($B = -9.91$, $P = .23$) and rate of rise of

elbow moment ($B = -10.38$, $P = .04$) on the trailing (right) side and lower the rate of rise of the shoulder and elbow resultant moments on the leading (left) side ($B = -1.39$ and -4.74, $P = .20$ and .14) (Table 5). Because this item combines multiple aspects of handgrips it is difficult to know exactly which attribute (e.g., type of finger grip or hand placement within the base of support) is more responsible for the kinetic outcomes. The rate of rise of the wrist resultant moment increased with better handgrip ($B = 7.51$, $P = .09$) which may be associated with the types of handgrips used by the subjects which were not explicitly documented in this study. Future research should be done to investigate the impact that different types of handgrips used in transfers have on the upper limb joint forces and moments.

Wheelchair users who use the head-hips technique appropriately (Item 12) experienced lesser rate of rise of moment on the leading (left) shoulder ($B = -.81$, $P = .05$) (Table 5). This technique has been associated with an increase in trunk forward flexion and a shift of the trunk center of mass forward and downward to create a moment which can facilitate lifting the buttocks during the transfer [54]. As with setting up the wheelchair angle appropriately, the increased trunk pivot motions may have helped to reduce the rate of rise of resultant shoulder moment.

Wheelchair users with proper overall transfer skills (higher part 1 summary scores) were more likely to experience lower moments on the trailing (right) elbow ($B = -5.86$, $P < .01$) and lower rate of rise of resultant moment on the leading (left) elbow ($B = -1.13$, $P < .01$) but increased extension shoulder moment on the leading (left) side ($B = 1.94$, $P = .03$) (Table 6). Shoulder and elbow movements are related to each other in a close chain activity [48]. As observed with the individual TAI items using good skills can shift loading off of one joint onto another or from one arm to the other. Offloading the elbows and loading the shoulders more may make for a more efficient transfer particularly for individuals who lack elbow extension function. Although triceps muscle function can make a transfer easier (assist with lifting the buttocks off the surface) it is not a primary mover in transfers. The primary movers for transfers are the actions of the pectoralis major muscles, serratus anterior, and latissimus dorsi muscle groups which are all attached to shoulder [17, 55]. The increasing extension shoulder moment may have resulted from the recruitment of the large primary movers, such as the latissimus dorsi and pectoralis major muscles [19, 56] which can drive the movement and shift the body weight during transfers [57].

As noted in our regression models (Tables 5 and 6), some transfer skills as measured by the TAI increase magnitudes and rates of rise of moments. By properly using different transfer skills in tandem, the risks associated with secondary injuries may be minimized. For example, wheelchair users should angle their wheelchairs appropriately relative to the target surface (20–45 degrees) to reduce the large internal rotation shoulder moments on the leading side which can occur when using a proper leading handgrip. Using the head-hip technique (item 12 skill), can reduce the increasing rate of rise of leading shoulder moments which was also associated with close wheelchair positioning. Wheelchair users may need to combine skills to reduce biomechanical loading on the upper extremities, to a greater effect than when utilizing only one or the other. For example, wheelchair users should combine close wheelchair positioning with the scooting forward in their wheelchair to reduce the extension moment on leading shoulder. Taking into consideration the kinetic effects of all transfer skills studied may help to relieve negative effects on the upper extremities during transfers.

5. Study Limitations

The small sample size may have negatively affected the power of the statistical analyses and response rate for some of the TAI items. For example not all of the items could be modeled because subjects were either too proficient on the item or the item did not apply to their transfer. This study only analyzed transfers from a wheelchair to a level-height bench located on the subjects' left side and required them to use the wheelchair side grab bar for positioning of the trailing hand (Figure 1). Subjects were given time to acclimate to the setup prior to testing. Furthermore, a prior study found no differences in muscular demand based on which side (dominant or nondominant) led the transfer or preferred direction of transfer [56]. Wheelchair users have to learn to be flexible with adapting to different setups when they transfer in public places where places to position their hands or the area to position their wheelchairs is limited. Future studies should consider the effects of skills on kinematic variables. Furthermore, the biomechanical effects of transfer training based on TAI principles should be investigated.

6. Conclusions

The study shows that the transfer skills that can be measured with the TAI are closely associated with the magnitude and timing of joint moments. Certain transfer skills helped to reduce the moments imparted on both upper limbs, while other skills had the effects of increasing the magnitudes or rates loading on the leading limb. Different skills have different kinetic effects on the upper extremities. Taking into consideration the kinetic effects from all the transfer skills studied may help to reach better load-relieving effects on the upper extremities during transfers. The study provides insight into the impact that a specific skill can have on upper limb loading patterns. As such the TAI may be useful for measuring the effects of a training intervention on reducing upper limb joint loading.

Disclosure

The authors certify that no party having a direct interest in the results of the research supporting this paper has or will confer a benefit on them or on any organization with which they are associated and, if applicable, they certify that all financial and material support for this research (e.g., VA) and work are clearly identified in the title page of the paper.

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Conflict of Interests

Chung-Ying Tsai, Nathan S. Hogaboom, Michael L. Boninger, and Alicia Koontz declare that they have no proprietary, financial, professional, or other personal interests of any nature or kind in any product, service, and/or company that could be construed as influencing the position presented in, or the review of, this paper.

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Research Article

Effects of Repetitive Shoulder Activity on the Subacromial Space in Manual Wheelchair Users

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This study investigated (1) the effect of repetitive weight-relief raises (WR) and shoulder external rotation (ER) on the acromiohumeral distance (AHD) among manual wheelchair users (MWUs) and (2) the relationship between shoulder pain, subject characteristics, and AHD changes. Twenty-three MWUs underwent ultrasound imaging of the nondominant shoulder in an unloaded baseline position and while holding a WR position before and after the WR/ER tasks. Paired t-tests and Spearman correlational analysis were used to assess differences in the AHD before and after each task and the relationships between pain, subject characteristics, and the AHD measures. A significant reduction in the subacromial space ($P < 0.01$) occurred when subjects performed a WR position compared to baseline. Individuals with increased years of disability had greater AHD percentage narrowing after WR ($P = 0.008$). Increased shoulder pain was associated with AHD percentage narrowing after ER ($P \leq 0.007$). The results support clinical practice guidelines that recommend MWUs limit WR to preserve shoulder function. The isolated repetitive shoulder activity did not contribute to the changes of subacromial space in MWUs. The ultrasonographic measurement of the AHD may be a target for identifying future interventions that prevent pain.

1. Introduction

Subacromial impingement syndrome (SIS) is a common shoulder dysfunction in manual wheelchair users (MWUs). The mechanisms of SIS can be divided into intrinsic and extrinsic factors. [1–3] The intrinsic factors include rotator cuff degeneration, aging, arthritis, acromial shape, and abnormalities including subacromial and acromioclavicular joint spurs. Extrinsic factors include misalignment of the shoulder joint caused by muscle weakness or improper trunk postures, altered scapular kinematics, and mechanical compression from forces that drive the humeral head further into the glenohumeral joint, causing impingement of the supraspinatus tendon under the acromioclavicular arch and inflammation. Intrinsic and extrinsic factors may not be mutually exclusive and are exacerbated by overuse syndromes [2].

MWUs commonly experience overuse because their upper extremities are used extensively for mobility and activities of daily living (ADL). The weight-relief raise (WR) is an ADL that requires heavy and frequent shoulder loading. During a WR, MWUs need to lift and support the weight of the body to reduce pressure on the buttocks. This activity results in excessive shoulder joint loading and requires rotator cuff muscles to maintain glenohumeral joint stability [4–6]. van Drongelen et al. simulated shoulder joint reaction forces during the WR using musculoskeletal modeling techniques. They found that large weight-bearing forces (1288 N) acted to drive the humerus into the glenohumeral joint during the WR [6]. Gagnon et al. compared shoulder mechanical

loads during WR and sitting pivot transfers among 13 MWUs with spinal cord injury (SCI). They reported that the bodyweight-normalized superior shoulder joint force during WR (2.91 N/kg) largely exceeded the amplitudes found during sitting pivot transfers in the leading arm (1.63 N/kg) and trailing arm (1.47 N/kg). Due to the limited size of the subacromial space, WR positioning is most likely to impinge the subacromial structures [7]. There is limited information on the impact of holding the WR position and isolated repetitive WR maneuvers on the subacromial space.

Shoulder external rotation (ER) is a commonly prescribed training among MWUs to strengthen the shoulder external rotators to act against potentially injurious forces during wheelchair activities [8]. Shoulder external rotators, including infraspinatus, supraspinatus, posterior deltoid, and teres minor, are important for maintaining glenohumeral joint positioning [9]. Previous studies have found MWUs with paraplegia have comparative weakness of shoulder external rotators compared to shoulder internal rotators, resulting in shoulder muscle imbalances [10]. Shoulder muscle imbalances can lead to functional instability of the glenohumeral joint, resulting in the subacromial space narrowing and placing the individual at a higher risk of developing SIS [11]. Previous studies have implied the narrowing of the subacromial space after isolated repetitive ER in subjects with SIS or rotator cuff tear. However, there is a knowledge gap regarding how the isolated repetitive ER contributes to subacromial space narrowing in the MWU population.

We recently described a reliable method to quantify the subacromial space by using ultrasound while holding a WR position [12]. Ultrasound has the advantage of enabling the shoulder to be scanned in a functional posture. The primary purpose of this study was to investigate the subacromial space with the shoulder in an unloaded neutral position (e.g., baseline) and in a WR position both before and within one minute after isolated repetitive WR and ER tasks. We hypothesized that the acromiohumeral distance (AHD), linear measurement of the subacromial space, in the WR position, would be narrower than the baseline AHD. We also hypothesized that the AHD would be narrower after subjects completed each protocol compared to before the protocol. A secondary goal of this study was to examine the relationship between shoulder pain, subject characteristics, and AHD.

2. Methods

2.1. Subjects. Study participants were a convenience sample recruited during the 2011 National Veterans Wheelchair Games (NVWG). For a power of 0.95, an α level of 0.05, and an effect size of 1.09 using a paired t-test, a sample size of 23 was required for this study. The effect size was calculated based on ultrasound data collected from wheelchair users in previous reliability study [12]. Inclusion criteria included using a manual wheelchair as primary means of mobility, able to perform at least 10 WR in a row without assistance, and between 18 and 65 years of age. The exclusion criteria included history of fractures or dislocations in the shoulder from which the subject had not fully recovered, upper limb dysesthetic pain as a result of a syrinx or complex regional pain syndrome, and history of cardiovascular or cardiopulmonary disease. Informed consent was obtained from all the subjects before participation in this study. The research protocol was approved by the Department of Veteran Affairs Institutional Review Board.

2.2. Questionnaires. Basic demographic information including age, height, weight, and date of injury/diagnosis was collected using self-report. All subjects completed the Wheelchair Users Shoulder Pain Index (WUSPI) [13]. The WUSPI is a reliable and validated 15-item self-report instrument that measures shoulder pain intensity in wheelchair users in the last seven days during various functional activities of daily living including transfers, wheelchair mobility, dressing, overhead lifting, and sleeping [14]. Each item is scored using a 10 cm visual analog scale anchored at the ends with the descriptors of "no pain" and "worst pain ever experienced." Total score was calculated by summing the individual scores divided by the number of performed activities and then multiplying by 15 [15].

The OMNI pain scale is a numerical rating scale ranging from 0 to 10 [16–18]. The OMNI pain scale has been previously validated for walking, running, and cycle ergometer exercise and for use by male and female adults during upper and lower body resistance exercise. The OMNI scale was administered prior to the beginning of testing, to establish a baseline measure of pain, and after each activity, to determine the intensity of activity-induced pain experienced during the testing.

2.3. Procedure. Shoulder circumference and upper arm length were obtained from all subjects at the beginning of testing. The shoulder circumference and upper arm length were measured while the subjects were in the seated anatomical position. The shoulder circumference was measured from the superior portion of the acromion to the axilla. The upper arm length was measured from the most lateral and superior portion of the acromion to the tip of the olecranon. A single investigator conducted all of the measurements using a standard tape measure. Using this method to record similar anthropometrical measures has been found to be reliable [19].

Subjects transferred to a Biodex System 3 dynamometer (Biodex Medical System, Inc., Shirley, NY) with custom-made adjustable height armrests. Armrests were fitted to each subject to allow pushing straight up with full elbow extension to off load the buttock tissue. The seat height was fixed during the entire testing. The WR entailed lifting and holding the buttocks off the seat with an elbow locked position [4, 5]. The WR task was repeated at a rate of 30 repetitions per minute to the auditory cue of a metronome. Subjects were instructed to stop when they were no longer able to continue or until they completed two minutes of activity. The total number of WR raises (60) is similar to the number that would be performed each day in case of following the recommended frequency of pressure relief (one time every 15 to 30 minutes) [20].

The ER task followed a similar protocol to a previous study involving neurologically intact individuals without

shoulder disorders and was designed to overuse the shoulder external rotators [21]. The nondominant side upper arm was adducted at the side with the elbow bent 90°. The subject was instructed to externally rotate the forearm from a shoulder neutral position to 45° or the maximum range of ER that they could reach comfortably [22]. The trunk was secured to minimize compensatory movements using straps from the Biodex that crossed the chest and lap. The strap has been used in previous studies to support targeted joint movements among spinal cord injured and able-bodied subjects [23, 24]. The dynamometer was adjusted to match the level of their tested and nontested shoulders before the ER activity. Resistance for the ER task was set for 5% of self-reported body weight [25]. To minimize the involvement of the shoulder internal rotators, the minimum resistance setting of 0.5 kg was used for the internal rotation direction from an externally rotated position back to neutral [26]. ER protocols were administered at the same pace and ended in the same manner as the WR task. The subjects rested in between the two protocols for a period of approximately 15 minutes.

2.4. AHD Ultrasound Examination. The subacromial space was quantified by measuring the AHD using ultrasound techniques as described in a previous reliability study [12]. The intrarater reliability of the AHD measurement with the shoulder in a neutral and WR position resulted in a standard error of measurement of 0.21 and 0.52 mm and intraclass correlation coefficient of 0.93 and 0.98, respectively [12]. A single examiner conducted all scans for each subject using a Philips HD11 1.0.6 ultrasound machine with a 5–12 MHz linear transducer. A water-based gel was applied on the skin to enhance conduction between the ultrasound probe and skin surface.

The nondominant side was chosen for all the AHD measures in order to minimize the effects caused by performing other types of activities of daily living on the dominant shoulder. The muscular demand of the nondominant shoulders among manual wheelchair users was also examined in previous studies [27]. The nondominant shoulder was scanned from the anterior aspect of glenoid to the flat segment of posterior acromion to capture the bright reflection of the bony contour of the acromion and humeral head (Figure 1). Ultrasound video was recorded at 60 Hz and scanning took approximately 10 seconds. A baseline US video was recorded with the shoulder in a neutral and resting position. Before and within one minute after each protocol, imaging was completed while the subject isometrically held the WR position [12]. We chose to examine AHD while subjects held the WR position because it provides a measure of what the AHD looks like under realistic, functional loading conditions.

2.5. Data Analysis. An investigator who was blinded to the timing of the video (e.g., pre or post) used a custom developed Matlab program to manually review each frame of the video and mark the inferior edge of acromion and superior margin of the humeral head. (Figure 1) The distance between the bony landmarks was calculated for each frame of the video

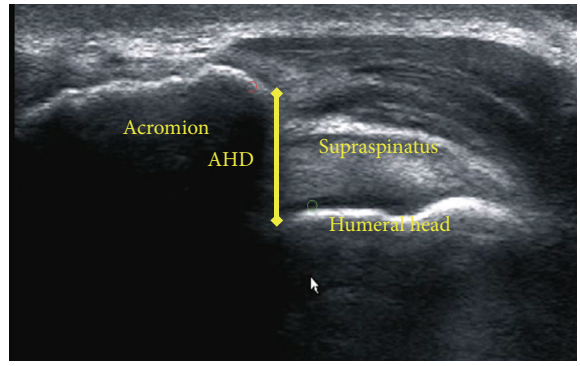

Figure 1: Ultrasonographic image of the acromiohumeral distance (AHD).

and the narrowest distance across all frames was used for statistical analyses. A Shapiro-Wilk test indicated that the data followed a normal distribution. Therefore two-tailed paired t-tests were used to assess the difference in AHD between the baseline (unloaded) and WR shoulder positions as well as before and after performing the WR and ER tasks for all subjects (neutral versus pre-WR, pre-WR versus post-WR, pre-ER versus post-ER, and pre-WR versus pre-ER). A Bonferroni correction for multiple comparisons was applied, with a resultant level of significance of $P < 0.013$. Pearson's or Spearman's correlation was used to investigate relationships between the continuous measures, including AHD, AHD percentage changes (1), WUSPI score, OMNI scale score, and demographic data (e.g., height, weight, shoulder circumference, arm length, age, and years since acquiring the disability or injury). The strength of correlation was defined as a good to excellent relationship (r is above 0.75), moderate to good relationship ($r = 0.50$ to 0.75), fair relationship ($r = 0.25$ to 0.50), and little or no relationship ($r = 0.00$ to 0.25) [28]. Subject demographic variables statistically associated with the AHD measures were controlled for when testing the relationships between the AHD and pain measures. An alpha level less than 0.05 was established for significant changes:

$$\text{AHD percentage change (\%)} = \frac{\text{post-AHD measure} - \text{pre-AHD measure}}{\text{pre-AHD measure}} \times 100\%. \quad (1)$$

3. Results

3.1. Subjects. Twenty-three MWUs (twenty-two men and one woman) participated in this study. Sixteen MWUs had a spinal cord injury (five cervical and eleven thoracic), one had a unilateral transfemoral amputation, three had bilateral transtibial amputations, and three had multiple sclerosis. Twenty-two participants were right hand dominant. Descriptive data are provided in Table 1.

There were no significant differences in the AHD before and after performing WR ($P = 0.89$) and ER ($P = 0.81$) (Table 2). The AHD in the pre-WR and pre-ER positions

Table 1: Subject demographics ($n = 23$).

Demographic	Mean ± standard deviation	Range
Age	46 ± 12	26–64
Height (m)	1.78 ± 0.08	1.65–1.93
Weight (kg)	81 ± 18	55–130
Time since injury (year)	15 ± 10	1.5–33.5
Number of WR	34 ± 16	10–61
Number of ER	39 ± 18	6–60
WUSPI	14.08 ± 18.07	0–60, median 12.6
OMNI pain scale baseline	1.04 ± 1.58	0–5
OMNI pain scale after WR	2.09 ± 2.56	0–8
OMNI pain scale after ER	2.30 ± 2.42	0–7

were not different ($P = 0.38$) but both were significantly smaller than the AHD in the baseline shoulder neutral position ($P < 0.001$). No relationship between baseline AHD and age, height, weight, or arm length was found. Individuals with narrower AHD at baseline had smaller shoulder circumferences ($r = 0.42$, $P = 0.044$, Figure 2(a)). Individuals with increased years of disability had greater AHD percentage narrowing after the WR task ($r = -0.54$, $P = 0.008$, Figure 2(b)). More shoulder pain on WUSPI was associated with greater percentage narrowing of the AHD after the ER task ($r = -0.41$, $P = 0.007$, Figure 2(c)). The OMNI pain scale results measured at baseline, after WR, and after ER were 1.04 ± 1.58, 2.09 ± 2.56, and 2.30 ± 2.42, respectively. Individuals with higher scores on the OMNI pain scale after ER had greater percentage narrowing of the AHD after ER ($r = -0.59$, $P = 0.003$) (Figure 2(d)).

4. Discussion

The results of this study indicate that AHD narrowing occurs when MWUs assume a weight-bearing position with their arms. When our subjects assumed the WR position, a statistically significant reduction in space occurred. In this position, the elbows are in full extension allowing the humeral head to be oriented more directly upward and into the joint while the scapula is anteriorly tilted and internally rotated [29]. The humeral head migration, scapular anterior tilting, and internal rotation as well as the large superior or posterior weight-bearing forces likely contribute to a reduction in subacromial space [30]. This narrowing of the AHD can lead to commonly experienced pathologies among MWUs, including enlarged bursa, tendon inflammation, or irregularities of the gliding surface [31]. Weight-bearing positions are difficult to avoid and occur daily at high frequency during wheelchair transfers and pressure relief, causing wheelchair users to be at risk for developing SIS. These findings support clinical practice guidelines that recommend MWUs limit the WR technique for pressure relief [8].

In contrast to our hypothesis, this study did not find differences in the AHD measures before and after isolated repetitive WR. One possible explanation is that not all subjects may have experienced overuse in the targeted rotator cuff muscles. Many individuals in our study were not able to complete two minutes of activity which could have been a result of fatigue or other reasons (e.g., pain onset, discomfort, boredom, etc.). Also, the triceps are the most active muscle group during a WR task and overuse would most likely occur in this muscle first [4]. In addition, a fine-wire electromyography (EMG) study showed that the rotator cuff and depressor muscles (e.g., supraspinatus, infraspinatus, subscapularis, and serratus anterior) were minimally active during the WR task (less than 25% of maximum voluntary contraction) whereas moderate to high activity was found for the sternal pectoralis major and latissimus dorsi muscle groups [4]. Activation of these muscles is believed to help transfer humeral loads onto the trunk, functionally circumventing the glenohumeral joint, and reducing the potential for impingement. Thus, it may be more difficult to detect changes in the AHD when the shoulder is in the WR position.

The ER activity is an overuse protocol targeting the external rotators and minimizing involvement of other shoulder muscles. Previous studies have shown superior migration of the humeral head occurs after overusing the shoulder external rotators [32]. However, we found no difference in the AHD after the task among MWUs except in subjects who had greater levels of shoulder pain. This difference could be because prior studies on neurologically intact individuals measured subjects with their arms in a nonweight-bearing position during scapular plane abduction. With arm elevation, the deltoid muscle enhances the upward pull of the humerus. Increased humeral head migration has been reported when the inefficient force of the fatigued rotator cuff muscles could not counteract the superior force from the deltoid muscle [33]. Active shoulder elevation would likely magnify the upward shift of the humeral head after the isolated repetitive ER in contrast to the WR position. Further studies are needed to understand if there are other arm positions that would be more sensitive to detecting changes with the overuse of shoulder external rotators.

Not finding differences in AHD after the isolated repetitive tasks may also result from compensatory scapular motions [34] and motor strategies or muscle firing patterns [35] used to preserve the subacromial space [36]. A recent meta-analysis reported that subjects with SIS often demonstrate altered three-dimensional scapular behavior [37], including decreased scapular posterior tilt, upward rotation, and external rotations, which may negatively impact the subacromial space [38–40]. However, other investigations have reported different behavior, such as increased upward rotation scapular rotation, suggesting these motions are compensatory mechanisms to avoid further shoulder pain during activity [34]. Increased upward rotation and external rotation of the scapula, which would have a positive effect on the AHD, have also been linked to inadequate rotator cuff function [41]. It is possible that in response to the overuse our subjects were demonstrating compensatory scapular kinematic patterns in favor of protecting the subacromial space [4]. Future studies are warranted to understand the relationship between scapular motions during isolated repetitive shoulder tasks and changes in the subacromial space.

Table 2: AHD for each subject.

Disability	Rest	Multiple weight-relief raises (mm)		Shoulder external rotation activity (mm)	
		Pre	Post	Pre	Post
C3 spinal stenosis	14.07	10.85	10.85	10.67	9.31
T4 com. SCI	11.53	9.32	11.27	9.18	9.04
C6 inc. SCI	10.96	10.14	9.03	8.77	8.90
MS	12.64	8.08	9.45	11.25	10.82
Amp (LAK)	12.88	12.76	10.17	10.17	10.34
T4 com. SCI	11.51	10.00	9.31	9.72	10.82
Amp (RBK, LAK)	12.37	9.83	11.93	9.32	10.41
T7 inc. SCI	11.64	10.28	11.37	10.70	10.69
T9 inc. SCI	12.50	11.64	11.51	11.67	11.10
MS	9.32	9.04	11.53	10.27	8.92
C3 inc. SCI	10.96	10.55	10.83	10.00	8.45
MS	12.36	9.03	7.16	8.47	8.36
T12 com. SCI	9.44	10.00	9.31	10.00	9.73
T12 inc. SCI	11.81	10.00	10.27	10.27	10.69
Amp (RAK, LAK)	10.00	8.38	9.31	7.95	8.08
T12 com. SCI	10.14	9.31	8.77	8.75	8.49
Amp (RAK, LBK)	10.14	8.49	7.36	7.50	9.04
C5 inc. SCI	13.06	11.39	10.82	11.39	10.95
C7 inc. SCI	16.32	10.86	9.66	10.52	10.17
T10 inc. SCI	13.83	12.71	12.28	12.41	13.10
T11 inc. SCI	11.22	10.27	9.45	10.14	10.96
T12 com. SCI	13.84	11.10	11.51	9.73	10.41
T9 inc. SCI	8.22	6.03	6.08	6.71	5.83
Group mean	$11.78 \pm 1.83^{\dagger,\ddagger}$	$10.00 \pm 1.51^{\dagger}$	9.97 ± 1.60	$9.81 \pm 1.36^{\ddagger}$	9.77 ± 1.47

SCI, spinal cord injury (com., complete; inc. incomplete); Amp, amputee; RAK, right leg above knee; RBK, right leg below knee; LAK, left leg above knee; LBK, left leg below knee; MS, multiple sclerosis. $^{\dagger,\ddagger}P < 0.05$.

Our study found that more shoulder pain was related to greater AHD percentage narrowing after ER. Not finding the same association after the WR task may point to the effectiveness of the ER task in targeting overuse of the infraspinatus muscle [42]. As mentioned, SIS has been associated with scapular kinematics, which reduce the subacromial space. However how pain influences scapular kinematics is not well understood. A recent study anesthetized painful impinged shoulders and found greater anterior tipping of the scapula during greater humeral elevation angles, which can further reduce the subacromial space [43]. Thus pain may help protect the space from greater narrowing to some degree. However our study found that greater amounts of pain did not hinder narrowing of the space after repetitive ER activity in our sample of wheelchair users. The AHD measure may be useful in the future for evaluating the effectiveness of interventions that are targeted at reducing shoulder pain among wheelchair users.

Our results were consistent with other studies that found that AHD measures were not significantly correlated with the characteristics commonly linked to SIS such as age and weight [44, 45]. However a positive correlation was found between percentage narrowing of the AHD after WR with years of injury/diagnosis. Several studies have found that a longer duration of wheelchair use is associated with greater pain and shoulder pathology [8, 46, 47]. In addition to shoulder pathology, other problems commonly seen in long-term wheelchair users, including muscle strength imbalances around the shoulder, joint instability, altered scapular kinematics and abnormal glenohumeral motion, and subluxation, likely contributed to a greater reduction in the space following the WR task [48].

Our study had several limitations. Because our protocol was conducted at a national wheelchair sporting event, it was difficult to control for the amount of upper limb activity experienced before the testing. We conducted the informed consent process and questionnaires at the beginning of the study (e.g., a process that took 15–20 minutes) which helped to provide some washout period for the participants before starting the protocol. Also the within-subject design helped to control the effects that varying amounts of preactivity may have had on the primary pre-/post-AHD measures. In addition, wheelchair users who participate in sporting events may be considered more active than the general population. However, Tolerico et al. found that veterans who participate in the NVWG are not significantly different with respect

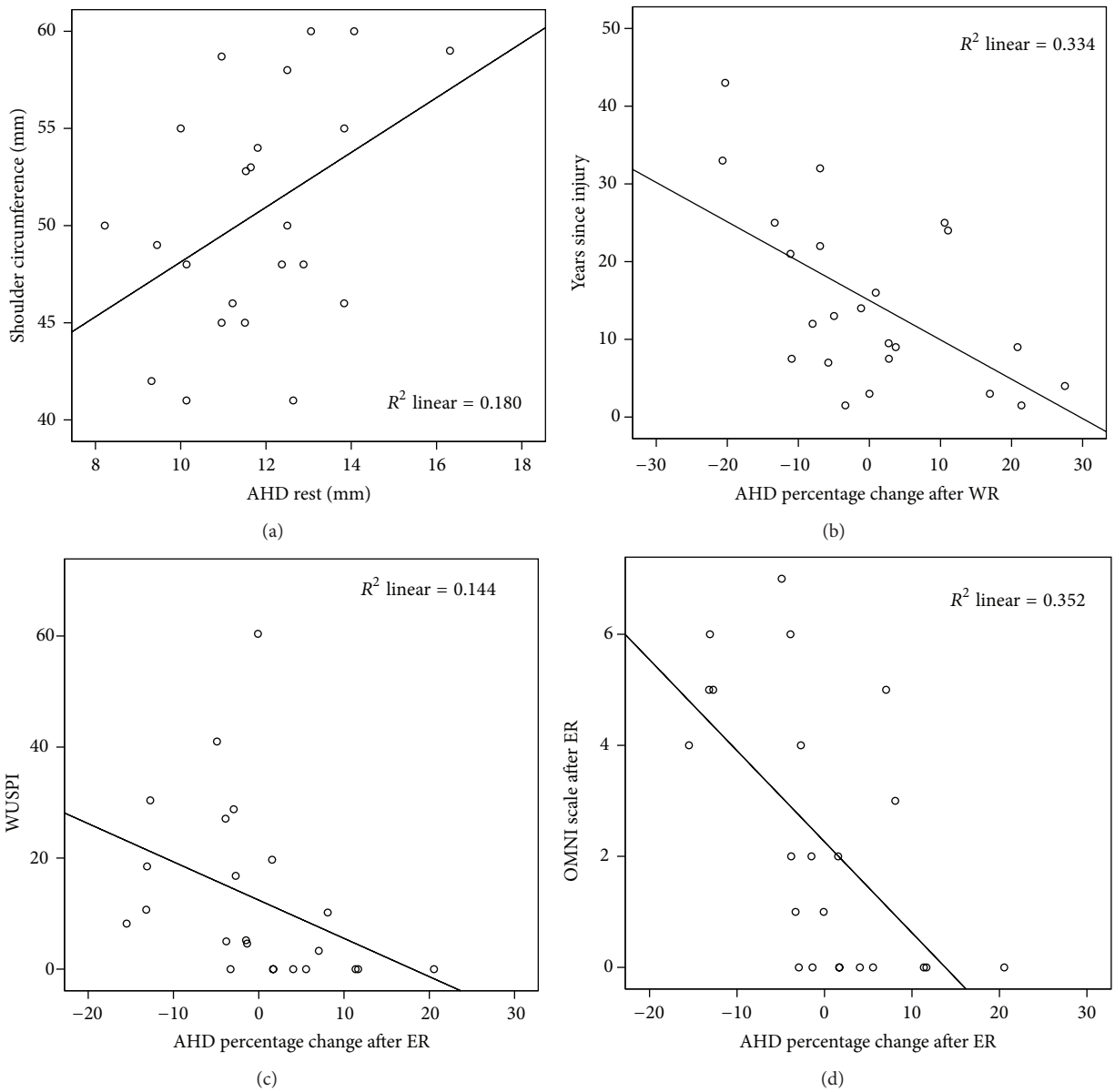

Figure 2: Correlation analysis for the AHD in neutral shoulder position with the shoulder circumference ($P = 0.044$, $n = 23$) (a), AHD percentage change after multiple weight-relief raises with years since injury ($P = 0.008$, $n = 23$) (b), AHD percentage change after shoulder external rotation activity with WUSPI ($P = 0.007$, $n = 23$) (c), and AHD percentage change with OMNI pain scale after ER ($P = 0.003$, $n = 23$) (d).

to mobility characteristics and activity levels from their community-dwelling wheelchair using counterparts [49]. Another limitation is that the two activities were performed in order (WR followed by ER) on the same day and it is possible that there was not enough recovery time to compensate for the overuse on the muscles. It is reassuring that the two AHD measures taken before each task were not significantly different.

Finding significant relationships between AHD changes and pain in our study implies that there is clinical relevancy with the AHD measure. However, because we did not specifically target symptomatic subjects for this study or study the effects of an intervention, more work is needed to define clinically meaningful changes in the AHD for manual wheelchair users. Limited research has been done so far in a wheelchair user population. A previous study on non-wheelchairs found that the ultrasonographic measurement of the AHD in affected shoulders among individuals with SIS (19.4 mm) was significantly narrower than the AHD in their nonaffected shoulder (22.2 mm, $P < 0.001$) [45]. Another study found statistical significance for mean differences in the AHD that ranged from 1.7 to 2.1 mm before and after a scapular assistant test designed to change scapular position and subacromial space among symptomatic subjects. Both

of these studies suggest that there may be clinical relevancy with ultrasound AHD changes on the order of 1.7 to 2.8 mm. This is within the realm of the statistically significant mean differences found in the AHD between the neutral and WR position. Other variables such as acromial shape, abnormal scapular kinematics, and impaired rotator cuff function were not investigated and could be additional sources to explain AHD narrowing. Acute changes were examined with the shoulders in a loaded position and differences may have been more apparent had the arm been scanned in an elevated, unloaded position. As scapular orientation has also been shown to affect AHD [50], future work should investigate scapular and humeral positioning to gain further insight into injury mechanisms.

5. Conclusions

The results of this study suggest that MWUs should limit WR for pressure relief, as placing the shoulder in a WR position led to a significant reduction in the subacromial space. The isolated repetitive shoulder activity did not contribute to the changes of subacromial space in MWUs. This study provides objective evidence that the AHD is associated with pain and long-term use of a wheelchair. Ultrasonographic measurement of the AHD may be useful for identifying interventions that prevent pain. A better understanding of the scapular and humeral kinematics may help to elucidate mechanisms leading to subacromial impingement in wheelchair users.

Conflict of Interests

The authors have no conflict of interests to disclose.

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Research Article

Sliding and Lower Limb Mechanics during Sit-Stand-Sit Transitions with a Standing Wheelchair

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Purpose. This study aimed to investigate the shear displacement between the body and backrest/seat, range of motion (ROM), and force acting on the lower limb joints during sit-stand-sit transitions by operating an electric-powered standing wheelchair. *Methods and Materials.* The amounts of sliding along the backrest and the seat plane, ROM of lower limb joints, and force acting on the knee/foot were measured in twenty-four people with paraplegia. *Results.* Without an antishear mechanism, the shear displacement was approximately 9 cm between the user's body and the backrest/seat surfaces. During standing up, the user's back slid down and the thigh was displaced rearward, but they moved in opposite directions when wheelchair sat back down. A minimum of 60 degrees of ROM at the hip and knee was needed during sit-stand-sit transitions. The maximal resultant forces acting on the knee restraints could reach 23.5% of body weight. *Conclusion.* Sliding between the body and backrest/seat occurred while transitioning from sitting to standing and vice versa. A certain amount of ROM at lower limb joints and force acting on the knee was necessitated during sit-stand-sit transitions. Careful consideration needs to be given to who the user of the electric powered standing wheelchair is.

1. Introduction

Standing is a routine activity of daily living (ADL) that allows individuals not only to talk and socialize with others, but also to perform many regular and important ADLs, such as cooking, grooming, or reaching objects. For people with paraplegia, standing with the help of a standing frame or a standing wheelchair helps to increase functional independence and improves an individual's physical and psychological well-being [1]. A standing wheelchair is an assistive device that enables people with paraplegia to achieve an upright posture and perform standing activities. A standing feature integrated into a wheelchair base allows the user to obtain a standing position without the need to transfer from the wheelchair. A mechanical or electromechanical system manipulated via the wheelchair's controller moves the seat surface from horizontal into a vertical or obliquely sloping position to anterior inferiorly position while maintaining vertical position of the legrest and backrest, thus extending the hip and knee joints. A full vertical standing position can be achieved directly from sitting, or through gradual angle changes from a supine position, or a combination of these positions [2]. People with paraplegia may use a standing wheelchair to stand up for a variety of health-related reasons, such as increasing their range of motion [3, 4], maintaining bone density [5, 6], improving circulation [7], decreasing spasticity [8, 9], preventing pressure sores [7, 9], and improving bowel and bladder function [4, 7, 9].

Moreover, rising from sitting to standing and its reverse are necessary and functionally important activities. While using a standing wheelchair, each sit-stand-sit movement cycle consisted of two phases: a rising phase during which the subject rose passively by a standing mechanism from the seated position to the standing position, and a descending phase during which the subject descended from standing

to the seated position. These two phases in the sit-stand-sit movement cycle are discrete. During the transition from sitting to standing, the seat and knee block of the wheelchair which supports the user are often rotated about the pivot points parallel to the user's hip and knee joint, causing the user to assume a standing position [10]. If the joints of the wheelchair do not follow the anatomical paths of the user, the sliding will occur. On the contrary, sitting down transition is performed with the assistance of gravity (downward movement). The forward trunk rotation into flexion required to sit down was operated by lowering the seat while keeping the backrest in a rearwardly inclined vertical position. As a result, large sliding displacement could occur and these shear forces will be developed because these forces between the user and the backrest/seat usually prevent the user from sliding back up to the original position. Large shear forces in the backrest and seat could lead to the development of pressure ulcer [11]. Study has shown that, without an antishear mechanism, up to 11 cm of displacement can take place between the person's back and the wheelchair's backrest while using powered reclining wheelchair [12]. But there were a few studies addressing these displacements with respect to the backrest and seat plane on the standing wheelchair. Therefore, it is worth investigating possible shear displacements with the use of a standing wheelchair.

In order to secure the standing mechanism, the larger force would be applied to the lower extremities, especially at knee joint, while using knee block placed just anteriorly to the knees. Excessive force exerted against the knee joint might be at high risk for a bone fracture [13]. So far, to the best of the author's knowledge, no studies investigated the magnitude of force over the knee block during transition of standing with a standing wheelchair. Furthermore, when individuals with paraplegia are unable to stand independently due to increased muscle tone, weakness, or imbalance, they are at risk of shortening (contracture) of the muscles which bend the hip (e.g., iliopsoas), those which straighten the hip and bend the knee (e.g., hamstrings), and the calf muscle which bends the knee and ankle. While being not a substitute for therapy, the user of standing wheelchair may benefit from partial weight bearing even if he or she already has fixed contractures of the lower extremities. Many people in wheelchairs have limited access to therapy or caregivers who can provide the necessary amount of ranging. Standers integrated with the wheelchair base allow them to perform this important activity on their own and with higher frequency [2]. However, in order to avoid overstretching contracted muscles, it is critical to know the range of motion (ROM) of lower extremity joints during sit-to-stand with a standing wheelchair. Most clinicians are familiar with requirements for obtaining proper static seating posture in a manual wheelchair; few are familiar with dynamic seating posture in a standing wheelchair. As standing wheelchairs are more complex than most manual or electrically power wheelchair, knowing the prerequisite ROM of these joints could help the clinician to make decisions prior to selecting and fitting the user on it.

For these reasons, the purpose of this study was to measure the shear displacement between the body in people with spinal cord injuries and backrest/seat during both the transition of standing up and sitting down by operating an electric powered wheelchair. We also were interested in investigating the ROM of low limb joint and force acting on the knee restraint to determine the prerequisite joint range and the magnate of forces against the knee during sit-to-stand-to-sit cycles. Due to different mechanical and postural conditions, we further hypothesized that standing up and sitting down would affect shear displacement and force acting on the knee block in different ways.

2. Materials and Methods

2.1. Participants. Twenty-four people (17 men and 7 women) with thoracic or lumbar spinal cord injuries (American Spinal Injury Association Impairment Scale type A or B) participated in this study. None of them had ever utilized an electric standing wheelchair before. The mean age was 41.60 ± 11.33 years old, mean weight was 63.92 ± 12.85 kg, and mean length of leg was 81.39 ± 6.84 cm, respectively. They have neither fixed contracture nor skeletal deformities of lower extremities. Known cases of osteoporosis were also excluded. All participants provided written informed consent in accordance with the procedures approved by the Institutional Review Board of the Kaohsiung Medical University Hospital prior to participation in the study.

2.2. Instrumentation. An electrically powered standing wheelchair (Model LY-ESB240, Comfort Orthopedic Co. Ltd., Chia-Yi, Taiwan) was used in this study. The overall dimension of length with footrests, seat height, and width were 113.5 cm, 56 cm, and 45.7 cm, respectively. The maximal angle to hoist a user upright was 80 degrees. No particular shear reduction technology either by back or seat frames was employed in this model. Each participant was required to wear tight fitting clothing (e.g., spandex shorts and sleeveless shirt) to prevent problems with "marker plucking." Reflective markers were placed bilaterally to the surface of the skin over the following bony landmarks: acromion process, great trochanter, lateral epicondyle of the femur, the medial and lateral malleoli, and the fifth metatarsal base. Moreover, eight additional markers were placed on both sides of the backrest frame and seat frame of the wheelchair to determine the backrest and seat reference plane. Three-dimensional marker trajectory data were measured using a six-camera motion analysis system (Qualisys Medical AB, Göteborg, Sweden) at a sampling rate of 120 Hz (Figure 1).

The biaxial force transducers (Model BFT100, Junzhi Engineering Co., Ltd., Kaohsiung, Taiwan) were mounted bilaterally at the knee block to measure the anteroposterior and vertical force acting at the knee restraint with a 10 cm wide nylon strap (Figure 1). Normal ground reaction forces at the footrest were measured using a force plate with a sample frequency of 120 Hz (Model 9286AA, Kistler Instrument, Amherst, NYA).

2.3. Experimental Protocol. At the beginning of testing, the length of the legrest was adjusted when the subject sat

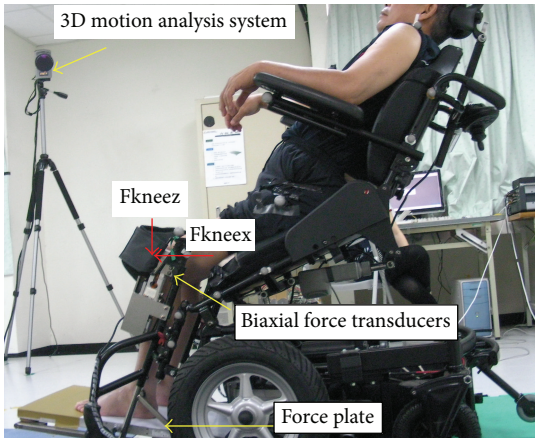

Figure 1: Experimental setting.

comfortably in the wheelchair so that the thighs rested parallel to the seat surface with the foot comfortably placed on the footrest. Afterwards, each subject was asked to transition from sit-to-stand and then back to sit through the operation of the electric powered standing wheelchair. The experimental trail would be stopped immediately for safety concerns and redone again if the participant could not continue sit-to-stand and stand-to-sit transitions, due to orthostatic hypotension or danger of sliding out of the wheelchair. Data were recorded throughout three complete sit-to-stand-to-sit cycles. Between each cycle, a one-minute break was provided.

2.4. Data Analysis. Displacements between the body and wheelchair backrest/seating surface represent the degrees of sliding along the backrest (BS) and sliding along the seat (SS). Therefore, BS was calculated as the distance between the acromion and the backrest frame marker on the backrest reference plane, and SS was defined as the distance between the greater trochanter and seat frame marker on the seat reference plane [14]. Positive values of BS indicated that the location of acromion was displaced upward along the backrest, whereas negative values indicated downward displacements from the initial position. Positive values of SS indicated rearward displacements of the greater trochanter along the seat plane, whereas negative values indicate forward displacements from the initial position. The sliding along the backrest and along the seat at left and right side was then averaged to represent the shear sliding for these two surfaces, respectively. When going through sit-to-stand or stand-to-sit transitions, the body and thigh may shift up/down or fore/back. Therefore, the cumulative values of BS and SS were used to represent the sum of shear displacements during sit-to-stand and stand-to-sit transitions. The difference between the maximum and minimum displacement of BS and SS was calculated as the range of BS and SS to describe the possible range of shear displacements. In addition, from the position of the markers three joint angles were calculated and analyzed: (1) the hip angle between the trunk (link between acromion process and great trochanter) and the thigh (link between the great trochanter and the lateral femoral epicondyle), (2) the knee angle between the thigh and the shank (link between the lateral femoral epicondyle and the external malleolus), and (3) the ankle angle between the shank and foot (link between the external malleolus and fifth metatarsal base). Joint ROM for the hip, knee, and ankle were calculated as the difference between the maximum and minimum joint angle. The magnitudes of force acting on the knee restraint, normal ground reaction force at the ankle joint, were also recorded and normalized with respect to individual body weight. Afterwards, the mean value for above kinematic and kinetic variable across three trials was computed using MATLAB (The MathWorks, Natick, MA) software.

2.5. Statistical Analyses. Descriptive statistics were used to describe selected dependent variables. The comparison of the BS, SS, ROM of low extremity joints, and forces acting on the knee restraint between sit-to-stand and stand-to-sit transitions was made using a paired t-test. All statistical tests were performed using the SPSS for Windows 13.0 (SPSS Inc., Chicago, IL) software package. The level of significance was set to 0.05.

3. Results

3.1. Relative Shear Displacement during Sit-to-Stand-to-Sit Transition. A representative plot of the shear displacements for the back to backrest and thigh to seat was shown in Figure 2. During sit-to-stand transition, the average BS values of each subject were negative, indicating that the upper body was displaced downward, but positive values of SS indicated rearward displacement of the greater trochanter along the seat plane. On the contrary, positive values of BS with negative values of SS were found during a stand-to-sit transition. It was shown that stand-to-sit displaced the upper body into upward direction and thighs into forward direction (Figure 2). The statistical analysis comparing sit-to-stand and stand-to-sit transitions revealed no significant differences in the cumulative values of BS ($P = 0.53$) and SS ($P = 0.07$), but significant differences in the range of BS and SS were found ($P < 0.05$). The range of shear displacements along the backrest plane (BS) was significantly larger during sit-to-stand transition ($P < 0.01$). On the other hand, during stand-to-sit transition the range of shear displacements along the seat plane (SS) was significantly larger ($P = 0.01$).

3.2. The Features of Joint Kinematics during Sit-to-Stand-to-Sit Transition. There are no significant differences in the ROM of hip, knee, and ankle joint when comparing sit-to-stand and stand-to-sit transitions (Table 1). The average ROM of hip and knee joint was 62.1 and 60.8 degrees, respectively, but much less ROM of ankle joint (2~3 degrees) was found. During sit-to-stand transition, the hip and knee joint angle increased in accordance with an increase in the seat-to-back angle. Moreover, the knee angle still increased despite the fact that the backrest was reclined to a flat position (Figure 3). On the contrary, the hip and knee joint angle both

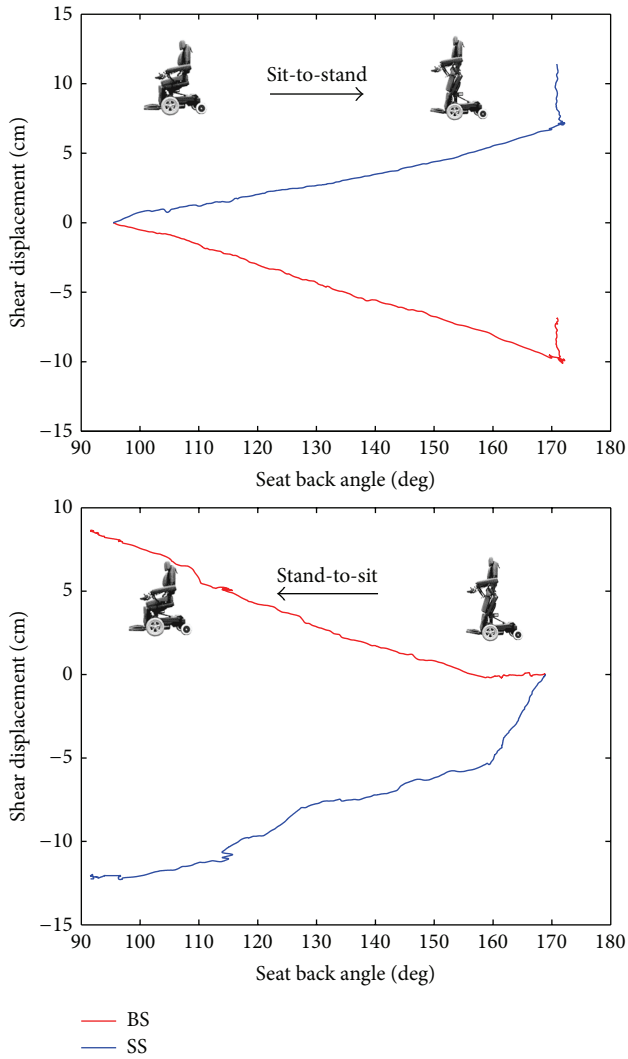

Figure 2: Representative plot (subject number 14, Trial 1) of shear displacement of the user's body sliding along the backrest (BS) and sliding along the seat (SS) as the seat-back angle of the standing wheelchair transformed from sit-to-stand and vice versa.

Table 1: Measurement of sliding along the backrest and seat plane, ROM of low extremities, and forces acting on the knee restraint between sit-to-stand and stand-to-sit transitions.

	Sit-to-stand	Stand-to-sit	P value
Cumulative BS (cm)	−4.84 ± 3.22	4.68 ± 3.57	0.53
Range of BS (cm)	8.72 ± 4.87	6.82 ± 4.04	<0.01*
Cumulative SS (cm)	8.66 ± 2.87	−9.33 ± 3.25	0.07
Range of SS (cm)	8.83 ± 3.21	10.05 ± 3.31	0.01*
ROM of hip joint (degrees)	61.78 ± 11.59	62.57 ± 9.74	0.59
ROM of knee joint (degrees)	60.66 ± 9.02	61.09 ± 7.82	0.71
ROM of ankle joint (degrees)	3.24 ± 1.91	3.14 ± 1.03	0.78
Maximal of resultant force (%BW)	23.51 ± 8.93	19.61 ± 8.04	<0.01*
Average of resultant force (%BW)	9.58 ± 3.04	8.44 ± 3.27	0.01*
Maximal anterior force (%BW)	22.98 ± 8.37	18.62 ± 8.09	<0.01*
Average anterior force (%BW)	9.07 ± 3.02	7.29 ± 3.32	<0.01*
Maximal downward force (%BW)	6.65 ± 2.67	7.48 ± 3.45	0.19
Average downward force (%BW)	2.04 ± 1.22	3.00 ± 2.00	0.01*

* $P < 0.05$.

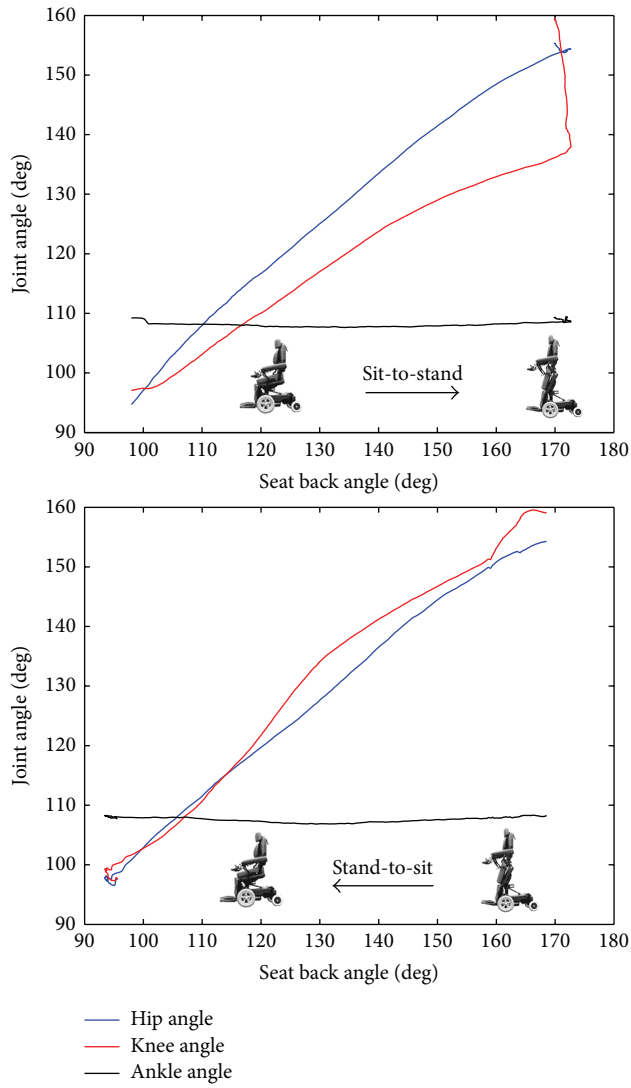

Figure 3: Representative plot (subject number 10) of lower extremity joint angle motion as the seat-back angle of the standing wheelchair transformed from sit-to-stand and vice versa.

decreased with decreasing seat-to-back angle during a stand-to-sit transition (Figure 2). No significant changes in ankle ROM were detected during sit-to-stand-to-sit transition.

3.3. The Force Acting on the Knee Restraint during Sit-to-Stand-to-Sit Transition. Results showed that the resultant forces acting on the knee restraints were significantly larger during sit-to-stand transition than during stand-to-sit transition ($P = 0.01$). During sit-to-stand transition, the resultant forces increased as the knee was extended (Figure 4). At the standing position, large force was applied on the knee to keep legs straight while standing, but by lowering the seat with increasing the knee flexion angle, the resultant forces acting on the knee were decreased when sitting back. Then, a more detailed analysis comparing separately anterior-posterior and upward-downward forces between these two transitions revealed that the maximal and average anterior force, which are against the knee restraint, were significantly greater during sit-to-stand transition ($P < 0.01$). But when moving back to sit the average downward forces were significantly larger ($P = 0.01$). No significant difference in the maximal downward forces was found between these two transitions ($P = 0.19$). Furthermore, while standing up in a standing wheelchair, an average 74.57 ± 13.67% of body weight by the user was attained onto the feet.

4. Discussion

Standing wheelchairs are confronted with the problem that relative displacements between the backrest/seat surfaces and the wheelchair user's body can occur during standing up or sitting down. A relative displacement between the wheelchair and the user's body is produced when the axis of rotation of the hinge mechanism does not correspond to the axis of rotation of the hip/knee joint. In reality, the situation is somewhat more complex because these anatomical joints

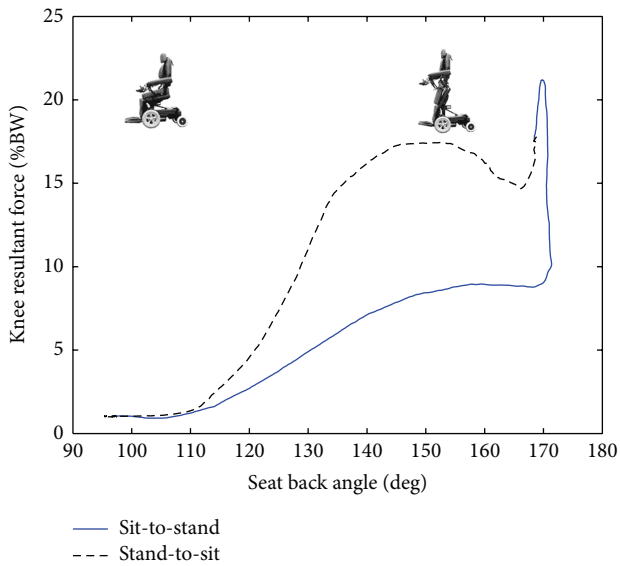

Figure 4: Representative plot (subject number 10) of the resultant forces acting on the knee as the seat-back angle of the standing wheelchair transformed from sit-to-stand and vice versa.

somehow do not perform a pure rotational movement. Our results showed that the range of shear displacements were up to 8.7 cm for between the user's body and the wheelchair backrest/seat surfaces during sit-to-stand. When the standing wheelchair transformed from sitting to standing, the user's back slid down, and the greater trochanter was displaced rearward relative to the seat surface. On the other hand, when wheelchair sat back down, up to 7 cm and 10 cm of displacement could take place between the user's body and the wheelchair's backrest and seat surface, respectively.

Specifically, during transition from sitting to standing with opening the seat-to-back angle, standing up was performed with the seat lift mechanism being against gravity. The gravity would cause the user's body to slide down and forward along backrest and seat plane. If the sliding happened without any other restrictions, the user was likely to slide and fall out of the wheelchair. Therefore, the knee restraint which was placed anteriorly to the knees limited forward movement of the thighs as a "block" to prevent sliding and helped to reposition the knee from a flexed position to a straight position while standing. When standing with legs straight, this will further prevent the user sliding down along backrest and seat plane. As a result, the acromion slid down from the initial position in the beginning and then rebounded upward due to keeping the knee straight. At the same time, the greater trochanter was also slid rearward, close to the backrest when the wheelchair raised users from sitting to standing as shown in Figure 2.

On the contrary, during the transition from standing to sitting with closing the seat-to-back angle, the backrest was more likely going to a squeeze posture and moved the user into upward and forward direction. The upward displacement of acromion and forward displacement of greater trochanter, in consequence, could be found, as shown in Figure 2. Furthermore, there was more significant forward displacements of the greater trochanter than acromion in upward direction. When the standing wheelchair tried to move back into the sitting position, the act of moving the backrest upright often tried to push the user out of the seat even further unless the user was repositioned. No matter the transition from sitting to standing positions or vice versa, relative displacements between the body's position with respect to the backrest and seat surface was attributed to this dynamic posture change. Because the backrest and seat of the wheelchair slid against the user's back and thigh when standing and returning to an seated position, the use of any type of shear reduction technology on these two surface planes might reduce the shear displacements and further prevent the risk of skin breakdown and pressure sores.

The standing wheelchair usually provided three-point stabilization in a fully upright or vertical position, with supports at the lower torso, hips, and knees. A backrest upheld the lower torso, a seat lift supported hip posteriorly, and knee pads stabilized the knees anteriorly [15]. This standing mechanism allowed individuals with complete paralysis to stand passively and subsequently load their lower extremity. However, there were a few limitations as to who can use a standing wheelchair. The biggest was that the user's legs and joints must be able to support the user's torso weight without pain, discomfort, or doing any damage and that the user's hips and knees must have a certain range of motion as they would be extended/flexed during use. Our results showed that a minimum of 60 degrees of ROM in the hip and knee with few degrees of ankle ROM was required for safety of operation of a standing wheelchair. Since standing wheelchairs are powerful devices, they extended a user's hip and knee joint mechanically to provide posture change, but the amount of extension has to be limited, especially in the case of a client without sensation who would be unable to detect and indicate harmful stretches. It may cause harm if attempting to overstretch contracted muscles within limited ROM of hip and knee joint. Based on our findings, it was recommended that at least 60 degrees of ROM in the hip and knee was necessitated in a standing wheelchair. Less ROM of the ankle joint was wanted.

Our results also showed that the maximal resultant forces acting on the knee restraints could reach 23.5% of BW and were significantly larger during sit-to-stand transition than stand-to-sit transition. Furthermore, the magnitudes of force in anterior direction were also significantly larger. This indicated that knee restraints were used as a block to secure the knee and applied pressure to the upper shin as the wheelchair mechanisms worked in concert to raise the body to the standing position. The user was raised by a movement which was produced by the reaction force at the knee and the raising seat. On the contrary, during transitioning from standing to sitting, the knee restraints were more likely to support lower limbs and prevent accidental sliding downward. Therefore, more downward forces were observed, as compared with sit-to-stand transition. Regardless of whether sitting or standing in a standing wheelchair, knee blocking is critical and necessary to keep the knees straight without

buckling. It was not possible to stand up or move back to sit without the knee restraints.

Many wheelchair users experience significant reduction in bone mineral density due to the lack of weight bearing which results in osteoporosis and a risk of fractures. Therefore, it is necessary to align the pivot mechanism for the knees with the center of joint rotation for the user's knee. Misalignment can place stress on the knee and tibia, which may lead to a fracture or joint laxity [13]. Our results showed that the average maximal resultant forces acting on the knee could reach 24% of BW (the range of 23.5% BW~19.6% BW). Dudley-Javoroski and Shields reported that in the first year post-SCI, between 15 and 35 percent of bone mineral density was lost in the region of the knee joint at greatest risk for fracture: the distal femur and proximal tibia [16]. Hence, it would more likely cause fracture around the knee joint if the standing wheelchair user had a lower bone mineral density. Careful consideration needs to be given to who the user of the standing wheelchair is.

Standing wheelchairs also promote weight bearing, which is essential to maintain bone density in the lower extremities. Our results indicated that an average weight-bearing load of 75% of body weight could be reached while standing upright in this model of standing wheelchair. This finding was consistent with previous studies using other standing devices. Although the amount of weight distributed through the lower limbs could vary depending on the standing device used, previous studies showed that weight bearing ranged from 68% to 85% of body weight [17, 18]. Ideally, it was expected that most weight bearing by the user of the standing wheelchair should be reached in the right circumstances. However, as sliding occurred between the user's body and backrest/seat plane, it was more challenging to achieve full hip and knee extension in a standing wheelchair than in a standing frame. In such situations, more weight was translated anteriorly into the proximal tibia against the knee restrains and posteriorly at the pelvis against the seat surface. Consequently, partial weight bearing was borne by the combination of the seating surface and knee restrains as found in our data. Moreover, during sit-to-stand transition, the participants in our study were asked to relax and rest their arms on the armrest for comfort. It was possible that the participants may grasp the armrest to particularly support their weights for stability during standing. A similar phenomenon was observed by Bernhardt et al. They indicated that supporting the arms on the standing frame tray while standing reduced the weight bearing by approximately 10% body weight. So, higher percentage of weight bearing could be obtained while standing in the standing wheelchair with their arms unsupported. Nevertheless, a large proportion of their body weight (approximately 75%) was still borne through the lower limbs even though their arms were resting on the armrest.

The scope of this study was limited because only one particular model of powered standing wheelchair was tested. There are numerous types of standing wheelchairs available on the market, including a manual standing wheelchair, a manual wheelchair with powered standing assist, or a fully powered wheelchair. Many manufactures now incorporate reclining mechanisms into the standing wheelchair mechanisms that are designed to eliminate sliding displacement during sit-stand-sit transition. Different models of wheelchair may incorporate these features including tilt-in-space, reclining, standing, and even elevating seat. The combination of these features might provide a smooth and low shear transition between standing and seating functions. Our study only demonstrated the possible shear displacements without implementing shear reduction technologies. Further research is needed to measure the effect of these shear reduction technologies for reducing shear and friction forces between the user's body and backrest/seat surface.

5. Conclusions

In spite of the numerous benefits, a standing wheelchair might be contraindicated without appropriate assessment. Our results showed that the shear displacement occurred between the user's body and the backrest/seat surfaces when using a stand-up function of the powered standing wheelchair. The shear forces from the backrest and seat surface while transitioning from sitting to standing and vice versa could result in a pressure sore. Furthermore, a user with fixed contractures of the lower extremities or existing osteoporosis was not an appropriate candidate for a standing wheelchair. Clinical practitioners should be aware of these contraindications and precautions.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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Clinical Study

Temporal Parameters Estimation for Wheelchair Propulsion Using Wearable Sensors

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Due to lower limb paralysis, individuals with spinal cord injury (SCI) rely on their upper limbs for mobility. The prevalence of upper extremity pain and injury is high among this population. We evaluated the performance of three triaxis accelerometers placed on the upper arm, wrist, and under the wheelchair, to estimate temporal parameters of wheelchair propulsion. Twenty-six participants with SCI were asked to push their wheelchair equipped with a SMART^{Wheel}. The estimated stroke number was compared with the criterion from video observations and the estimated push frequency was compared with the criterion from the SMART^{Wheel}. Mean absolute errors (MAE) and mean absolute percentage of error (MAPE) were calculated. Intraclass correlation coefficients and Bland-Altman plots were used to assess the agreement. Results showed reasonable accuracies especially using the accelerometer placed on the upper arm where the MAPE was 8.0% for stroke number and 12.9% for push frequency. The ICC was 0.994 for stroke number and 0.916 for push frequency. The wrist and seat accelerometer showed lower accuracy with a MAPE for the stroke number of 10.8% and 13.4% and ICC of 0.990 and 0.984, respectively. Results suggested that accelerometers could be an option for monitoring temporal parameters of wheelchair propulsion.

1. Introduction

According to the 2010 Survey of Income and Program Participation (SIPP), about 3.6 million people aged 15 years and older in the USA use a wheelchair [1]. Most of these individuals use a manual wheelchair for mobility [2]. Manual wheelchair users often rely on their upper extremities for almost all activities of daily living (ADLs). Some of their daily activities such as wheelchair propulsion and transfers require high forces and repetitiveness of upper extremities movements. Therefore, it is not surprising that the incidence of upper extremity pain and injury among manual wheelchair users is high, ranging from 49% to 78% [3–11].

Given the negative impact that upper extremity pain and injury may have on the lifestyle and quality of life of manual wheelchair users [9, 12–14], the Consortium for Spinal Cord Medicine published the monograph, *Preservation of Upper Extremity Function Following Spinal Cord Injury: A Clinical Practice Guideline for Health Care Professionals*, where it provides concise ergonomic and equipment recommendations based on the review of published evidence [15]. The guideline recommends reducing the frequency of repetitive upper limb tasks, minimizing forces required to complete tasks, and minimizing extremes of wrist and shoulder motions. It also makes recommendations on wheelchair propulsion techniques such as reducing the stroke number and push frequency.

Temporal parameters of wheelchair propulsion such as the stroke number and push frequency have been quantified in laboratory settings using motion capture systems and SMART^{Wheels} (Three Rivers Holdings, LLC) a force sensing wheel that can replace the wheelchair wheel to collect propulsion parameters [5, 16–18]. Unfortunately, due to the cost and intricate settings, these valuable tools are not appropriate for assessing upper extremity movement in the home and community environment. Therefore, the repetitiveness of upper extremity movement for wheelchair propulsion out of clinical settings is unclear. With the recent advancement of sensors and miniature technologies, accelerometers emerge as a possible solution for monitoring wheelchair propulsion

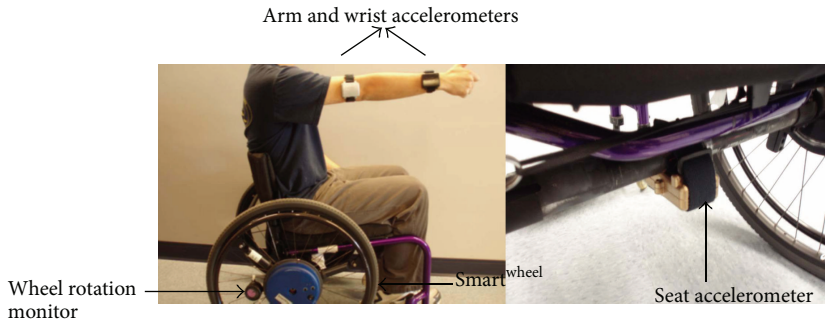

Figure 1: Instrumentation setup.

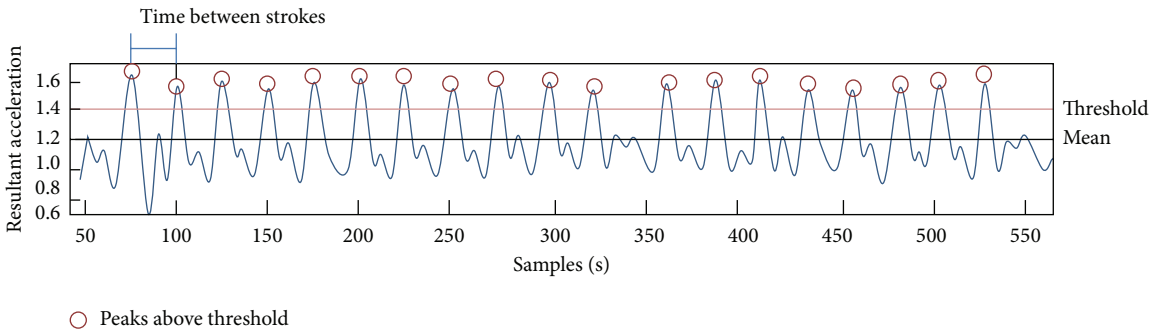

Figure 2: Visual example for stroke number and push frequency estimation.

parameters, potentially contributing to the understanding and prevention of upper extremity pain and injury among manual wheelchair users.

Previous studies have used accelerometers and other sensors to track gross mobility of manual wheelchair users. A pilot study conducted by Kumar et al. used a customized data-logging device to determine driving characteristics including distance, speed, and driving time of 19 power wheelchair soccer players [19]. A similar study conducted by Coulter et al. used two triaxial accelerometers placed on the wheels of a wheelchair to estimate gross mobility of 14 manual wheelchair users with spinal cord injury (SCI). The results showed that the accelerometers were able to recognize wheelchair propulsion episodes with an overall accuracy of 92% [20]. A study conducted by Gendle et al. investigated the revolutions, duration, and direction of movements. They found that the activity counts from the accelerometer were significantly different between light and moderate effort [21]. Other researchers have evaluated the performance of accelerometers in detecting manual wheelchair users' activities. A study conducted by Postma et al. used six two-axis accelerometers placed around the wrists, thighs, and along the sternum, respectively, to detect wheelchair propulsion episodes and its intensity from a range of ADLs among 10 manual wheelchair users. All six accelerometers were wired to a data recorder attached to the waist. Wheelchair propulsion episodes were identified using a wheelchair detection knowledge based on different body postures. The study showed that the six accelerometers were able to detect wheelchair propulsion episodes with an overall agreement of 92%. However, having 6 accelerometers on the body may prevent the user from moving freely in a natural environment [22]. Despite the fact that gross mobility and its intensity are, to some extent, indicative of manual wheelchair users' upper extremity movements, they cannot tell the exact amount and repetitiveness of upper extremity movements for wheelchair propulsion.

Knowing the repetitiveness of upper extremity movements for wheelchair propulsion that occur on a daily basis could be important for understanding and preventing upper extremity pain and injury. However, research looking into using wearable sensors to directly estimate temporal parameters of wheelchair propulsion is limited. A study conducted by Koontz et al. estimated temporal parameters of wheelchair propulsion including push time, propulsion time, and recovery time based on hand acceleration collected via a 6-camera VICON motion analysis system among 29 manual wheelchair users. Position of the third metacarpal phalangeal joint was converted into instant velocity and instant acceleration. Push, propulsion, and recovery time were estimated by detecting acceleration sign change. Estimated parameters were compared with temporal parameters obtained from the SMART^{Wheels} (Three Rivers Holdings, LLC). Results showed high intraclass correlation between the estimated and criterion measures [23]. This study showed the feasibility of using hand acceleration to determine propulsion parameters. However, the hand acceleration in this study was derived from the 6-camera VICON system instead of a wearable accelerometer. A study conducted by Turner investigated the use of an accelerometer placed beneath the chair and a wheel-mounted magnet to detect wheelchair propulsion

Table 1: Participant demographics.

Demographic variables	Mean ± standard deviation
Sex	
Female	6
Male	20
Age (years)	40 ± 14
Weight (lb.)	159 ± 41
Manual wheelchair usage (years)	13 ± 8
Injury level range	
Paraplegia (T4 and below)	20
Tetraplegia (T3 and above)	6
Self-reported pain (WUSPI)	7 ± 10

Table 2: Criterion and estimated stroke number.

	Video	Arm	Wrist	Seat
Level surface	24.6 ± 4.1	24.6 ± 4.0	24.6 ± 4.6	25.0 ± 4.3
Sloped surface	18.1 ± 1.1	17.2 ± 1.3	17.0 ± 1.4	17.7 ± 2.0
Overall	22.4 ± 3.6	22.2 ± 3.6	22.1 ± 4.0	22.6 ± 3.8

Table 3: Criterion and estimated push frequency (stroke/sec).

	SMART^{Wheel}	Arm	Wrist	Seat
Level surface	0.95 ± 0.15	0.93 ± 0.09	0.94 ± 0.09	0.82 ± 0.19
Sloped surface	1.06 ± 0.09	1.02 ± 0.04	1.03 ± 0.13	0.94 ± 0.22
Overall	0.98 ± 0.11	0.96 ± 0.06	0.98 ± 0.09	0.86 ± 0.18

parameters including the stroke number, push frequency, distance, and speed. Ten manual wheelchair users were asked to propel their wheelchair on indoor and outdoor surfaces. Estimated parameters were compared with criterion values obtained from OptiPush wheels. Results showed the average percentage of errors were −1.0% for the stroke number and −1.7% for push frequency [24].

The purpose of this study is to assess the validity of a triaxis accelerometer placed at three locations (i.e., wrist, upper arm, and underneath the wheelchair seat) in estimating temporal parameters of wheelchair propulsion including the stroke number and push frequency. The information obtained can guide the use of accelerometers for monitoring temporal parameters and upper extremity movements during wheelchair propulsion.

2. Material and Methods

2.1. Study Participants. The Institutional Review Board at the University of Pittsburgh approved this study. A total of 26 manual wheelchair users with SCI volunteered and provided informed consent prior to their participation in the study. Subjects were identified through the IRB approved wheelchair user registries developed by the Human Engineering Research Laboratories (HERL) and the Department of Physical Medicine and Rehabilitation at the University of Pittsburgh. Subjects were included in the study if they (1) were 18 years of age or greater; (2) use a manual wheelchair as a primary means of mobility; and (3) have SCI. Subjects were excluded if they were unable to tolerate sitting for 2 hours and/or have upper limb pain that limits their mobility.

2.2. Instrumentation. Subjects were fitted with four monitoring devices and a SMART^{Wheels} (Three Rivers Holdings, LLC). As shown in Figure 1, the four monitoring devices included a custom wheel rotation monitor attached to the wheelchair wheel and three off-the-shelf triaxis accelerometers (Shimmer Research, Dublin) worn on the dominant upper arm, dominant wrist, and underneath the wheelchair seat, respectively.

(i) The wheel rotation monitor was developed at the HERL. It is a lightweight and self-contained device that can be easily attached to the wheelchair's wheel without any modifications to the wheelchair. It tracks the wheel rotation through three reed switches mounted 120° apart on the back of the printed circuit board and a magnet mounted at the bottom of a pendulum. As the wheel rotates and exceeds 120° of rotation, one of the reed switches is triggered, and a date and time stamp is recorded. This information can be further processed to obtain the distance, speed, and time of movement [25]. The wheel rotation monitor has been used in previous studies to collect mobility characteristics of manual wheelchair users with different diagnoses [19, 26, 27].

(ii) The triaxis accelerometer (Shimmer Research, Dublin) used in this study is a small low-power device that can record the motion data into a micro SD card. The two upper arm accelerometers were sampled at 20 Hz and the accelerometer underneath the seat was sampled at 60 Hz.

(iii) The SMART^{Wheels} (Three Rivers Holdings, LLC) is a 3D force and torque-sensing wheel that measures push forces, push smoothness, push frequency, speed, and push length in every push cycle. It is sampled at 240 Hz. Subjects' wheelchair wheels were replaced with a SMART^{Wheels} (Three Rivers Holdings, LLC) at the dominant side and a dummy wheel at the other side to balance the weight of the SMART^{Wheels} (Three Rivers Holdings, LLC). The use of SMART^{Wheels} (Three Rivers Holdings, LLC) did not change the camber or the axle position.

2.3. Experimental Protocol. Subjects were asked to pay two visits to HERL with each visit lasting about 2.5 hours. During the first visit, subjects completed a demographics survey and the Wheelchair Users Shoulder Pain Index Questionnaire (WUSPI). The WUSPI questionnaire measures shoulder pain based on 15 questions using a 10 cm visual analogue scale, resulting in a total score from 0 (no pain) to 150 (extreme pain) [28]. After subjects were fitted with the instrumentation described in the previous section, they were asked to propel their own wheelchairs on two surfaces including a level surface of 33 meters and a sloped surface of 15 meters with 5 degrees of incline. A total of 24 level-surface trials performed

Table 4: Mean absolute error (MAE) and mean absolute percentage of error (MAPE) for the stroke number.

	MAE			MAPE %		
	ARM	WRIST	SEAT	ARM	WRIST	SEAT
Level surface	1.7 ± 1.5	2.4 ± 2.3	2.9 ± 3.5	7.7 ± 6.6	11.0 ± 10.2	13.5 ± 16.4
Sloped surface	1.5 ± 1.2	1.8 ± 1.3	2.4 ± 2.1	8.6 ± 7.0	10.3 ± 7.9	13.4 ± 11.8
Overall	1.6 ± 1.4	2.2 ± 2.1	2.7 ± 3.2	8.0 ± 7.1	10.8 ± 9.8	13.4 ± 15.6

Table 5: Mean absolute error (MAE) and mean absolute percentage of error (MAPE) for the push frequency.

	MAE			MAPE %		
	ARM	WRIST	SEAT	ARM	WRIST	SEAT
Level surface	0.1 ± 0.1	0.2 ± 0.2	0.3 ± 0.2	16.1 ± 16.7	21.5 ± 21.4	25.4 ± 16.9
Sloped surface	0.1 ± 0.1	0.1 ± 0.1	0.2 ± 0.2	6.4 ± 4.6	8.0 ± 6.1	21.8 ± 14.6
Overall	0.1 ± 0.1	0.1 ± 0.1	0.2 ± 0.2	12.9 ± 15.1	17.2 ± 19.3	24.2 ± 16.6

Table 6: Stroke number and push frequency intraclass correlation coefficient (ICC 3, 1).

		ICC	95% CI	*P* value
Stroke number	ARM	0.994	0.988~0.997	<0.001
	WRIST	0.990	0.980~0.995	<0.001
	SEAT	0.984	0.972~0.991	<0.001
Push frequency	ARM	0.916	0.843~0.953	<0.001
	WRIST	0.889	0.802~0.936	<0.001
	SEAT	0.690	0.071~0.868	<0.001

CI: confidence interval.

at self-selected speed, low speed, and fast speed, and 12 sloped-surface trials at a self-selected speed were completed by each subject. All trials were videotaped using a hand-held digital video recorder.

During the second visit, participants were first asked to perform the propulsion trials as detailed for the first visit. Participants were then asked to complete a training session where they watched a multimedia instructional program on a laptop computer that aimed to teach appropriate propulsion techniques. The multimedia instructional program was developed by a previous study based on propulsion biomechanics literature and the Clinical Practice Guideline, which emphasized reducing push frequency and increasing push angle [29]. Examples of good and bad techniques were provided. After subjects practiced the propulsion techniques following the video training, they were asked to perform the same propulsion trials. This visit allowed us to assess if the accelerometers were capable of capturing propulsion changes due to training.

2.4. Data Collection and Analysis. Videos recorded during the two visits served as the criterion measure of the stroke number. Two investigators independently counted the stroke number for each propulsion trial, and video footages were reexamined when there was a discrepancy between the two investigators. The criterion push frequency was directly obtained from the SMART^{Wheels} (Three Rivers Holdings, LLC).

Data from the wheel rotation monitor was converted to the wheel speed, which was used to identify wheelchair propulsion episodes and segment the acceleration data for each trial. Acceleration signals obtained from the accelerometers on the wrist, upper arm, and underneath the seat were filtered to remove high frequency noise using an 8th-order Butterworth low-pass filter. Butterworth filters have response characteristics that are appropriate for filtering wheelchair propulsion kinematic data as shown in previous studies [30–32]. Butterworth filters are commonly used to filter noisy signals because they introduce almost no distortion on the pass band while zeroing the noise on higher frequencies. A higher order filter (8th-order) was used to narrow the transition bandwidth which is wide in this type of filters [33]. The cutoff frequency was defined by the fundamental frequency calculated based on each propulsion trial with values ranging from 2 to 6 Hz. For the arm and wrist accelerometers, the resultant accelerations (the vector sum of three directions) were used to obtain the stroke number. For the seat accelerometer, only the longitudinal component (parallel to the propulsion direction) was used. An algorithm was developed to extract the stroke number from each propulsion trial. The algorithm first calculated a threshold defined as the mean acceleration plus 0.5 standard deviation over each trial. The stroke number was then counted as the number of acceleration peaks over the established threshold. Push frequency was calculated as the mean propulsion time between each two consecutive strokes. Figure 2 shows a visual example of the stroke number and push frequency estimation.

Custom MATLAB (Version 7.11.0 R2010b, The Mathworks, Inc., USA) programs were used to process the acceleration signals.

The estimated stroke number and push frequency from the three accelerometers were compared with the criterion by calculating the mean absolute error (MAE) which was calculated as the average of the absolute difference between the estimated and the criterion, and mean absolute percentage of error (MAPE) calculated as the average ratio between the absolute difference and the criterion MAE $= (1/n)\sum_{i=1}^{n}|E_i - C_i|$ and MAPE $= (1/n)\sum_{i=1}^{n}(|E_i - C_i|/C_i)$, where E_i is

Table 7: Criterion and estimated stroke number before and after training, change, and P value.

		Stroke number				Change		P value
		Before		After				
		Mean		Mean		Mean		
Video	LS	25.5	±7.6	22.3	±5.7	−3.2	±0.96	0.09
	SS	18.2	±6.2	16.7	±4.4	−2.1	±1.13	0.68
	OA	21.8	±7.8	19.6	±5.8	−2.6	±1.02	0.16
Arm	LS	25.2	±7.2	22.7	±5.6	−2.5	±0.94	0.17
	SS	17.1	±5.9	15.8	±4.8	−1.9	±1.06	0.70
	OA	21.2	±7.7	19.3	±6.2	−2.2	±0.98	0.26
Wrist	LS	25.0	±7.0	23.2	±5.7	−1.8	±0.93	0.30
	SS	16.9	±6.0	16.1	±4.1	−1.4	±1.08	0.93
	OA	20.9	±7.7	19.7	±6.1	−1.6	±0.98	0.47
Seat	LS	26.3	±8.9	25.2	±6.4	−1.1	±0.84	0.12
	SS	18.0	±6.7	17.3	±5.6	−0.7	±1.06	0.66
	OA	22.1	±8.9	21.3	±6.8	−0.8	±0.95	0.19

LS: level surface propulsion trials, SS: sloped surface propulsion trials, and OA: level surface and sloped surface combined.

Table 8: Criterion and estimated push frequency before and after training, change, and P value.

		Push frequency (stroke/sec)				Change		P value
		Before		After				
		Mean		Mean		Mean		
SMW	LS	0.96	±0.16	0.88	±0.16	0.09	±0.12	0.06
	SS	1.13	±0.18	0.98	±0.15	0.18	±0.20	0.001
	OA	1.04	±0.19	0.93	±0.16	0.13	±0.17	0.001
Arm	LS	0.94	±0.14	0.89	±0.13	0.05	±0.13	0.197
	SS	1.05	±0.14	0.95	±0.13	0.14	±0.19	0.007
	OA	1.00	±0.15	0.92	±0.14	0.09	±0.17	0.007
Wrist	LS	0.93	±0.11	0.90	±0.12	0.03	±0.09	0.327
	SS	1.08	±0.19	0.98	±0.14	0.13	±0.21	0.024
	OA	1.00	±0.17	0.94	±0.13	0.08	±0.17	0.022
Seat	LS	0.84	±0.17	0.77	±0.14	0.07	±0.14	0.134
	SS	0.98	±0.30	0.87	±0.29	0.04	±0.40	0.081
	OA	0.91	±0.25	0.82	±0.19	0.06	±0.30	0.028

SMW: Smart^{Wheel}, LS: level surface propulsion trials, SS: sloped surface propulsion trials, and OA: level surface and sloped surface combined.

Table 9: Stroke number and push frequency ICC (3, 1) before and after training.

		ICC	95% CI	P value
Stroke number	ARM	0.980	0.964~0.989	<0.001
	WRIST	0.969	0.916~0.986	<0.001
	SEAT	0.870	0.773~0.925	<0.001
Push frequency	ARM	0.856	0.684~0.899	<0.001
	WRIST	0.822	0.711~0.923	<0.001
	SEAT	0.568	0.248~0.752	<0.001

CI: confidence interval.

the estimated measure and C_i is the criterion measure. In addition, the intraclass correlation coefficients (ICC 3, 1) were used to assess their agreements. Bland-Altman plots were performed to provide a visual analysis of their agreements. 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 [34]. Propulsion trials during the first and the second visit were compared to assess the agreement between the estimation and the criterion.

Intraclass Correlation Coefficients (ICC 3, 1) were calculated to assess the validity of the accelerometers in detecting changes after training. Independent paired t-test was performed to evaluate significant differences before and after training. All statistical analysis was performed using SPSS software (ver. 18.0, SPSS Inc., Chicago, IL, USA).

3. Results

The demographics of the participants are described in Table 1. Tables 2 and 3 show the mean and standard deviation of the

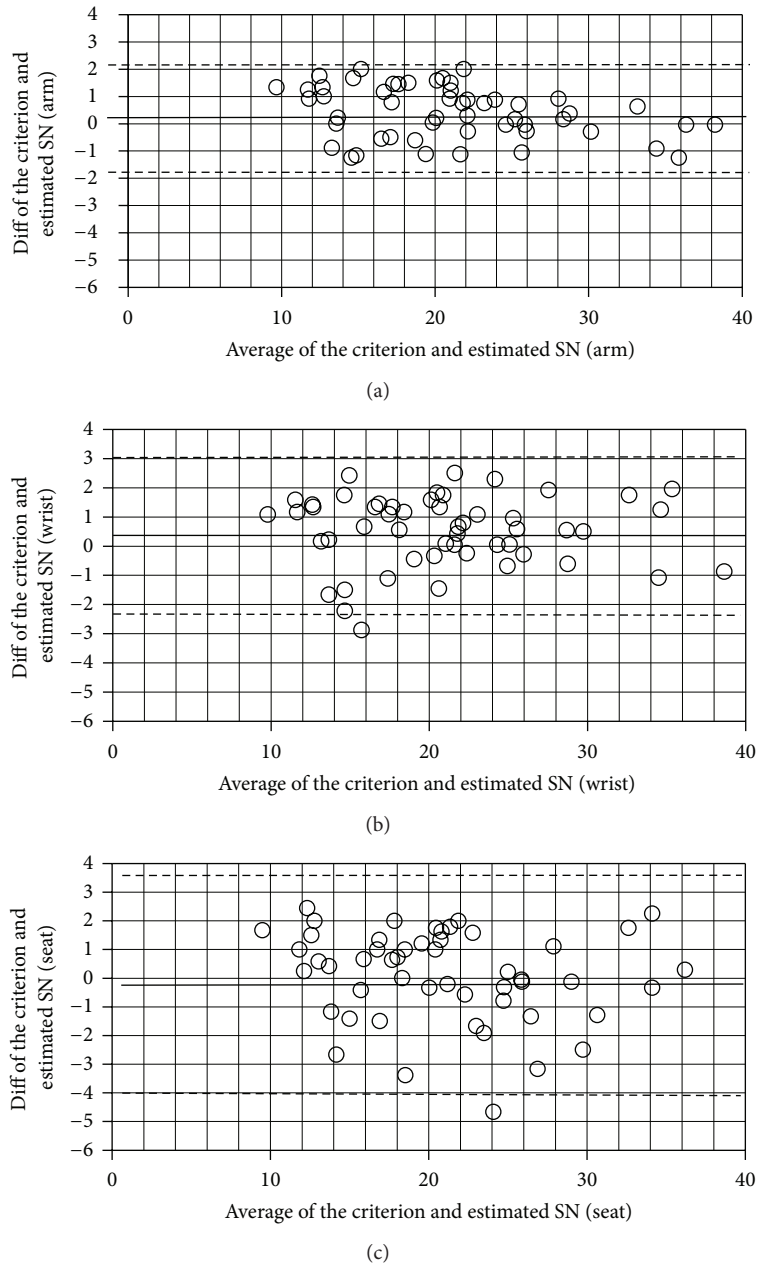

Figure 3: Stroke number Bland-Altman plots from the arm (a), wrist (b), and seat (c) accelerometers.

criterion and estimated stroke number and push frequency. Table 4 shows the MAE and the MAPE between the criterion and estimated stroke number from each accelerometer. Table 5 shows the MAE and MAPE between the criterion and estimated push frequency from each accelerometer. Table 6 shows the ICC (3, 1) between the criterion and estimated temporal parameters for each accelerometer. Table 7 shows the criterion and estimated stroke number before and after training, changes, and P values. Table 8 shows the criterion and estimated push frequency before and after training, changes, and P values. Table 9 shows the ICC (3, 1) for the criterion and estimation before and after training. All variables were calculated for the level surface trials, the sloped surface trials, and the overall trials. Figures 3 and 4 show the Bland-Altman plots between the criterion and estimated stroke number and push frequency from each accelerometer, respectively.

4. Discussion

This study provides insight into the usage of portable devices (e.g., triaxis accelerometers and wheel rotation monitor) to track upper extremity movements for wheelchair propulsion. The small discrepancies between the criterion and estimated parameters shown in Tables 4 and 5 suggest that wearable

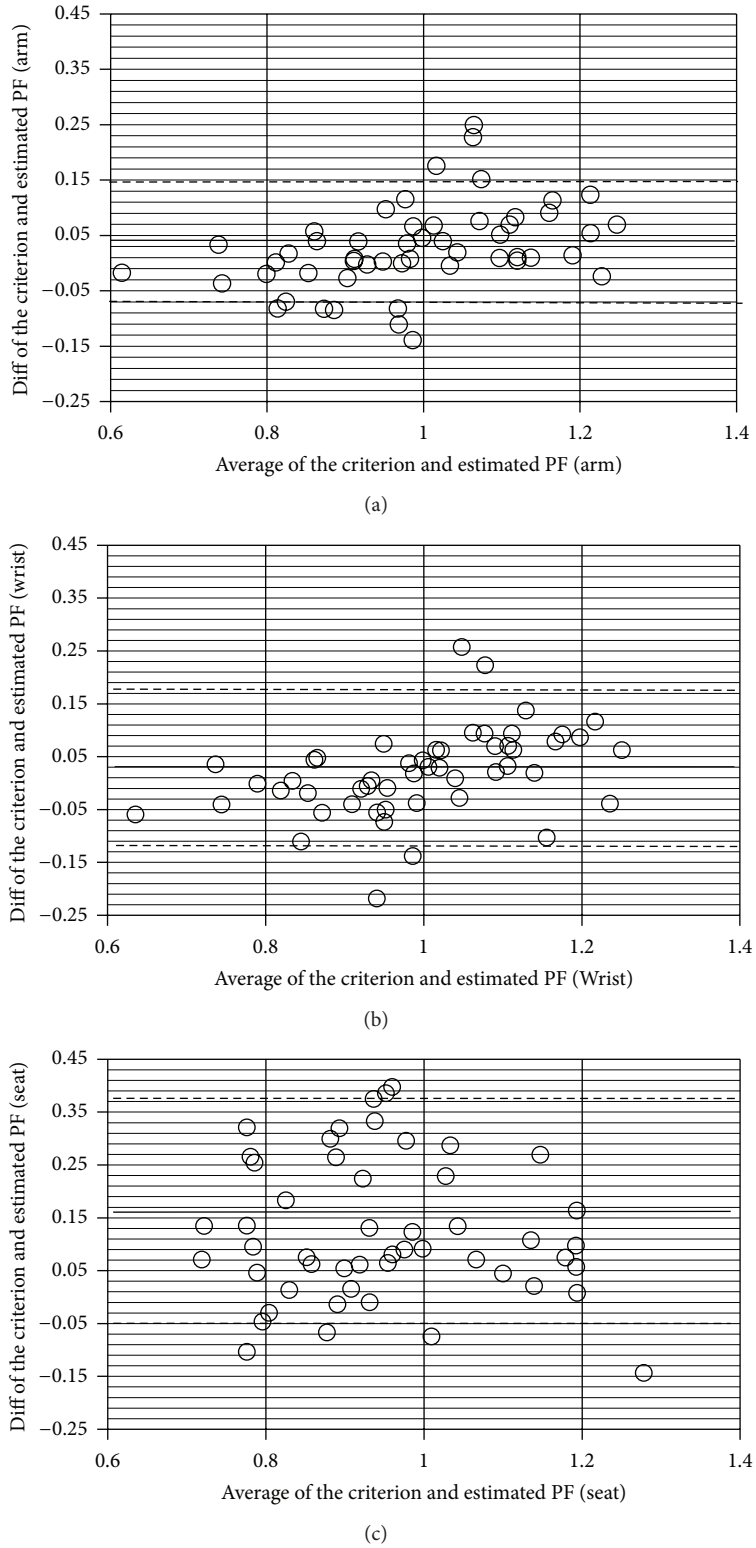

Figure 4: Push frequency Bland-Altman plots from the arm (a), wrist (b), and seat (c) accelerometers.

sensors have the potential to not only detect gross mobility levels of manual wheelchair users [20, 22, 35] but also to quantify the quality of upper extremity movements for wheelchair propulsion in terms of the repetitiveness.

In terms of estimating the stroke number and push frequency, the arm accelerometer showed the highest accuracy among the three accelerometers with a MAPE of 8.0% for stroke number and 12.9% for push frequency, indicating that the upper arm could be a good location for detecting temporal parameters of wheelchair propulsion. The wrist accelerometer showed higher MAPE than the arm accelerometer and this could be because the wrist accelerometer can be more sensitive to small upper extremity movements, possibly leading to the increased error. The seat accelerometer showed the lowest accuracy with a MAPE of 13.4% for the stroke number and 24.2% for the push frequency. The estimation errors for the seat accelerometer were greater than the study by Turner where an accelerometer was placed beneath the wheelchair seat to estimate the stroke number and push frequency among 10 manual wheelchair users. Unfortunately, the data analysis results were not described in detail. The study only reported an average percent error (i.e., −1.0% for stroke number and −1.7% for push frequency) instead of the MAPE averaged by each trial of each subject. An average percent error only indicates the estimation bias and may not be sufficient to show the estimation accuracy, as the positive and negative estimation errors from the trials may cancel each other, resulting in smaller overall errors [24].

Compared with the stroke number estimation, push frequency estimation was less accurate, which could be due to the estimation of the total cycle time comprised of push and recovery phases. The estimation algorithm based on the accelerometer signals was able to identify the push phase more accurately but unable to accurately determine the end of recovery phases, possibly leading to the inaccuracy when estimating the cycle time.

Tables 7 and 8 showed that subjects reduced the stroke number and push frequency after the propulsion training program, but there was only statistically significant difference on the push frequency along the upsloped surfaces. The accelerometers on the arm and wrist were also able to detect the difference. The ICC (3, 1) values in Table 9 also show that the accelerometers especially the ones on the arm and wrist were consistent with the criterion measures for detecting changes in stroke number and push frequency after the propulsion training. The responsiveness of the accelerometer and its estimation algorithm for propulsion parameters could potentially enable the evaluation of training interventions out of clinical settings, contributing to the preservation of upper limb functions of manual wheelchair users with SCI [36].

Considering the negative impact that upper extremity pain and injuries can have on manual wheelchair users with SCI, it is important to monitor and understand how the use of upper limbs during wheelchair propulsion and other ADLs are related to such pain and injury. The Clinical Practice Guideline on the Preservation of Upper Limb Function Following Spinal Cord Injury stresses the importance of reducing the frequency of repetitive upper limb tasks [15]. This study could result in a potential tool that can monitor the actual usage of upper extremities in terms of the repetitiveness during wheelchair propulsion and provide clinical professionals and researchers with an indication of activity levels as well as propulsion skills of manual wheelchair users in their daily life. Results in this study suggest that the use of accelerometers and wheel rotation monitors could potentially provide an objective measure of the repetitiveness of upper extremity movements of wheelchair users. This information may help clinicians to better understand and prevent upper extremity pain an injury among manual wheelchair users.

With the accelerometry technology getting cheaper and smaller, it is also possible to provide near real-time feedback to manual wheelchair users about their upper limb use and repetitiveness, further contributing to the prevention of upper limb pain and injury among this population. We envision the tools described in this study will be used during everyday living as follows: the wheelchair rotation monitor attached to the wheelchair wheel continuously monitors the wheelchair movement and determines the wheelchair propulsion episodes based on the wheelchair speed. If the wheelchair is determined to be moving continuously for a certain amount of time (e.g., 30 seconds), the accelerometer data for that period will be analyzed using the method described earlier, yielding the estimated stroke number and push frequency. The method can also accommodate the variations in propulsion style and speed within and between individual users by using their own movement data as reference. The estimated parameters could be used to provide feedback to the user in near real-time (if paired with a display or smartphone) or summary format to inform their progress over time. The push efficiency calculated by stroke number per feet or meter could also be obtained as an indicator of the user propulsion performance. The summary information could also help clinicians justify wheelchair prescription by knowing whether a user propels more efficiently using a specific type of wheelchair over another for a period of time at his/her home and community and evaluate the effect of interventions such as a propulsion training program or a new wheelchair/seating component. The devices could also support research that investigates the relationship between upper extremity usage and upper extremity pain and injury in a more accurate manner, contributing to our understanding of the etiology and prevention of upper extremity pain and injury prevalent in this population.

One limitation of the study is that the testing protocol was highly structured and involved only straight courses. Also using the wheel rotation monitor to identify self-propulsion episodes may not be accurate in real-life settings. Our previous study showed that the monitors we used here were able to detect self-propulsion, external pushing, sedentary activities, and other activities with an accuracy of 90% using a laboratory-based protocol [37]. Future testing should consider real-life testing with a mixture of wheelchair propulsion and other activities of daily living in the home and community settings and combine the detection of wheelchair episodes with the estimation of propulsion parameters when assessing the overall estimation accuracies of temporal propulsion parameters.

5. Conclusion

Results in this study suggest that the use of triaxis accelerometers and a wheel rotation monitor could be a viable option to accurately monitor temporal parameters of wheelchair propulsion such as stroke number and push frequency especially when the accelerometer is worn on the upper arm. This study could result in a potential tool that can monitor the actual usage of upper limbs in terms of the repetitiveness and contribute to the preservation of upper limb functions among manual wheelchair users with SCI.

Conflict of Interests

The authors declare that they do not have any competing interests.

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Clinical Study

Investigation of Peak Pressure Index Parameters for People with Spinal Cord Injury Using Wheelchair Tilt-in-Space and Recline: Methodology and Preliminary Report

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The purpose of this study was to determine the effect of the sensel window's location and size when calculating the peak pressure index (PPI) of pressure mapping with varying degrees of wheelchair tilt-in-space (tilt) and recline in people with spinal cord injury (SCI). Thirteen power wheelchair users were recruited into this study. Six combinations of wheelchair tilt (15°, 25°, and 35°) and recline (10° and 30°) were used by the participants in random order. Displacements of peak pressure and center of pressure were extracted from the left side of the mapping system. Normalized PPI was computed for three sensel window dimensions (3 sensels × 3 sensels, 5 × 5, and 7 × 7). At least 3.33 cm of Euclidean displacement of peak pressures was observed in the tilt and recline. For every tilt angle, peak pressure displacement was not significantly different between 10° and 30° recline, while center of pressure displacement was significantly different ($P < .05$). For each recline angle, peak pressure displacement was not significantly different between pairs of 15°, 25°, and 35° tilt, while center of pressure displacement was significantly different between 15° versus 35° and 25° versus 35°. Our study showed that peak pressure displacement occurs in response to wheelchair tilt and recline, suggesting that the selected sensel window locations used to calculate PPI should be adjusted during changes in wheelchair configuration.

1. Introduction

Sitting-acquired pressure ulcers result from loading-induced soft tissue necrosis [1–3]. Pressure ulcers are both common (up to 85% lifetime incidence) and chronic (up to 70% recurrence) among people with spinal cord injury (SCI), largely due to the heightened pressure ulcer risk associated with diminished capacities to sense pain and to perform weight shifts [4, 5]. In the United States, treatment for pressure ulcers has been estimated to cost $1.2 billion annually, accounting for one-quarter of the total cost of SCI care [6].

Although pressure ulcer etiology is multifactorial, leading hypotheses are that tissue ischemia and tissue deformation are associated with the precursory tissue necrosis [1–3]. The former proposes the fact that mechanical loading prevents arterial vessels from resupplying tissues with oxygen and nutrients, leading to tissue ischemia and ultimately tissue necrosis. The latter proposes the fact that mechanical loading causes compressive and shearing deformation at the cellular level, leading to individual cell deaths and ultimately tissue necrosis. The apparent link between tissue necrosis and mechanical loading has prompted the development and evaluation of seating support surfaces in terms of optimizing seating interface pressure distributions [7–11]. Empirically, there is evidence linking increased mechanical loading with increased pressure ulcer incidence in elderly wheelchair

users [12]. Clinically, best-practice guidelines recommend routinely performing pressure-relieving maneuvers to further manage pressure ulcer risk [13–15].

To evaluate seating pressure distributions, interface pressure mapping (IPM) is commonly used to measure the normal forces at the seating interface. Brienza et al. [12] used IPM in a randomized clinical trial to explore the relationship between seating interface pressure and pressure ulcer incidence. Among 32 elderly wheelchair users, seating interface pressure was significantly higher among those who developed pressure ulcers during the trial compared to those who did not. Others have used IPM to assess tissue viability in response to postural interventions among people with SCI [16]. Various IPM metrics have been reported in the literature, including average pressure, peak pressure, peak pressure index, peak pressure gradient, peak pressure ratio, and dispersion index [2, 17, 18]. For any given IPM metric, there is always a tradeoff between reliability and repeatability versus descriptiveness. For example, while peak pressure (i.e., the peak sensel value) provides very precise information about ischial loading, it is relatively unreliable and unrepeatable; and while average pressure (i.e., the sensel average throughout the entire contact area) is relatively reliable and repeatable, it masks precise information about ischial loading [19]. In the literature, peak pressure index (PPI, the sensel average within a 10 cm^2 window of the peak pressure sensel) has been reported to strike a reasonable balance during cushion bench tests and tests of interface pressure with manual wheelchair seat angle changes [18].

The purpose of this study was to investigate the selection of two sensel window parameters of the PPI metric in response to wheelchair tilt-in-space (tilt) and recline. Recent studies have begun using PPI to assess the effectiveness of tilt and recline on relieving seating interface pressure [20–22]. Due to seating perturbations during dynamic seating conditions, special care may be needed when analyzing and interpreting seating interface pressures. Studies have reported 1.5 cm and 6 cm of sliding displacement during various dynamic seating tests, including tilt, recline, and forward flexion maneuvers [19, 23–26]. Depending on the analyzed region of interest, these interface displacements may significantly alter computed seating pressure metrics, such as PPI. In addition, secondary metrics, such as center of pressure, may be affected. Center of pressure refers to the coordinates obtained by summing the product of each sensel's pressure value with its grid coordinate and dividing the result by the total pressure sum; that is, $(\sum_{i=1}^{n} P_i x_i / \sum_{i=1}^{n} P_i, \sum_{i=1}^{n} P_i y_i / \sum_{i=1}^{n} P_i)$, where n was the number of sensels, P_i was the pressure of the ith sensel, and x_i and y_i were the x and y coordinates of the ith sensel [27]. Sonenblum and Sprigle [22] explored center of pressure displacement and found a significant displacement when tilting 15° beyond the upright posture. We investigated the changes in PPI and center of pressure in response to two calculation parameters: sensel window location and sensel window size. The sensel window refers to the group of sensels considered by the PPI computation. The sensel window dimension refers to the number of sensels along each side (e.g., 3 sensels × 3 sensels); the sensel window size refers to the 2-dimensional area occupied by the window (e.g., 10 cm^2); and the sensel window location refers to its placement within the pressure value grid (Figure 1). We explored the sensel window location via calculating the Euclidean displacements of peak pressure and center of pressure in response to tilt and recline, and we explored the sensel window size via comparing normalized PPI in response to different sensel window sizes.

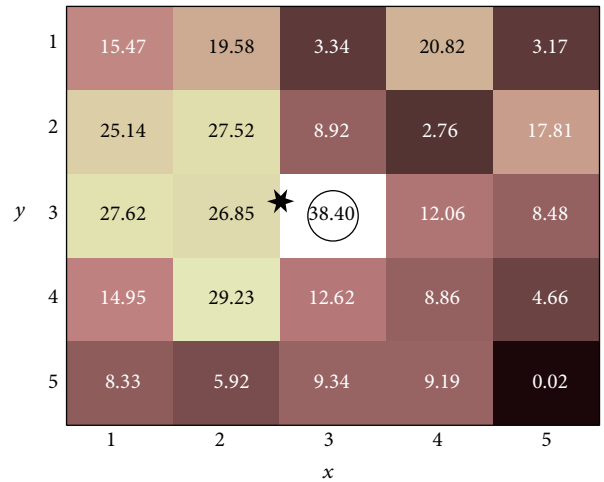

Figure 1: A sample 5 × 5 sensel window. The peak pressure refers to the maximal sensel value in the window. The peak pressure index refers to the average sensel value in the window. The center of pressure coordinate refers to the summation of the product of each sensel's pressure value with its grid coordinate, divided by the total pressure sum; that is, $(\sum_{i=1}^{n} P_i x_i / \sum_{i=1}^{n} P_i, \sum_{i=1}^{n} P_i y_i / \sum_{i=1}^{n} P_i)$, where n is the number of sensels, P_i is the pressure of the ith sensel, and x_i and y_i are the x and y coordinates of the ith sensel. For this sample, the peak pressure (circled) is 38.40 mm Hg, the peak pressure index is 14.44 mm Hg, and the center of pressure coordinate (starred) is at $x = 2.53$, $y = 2.80$.

2. Method

This study used an intervention and outcomes research design with repeated measures.

2.1. Participants. We enrolled 13 wheelchair users with SCI into the study. Participants were recruited via research flyers and referrals from a local rehabilitation hospital. Inclusion criteria included having traumatic SCI between the levels of C4 and T5, being at least 6 months after spinal injury, using a power wheelchair as the primary means of mobility and using a wheelchair seat width of 43 cm (17 in) to 53 cm (21 in). Exclusion criteria included having diseases that may affect cardiovascular function, diagnosed skeletal deformities (e.g., scoliosis, pelvic obliquity, and hip and knee contracture), or active pressure ulcers. All participants provided informed consent to this study, which was approved by the Institutional Review Board at the University of Oklahoma Health Sciences Center. The demographic data of the participants were as follows (values are mean ± SD): age 36.2 ± 10.0 years, body

weight 77.4 ± 17.9 kg, body mass index 24.6 ± 4.6 kg/m^2, and duration of injury 5.8 ± 5.9 years. The 13 wheelchair users included 4 women and 9 men: 3 African Americans, 1 Asian American, 1 multirace American, and 8 Caucasian Americans. Four participants had sensory complete injury (American Spinal Injury Association Impairment Scale (AIS) A), 2 participants had motor complete injury (AIS B), and 7 participants had incomplete injury (AIS C). All participants used a power wheelchair for mobility.

2.2. Apparatus. Seating interface pressures were recorded with an interface pressure mat (CONFORMat 5330, Tekscan, Boston, MA). The mat contains a 32 × 32 grid-based array of extremely thin, flexible tactile sensors. The sensor array can measure an area up to 47.1 cm × 47.1 cm with each sensel measuring approximately 1.47 cm × 1.47 cm. The mat system was calibrated before each subject's data collection, based on the manufacturer's guidelines.

A power wheelchair (C300 Corpus, Permobil, Lebanon, TN) with tilt and recline seat positioning functions was used in this study. The seat width was 48 cm (19 in). A standard high-density precontoured foam seat cushion (Corpus seating system, Permobil, Lebanon, TN) was used in this study. Tilt was defined as "a change of seat angle orientation in relation to the ground while maintaining the seat to back angle" [4, 5]. In this study, tilt referred only to the backward direction. Wheelchair recline was defined as "a change in seat to back angle while maintaining a constant seat angle with respect to the ground" [4, 5]. Configurations of tilt and recline are shown in Figure 2 and also described in our previous studies with the exception of recline measurements [14, 28, 29]. Previously, we reported recline measurements as the angle between the seat and the back support. In keeping with current clinical practice, we have modified our recline measurements to represent the sagittal angle of the back support from the vertical [30]. That is to say, our previously reported recline angles of 100° and 120° are now reported as 10° and 30°, respectively, although the actual recline configurations have not changed. Two digital angle gauges (WR300, Wixey, online-based company) were used to measure the tilt and recline angles.

2.3. Protocol. The wheelchair configuration protocol has been explained in our previous studies [14, 28, 29]. These studies were conducted in parallel to assess skin perfusion response to tilt and recline. The same subjects and protocols were used in these parallel studies. The tilt and recline protocol has been summarized in Table 1.

2.4. Procedure. Before the experiment, participants first provided informed consent. They were then asked to empty their bladders and to remain in the laboratory for 30 min to acclimatize to room temperature (23°C ±2°C). During the acclimatization period, the IPM mat was placed atop the standard precontoured seat cushion of the test power wheelchair. Upon completion of the acclimatization period, the participant was transferred to a mat table to affix a laser Doppler flowmetry sensor, which was used to measure skin perfusion for our parallel study regarding skin blood flow response to tilt and recline [14]. The participant was then transferred to the test power wheelchair, which contained an IPM mat at the interface between the cushion and buttocks. The participant was asked to place his or her hands in the lap and to sit as far back as possible while remaining comfortable. The foot support was adjusted to ensure that the femurs were parallel to the floor. The ischial tuberosities and coccygeal areas were palpated to ensure that they were positioned over the IPM mat. After a 6 min settling period to reduce the effects of creep [31], the IPM mat was calibrated for the given participant to minimize drift and hysteresis according to the manufacturer's procedure.

Table 1: Repeated measures before and after trial design.

Randomized configurations	Baseline condition (5 min)	Testing condition (5 min)	Washout condition (5 min)
1	0°T & 0°R	15°T & 10°R	35°T & 30°R
2	0°T & 0°R	25°T & 10°R	35°T & 30°R
3	0°T & 0°R	35°T & 10°R	35°T & 30°R
4	0°T & 0°R	15°T & 30°R	35°T & 30°R
5	0°T & 0°R	25°T & 30°R	35°T & 30°R
6	0°T & 0°R	35°T & 30°R	35°T & 30°R

R: recline; T: tilt-in-space.

Each experiment began with a washout configuration of 35° tilt and 30° recline. During the experiment, IPM samples were recorded at 10 Hz, and the range of acceptable angles was ±3° of the desired angle. To minimize operator effects, the same research assistant performed the tilt and recline adjustments for all experiments in this study. To minimize sequence effects, we used a balanced design with randomized testing protocols. To minimize carryover effects, we ended every testing condition with a washout configuration of 35° tilt and 30° recline, which also served to provide a recovery period of maximal pressure relief to the participant at least every 15 min, which satisfies the recommended pressure ulcer prevention guidelines [14]. Each participant spent roughly 100 min completing the entire protocol.

2.5. Data Analysis. A bicubic spline was first applied to the data prior to the metric calculations. Computations were performed using MATLAB 2012a (Mathworks, Natick, MA). For the displacement analysis, the independent variable was the seating orientation, which included six combinations of tilt (15°, 25°, and 35°) and recline (10° and 30°). The dependent variable was the displacement of either peak pressure or center of pressure among the various seating configurations. We performed two sets of comparisons. First, we grouped the recline angles to assess the effect of tilt on displacement. One-way analysis of variance (ANOVA) with Fisher's least significant difference (LSD) post hoc tests were used to compare pairwise displacement differentials between 15°, 25°, and 35° tilt for the two tested recline angles. Second, we grouped the tilt angles to assess the effect of recline on displacement. Paired samples t-tests were used to compare

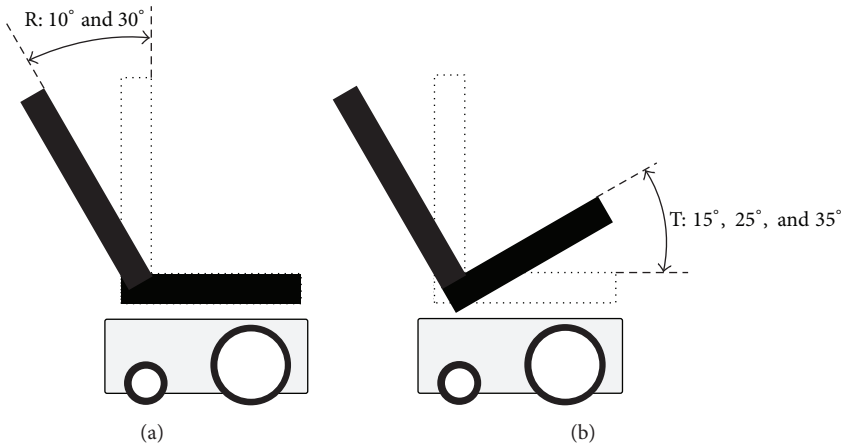

Figure 2: Configurations of wheelchair (a) recline and (b) tilt-in-space. R: recline; T: tilt-in-space.

the displacement differential between 10° and 30° recline for the three tilt angles. We considered using Bonferroni corrections; however, because of the high number of measures, low number of participants, and exploratory nature of the pilot study, we decided not to use these corrections due to the increased likelihood of committing a type II error, which was not acceptable for an exploratory study [32].

For the sensel window size analysis, the independent variable was the sensel window area, which included three configurations: 3 sensels × 3 sensels (19.45 cm^2), 5 × 5 (54.02 cm^2), and 7 × 7 (105.88 cm^2) (Figure 1). We chose these sensel window dimensions primarily for two reasons. First, a 2 × 2 sensel window would have fallen under the recommended 9-10 cm^2 area (i.e., estimated contact area of the ischial tuberosity) of the PPI metric [9, 18]. Second, the use of odd and square dimensions allowed us to consistently situate the peak sensel in the same location (i.e., the center) of the sensel window. The dependent variable was the peak pressure average within the sensel window. Seating pressures were averaged across each 5 min testing period. Values during each tilted and reclined period were normalized to their corresponding baseline values (Table 1). Because up to 30 sec were needed to complete the tilt and recline angle adjustments, seating pressures obtained during the first 30 sec of each 5 min tilted and reclined period were excluded from analysis. A one-way ANOVA with Fisher's LSD post hoc test was used to examine the efficacy of repeated measures between the 3 × 3, 5 × 5, and 7 × 7 sensel window dimensions. All statistical tests were performed using SPSS 22 (IBM, Somers, NY) at the significance level of .05.

3. Results

3.1. Displacement. Overall, peak pressure displacement between the upright and testing configurations ranged from 3.3 cm to 6.6 cm. Center of pressure displacement between the upright and testing configurations ranged from 0.6 cm to 1.7 cm.

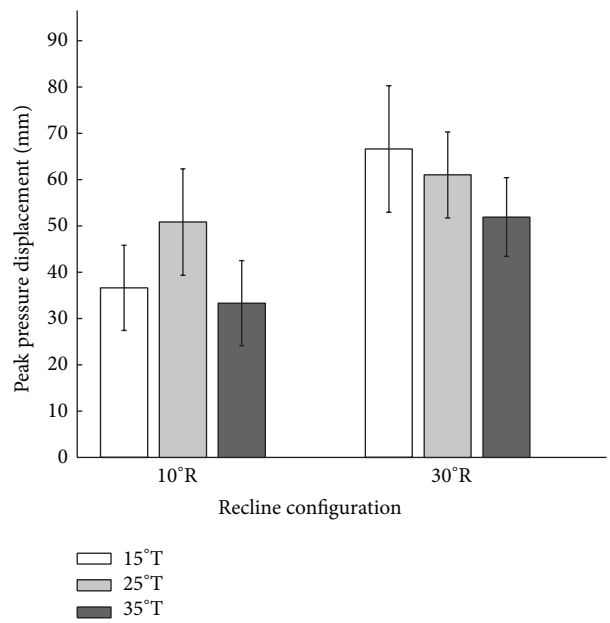

Figure 3: Pairwise comparisons of peak pressure displacement in the left side of the seat in response to six combinations of wheelchair tilt-in-space (15°, 25°, and 35°) and recline (10° and 30°). Data are shown as mean ± SE. R: recline; T: tilt-in-space.

3.1.1. Tilt Angle Effect. Pairwise comparisons of peak pressure displacement did not reveal significant differences between 15°, 25°, and 35° tilt for each of the two tested recline angles (Figure 3).

At 10° recline, pairwise comparisons of center of pressure displacement revealed significant differences between the following pairs of tilt angles: 15° versus 35° and 25° versus 35° ($P < .05$; see Figure 4). At 30° recline, pairwise comparisons of center of pressure displacement revealed significant differences between the same pairs of tilt angles: 15° versus 35° and 25° versus 35° ($P < .05$; Figure 4).

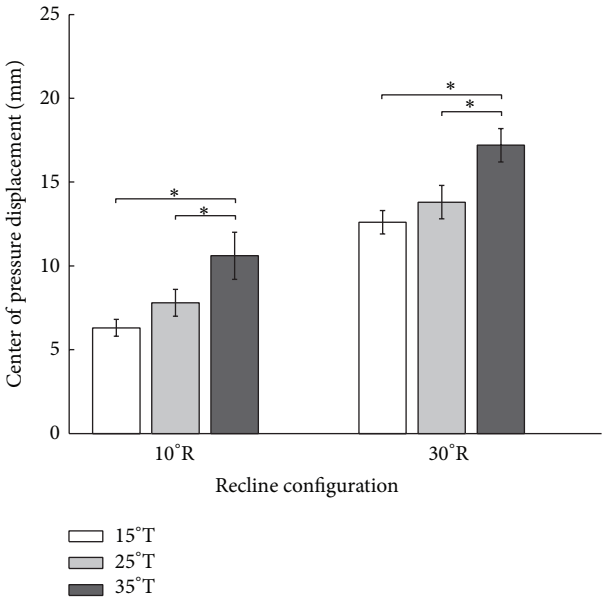

Figure 4: Pairwise comparisons of center of pressure displacement in the left side of the seat in response to six combinations of wheelchair tilt-in-space (15°, 25°, and 35°) and recline (10° and 30°). Data are shown as mean ± SE. R: recline; T: tilt-in-space; *a significant difference ($P < .05$).

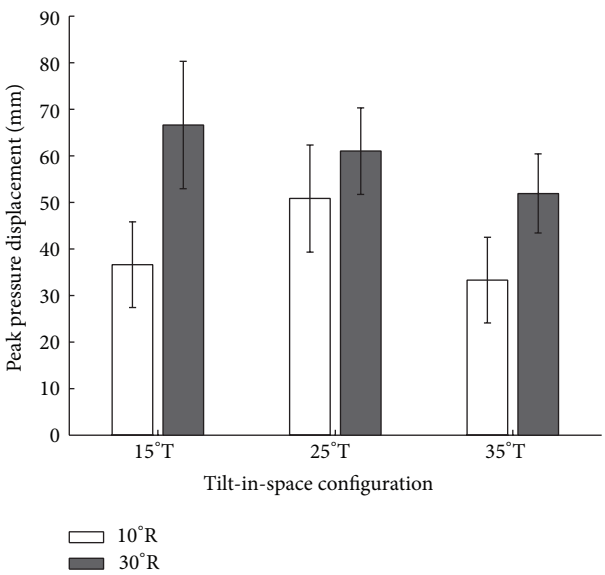

Figure 5: Comparisons of peak pressure displacement in the left side of the seat for three configurations of wheelchair tilt-in-space (15°, 25°, and 35°) in response to recline (10° and 30°). Data are shown as mean ± SE. R: recline; T: tilt-in-space.

3.1.2. Recline Angle Effect. Comparisons of peak pressure displacement did not reveal significant differences between 10° and 30° recline for each of the three tilt angles (Figure 5).

At 15° tilt, center of pressure displacement was significantly different between 10° and 30° recline ($P < .05$; see Figure 6). At 25° tilt, center of pressure displacement was significantly different between 10° and 30° recline ($P < .05$; see Figure 6). At 35° tilt, center of pressure displacement was significantly different between 10° and 30° recline ($P < .05$; Figure 6).

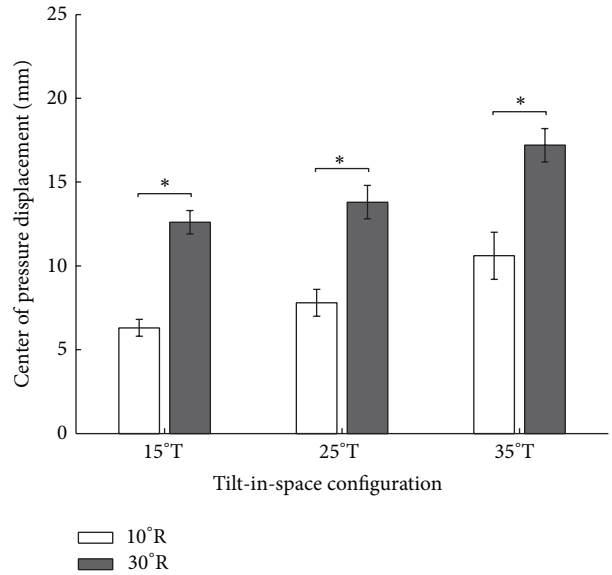

Figure 6: Comparisons of center of pressure displacement in the left side of the seat for three configurations of wheelchair tilt-in-space (15°, 25°, and 35°) in response to recline (10° and 30°). Data are shown as mean ± SE. R: recline; T: tilt-in-space; *a significant difference ($P < .05$).

3.2. Sensel Window Size. Comparisons of normalized PPI calculations among the three sensel window sizes for each wheelchair configuration revealed no significant differences. Figure 7 shows the normalized PPI in each wheelchair configuration for all three sensel window sizes. The statistical significance of normalized PPI versus the upright seating position was identical across all sensel window areas, except under one testing condition (25° tilt and 10° recline), in which the 5 × 5 and 7 × 7 sensel window sizes yielded significantly different PPI values while comparisons with the 3 × 3 window did not.

4. Discussion

Our study confirmed the occurrence of peak pressure and center of pressure displacement during combinations of tilt and recline. Although sliding displacement has been reported in the literature as a clinical challenge of wheelchair seating [33], few studies have investigated its effects on clinical and research applications. Cooper et al. [24] used motion capture cameras to measure sliding displacement during wheelchair sit-to-stand and recline operations. For hybrid and air cushions, they reported approximately 1.5 cm and 3.5 cm, respectively, of thigh displacement during recline. However, their testing did not include participants with SCI (rather, anthropometric test dummies) or standard foam cushions (rather, hybrid and air cushions). Hobson

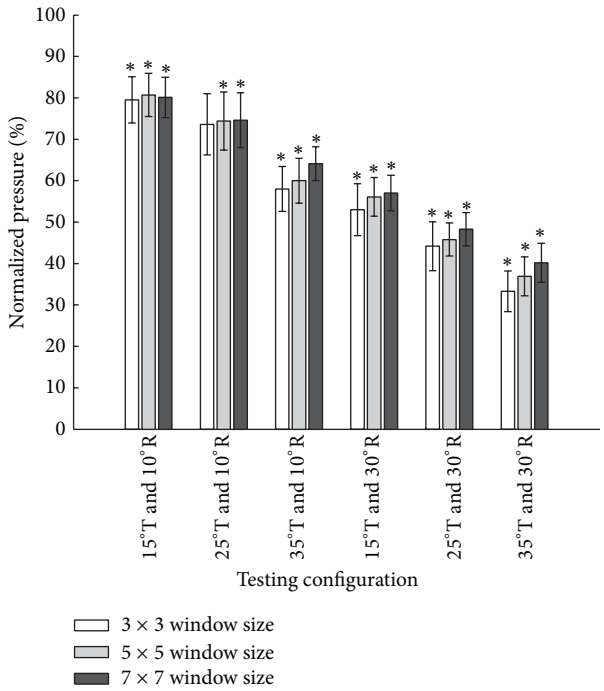

Figure 7: Comparisons of three sensel window sizes (3×3, 5×5, and 7 × 7) of normalized ischial pressure in response to combinations of wheelchair tilt-in-space (15°, 25°, and 35°) and recline (10° and 30°). Data are shown as mean ± SE. R: recline; T: tilt-in-space; *a significant difference compared to the upright sitting position ($P < .05$).

and Tooms [19, 23] used radiography to investigate pelvic movement across various body postures in people with SCI. When comparing a slight back support recline with 30° forward trunk flexion, they found the ischial tuberosity to be displaced by an average of 2.7 cm; and across all tested postures, they reported an average shift of approximately 4 cm. However, their experiment did not include IPM data to correlate with the radiographic data. Aissaoui et al. [25] quantified the sliding displacement of the buttock along the seat plane during repositioning and found approximately 6 cm of horizontal sliding at 30° tilt and 30° recline. However, their study only included participants without disability.

In our study, we recruited people with SCI and tested combinations of tilt and recline while seated on a precontoured foam cushion. Instead of relying on external measures of displacement, we examined the direct displacement of peak pressures at the buttock-cushion interface. Even at the lowest observed levels of sliding, we recorded peak pressure displacements of more than 3 cm. At the highest extreme, we observed nearly 7 cm of peak pressure displacement. These displacements indicate that when analyzing IPM data of people with SCI, the PPI sensel window should be moved in response to tilt and recline maneuvers. Otherwise, the peak pressures would frequently shift outside of the traditional PPI sensel window size of 10 cm^2.

Because pairwise comparisons of peak pressure displacement did not reveal significant differences, peak pressure displacement may be relatively insensitive to varying wheelchair seating configurations. That is to say, for a given tilt angle, changing the recline angle does not appear to produce significantly different displacements from one another and vice versa. However, this perhaps does not adequately represent the effect of wheelchair seating configuration on IPM measures. The effect of tilt and recline adjustments becomes more pronounced when we consider secondary metrics that have been computed from the raw IPM values, such as center of pressure. Among all three tilt angles, displacements of the computed centers of pressure were significantly different between the two tested recline angles. For each recline angle, center of pressure displacements was significantly different between two of the three tilt angle pairs. Thus, while raw peak pressure displacement was not significantly sensitive to varying wheelchair configurations, secondary metrics computed from the raw IPM values appear to be significantly affected.

Few studies have investigated center of pressure displacement in response to varying wheelchair configurations. One study used an instrumented simulator chair to mechanically actuate postural changes in 8 people without disability [34]. The study measured, among other metrics, seating interface pressure response to postural change and found center of pressure displacement to be a sensitive measure in response to wheelchair recline and pelvic rotation. Sonenblum and Sprigle [22] assessed 11 participants with SCI and found that the center of pressure was displaced significantly at 15° tilt compared to the upright position. Our study augments the previous reports of center of pressure displacement, providing center of pressure displacement in response to six wheelchair configurations among people with SCI.

In particular, the sensitivity of the center of pressure displacement metric is promising because it may be useful in the future to provide insight into currently unexplained physiological phenomena. Our previous work indicated that combinations of tilt and recline angles with lower than 25° tilt and 30° recline may not be sufficient to stimulate an effective skin perfusion response [14]. However, those findings are not universally representative of every person with SCI. We are optimistic that specific characteristics of center of pressure displacement may be helpful in understanding individual differences in skin perfusion response to tilt and recline. For example, two people may exhibit similar interface pressure measures at the ischial tuberosity, but significantly different perfusion responses. Increased center of pressure displacement may be an indicator for increased biomechanical changes in local soft tissues, stimulating higher microcirculation in those tissues. Further work is needed to explore whether lower-than-expected displacements in center of pressure displacement can be correlated with lower-than-expected skin perfusion measures in response to tilt and recline.

Sliding displacement characteristics may also be relevant to skin perfusion response in terms of the associated shear forces, which are known to contribute to internal tissue strain and blood vessel occlusion [35]. Hobson [19] found a 25% increase of shear forces in response to 30° recline and the elimination of shear forces in response to 20° tilt. Although shear forces are of significant interest to pressure

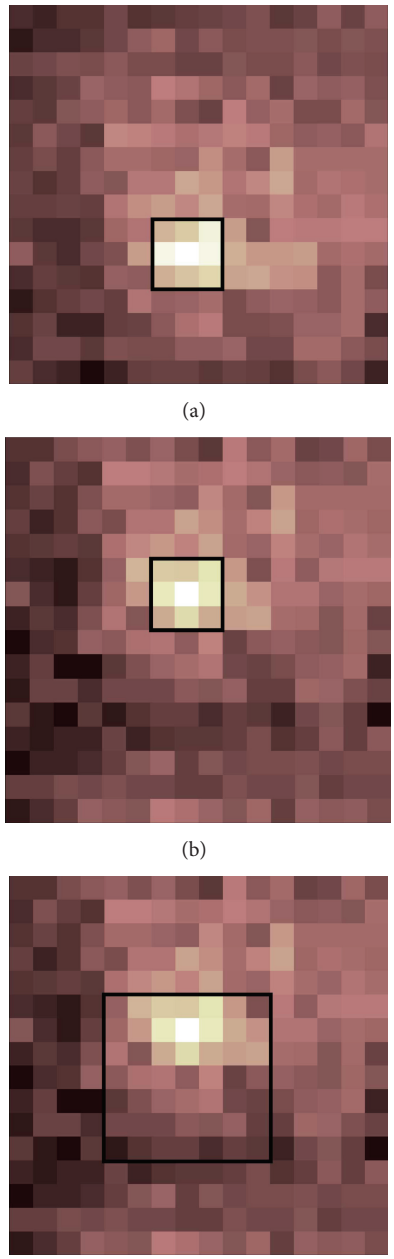

Figure 8: For IPM analysis, sliding displacement complicates the analysis. Two solutions are immediately apparent: move the sensel window location (a and b) or expand the sensel window area (a and c).

ulcer researchers, shear forces are difficult to measure without significantly altering the interface properties. While sliding displacement is not a direct measure of shear, it may still serve as a useful surrogate measure of shear forces at the seating interface. Because interface pressure displacement can be measured with IPM systems that are commercially available and do not significantly alter the seating interface, it may be an appropriate proxy measure for shear force response to tilt and recline.

For IPM analysis, sliding displacement complicates the analysis. Two solutions are immediately apparent: move the sensel window location (Figure 8(b)) or expand the sensel window area (Figure 8(c)). Although one study customized a sensor array (7 cm diameter) of interface pressure sensors and directly attached it to the buttock skin [22], a more practical solution for general clinical and research usage would be to adjust the sensel window parameters during IPM analysis. In our study, we adjusted the sensel window location for each tilt and recline maneuver. Sensel window dimensions were then increased to examine the effect of the sensel window area on the PPI metric. Preliminary findings indicate that if the sensel window location is adjusted appropriately during dynamic seating, the PPI metric does not appear to be sensitive to varying sensel window sizes. Despite observed peak pressure

displacements and significant center of pressure displacements, only one of the sensel window sizes within one testing condition differed statistically in normalized PPI. In all other conditions, the statistical significance of normalized PPI was not affected by sensel window size. Moreover, the one-way ANOVA did not reveal any significant pairwise differences in normalized PPI amongst the three sensel window sizes. These findings suggest that as long as sensel window locations are appropriately selected following tilt and recline changes, larger sensel window areas may not be necessary.

Our study was designed with clearly demarcated time periods for each wheelchair configuration. Thus, we afforded the benefit of distinct cutoff points to indicate when the sensel window should be moved. For more generalized settings in which the seating posture may be less structured and more dynamic, it may not be feasible to know when the sensel window location should be moved. In such cases, we propose two options. First, it may be preferable to perform a PPI time series analysis. That is, PPI would be computed at each time point, and the sensel window location would be dynamically selected for each frame. Further work is needed to determine the reliability and repeatability of this method. Second, the sensel window size could be enlarged. That is, the sensel window location would be maintained statically throughout the entire data set, but the increased sensel window size would be able to accommodate unpredictable displacements. Further work is needed to determine an appropriate sensel window size and whether the noncentered peak concentrations would significantly affect PPI calculations.

There are limitations of this study. First, a laser Doppler flowmetry sensor was attached at the site of the ischial tuberosity as part of our parallel study regarding skin perfusion response to tilt and recline [14]. However, the probe was small and thin, and we analyzed normalized pressure measures, instead of pure magnitudes. While there may be some residual effects from the attached probe, we felt it was a justifiable limitation to facilitate our long-term goal of integrating skin perfusion measures with interface pressure measures to develop a comprehensive assessment of wheelchair configurations. Second, our inclusion criteria specified people who used a seat width between 43 cm and 53 cm. In future work, multiple seat widths will be tested to accommodate a broader demographic of participants. Third, we only recruited 13 wheelchair users with SCI. This was initially a feasibility study to examine whether our protocols could be used to assess the efficacy of tilt and recline. In future work, a larger sample size using this protocol should be conducted to verify our findings.

5. Conclusions

Our study demonstrated the existence of peak pressure displacement and center of pressure displacement during tilt and recline maneuvers. Thus, the PPI sensel window location should be adjusted during position changes. Preliminary evidence also indicates that if the sensel window location is adjusted appropriately, increasing the sensel window dimensions from 3×3 up to 7×7 does not have a significant effect on the PPI metric. In future work, we hope to establish peak pressure displacement as a surrogate measure for shear force and center of pressure displacement as an indicator for individual differences in skin perfusion response to tilt and recline.

Abbreviations

ANOVA: Analysis of variance
IPM: Interface pressure mapping
PPI: Peak pressure index
SCI: Spinal cord injury.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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Research Article

Measurement of Hand/Handrim Grip Forces in Two Different One Arm Drive Wheelchairs

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Purpose. The aim of this study was to explore the total and regional grip forces in the hand when propelling two different manual one arm drive wheelchairs: the Neater Uni-wheelchair (NUW) and a foot steered Action3 wheelchair. *Methods*. 17 nondisabled users were randomly assigned to each wheelchair to drive around an indoor obstacle course. The *Grip*, a multiple sensor system taking continuous measurement of handgrip force, was attached to the propelling hand. Total grip force in each region of the hand and total grip force across the whole hand were calculated per user per wheelchair. *Results*. The Action3 with foot steering only generated significantly greater total grip force in straight running compared to the NUW and also in the fingers and thumb in straight running. *Conclusions*. The results suggest that the Action3 with foot steering generated greater grip forces which may infer a greater potential for repetitive strain injury in the upper limb. Further work is required to explore whether the difference in grip force is of clinical significance in a disabled population.

1. Introduction

Manual wheelchair propulsion is known to be an inefficient means of ambulation which has been associated with a high prevalence of upper limb injuries [1, 2]. Such injuries are thought to occur from a combination of repetitive movements, upper limb weakness, and inefficient propulsive technique [3, 4]. Hemiplegic users are particularly vulnerable to upper limb injury and pain [5, 6] because of being reliant on only one arm for propulsion. Moreover, the action of manual propulsion necessitates that the hand exerts repetitive forces to the handrim [6, 7] in order that the hand/handrim coupling be stable to accommodate the transfer of the forces from the shoulder and arm muscles onto the handrim [7]. This repetitive grip action may contribute to the development of upper limb repetitive strain injury [6, 8]. In a standard wheelchair, the handrim design has also been suggested to be one of the factors responsible for the low mechanical efficiency in manual propulsion [9].

There is currently very little choice of one arm drive wheelchairs. The most commonly prescribed include the one arm ratchet arm or lever-drive mechanism and the dual handrim mechanism. There are deficiencies associated with both of these designs, particularly with respect to the user interface. For a large number of users, the overall ergonomics of operation is not efficient and recent research has suggested that neither may be suitable for hemiplegic users [10]. A recent alternative to these has been the development of the Neater Uni-wheelchair (NUW) which has been designed specifically for hemiplegic users in response to the identified problems associated with the other two market alternatives. The NUW is an Action3 wheelchair to which a rear wheel differential and front wheel steering kit are attached. These features have been described in detail in a range of studies comparing it to both the dual handrim and lever-drive alternatives [11–15]. The differential enables a single pushrim to propel both rear wheels equally resulting in the wheelchair moving in a straight line. The steering mechanism is attached to one foot plate and operates independently of the drive mechanism. Steering is intuitive: rotating the foot to the right turns the wheelchair to the right; rotate the foot to the left and chair turns left. In addition the kits can be attached to

either side for use by either right- or left-handed users (insert Figures 1 and 2) and can include both components or only the steering mechanism. The body of work to date, comparing the NUW to existing provision, suggests that the NUW is ergonomically more efficient to drive and preferred by users in a laboratory setting [11, 12], in their own homes [12, 13], and in simulated activities of daily living setting [14]. These studies suggested that NUW could meet the unmet needs of the hemiplegic user group and provide them with additional choice in their wheelchair provision. However, there is no research exploring grip action in different one arm drive wheelchairs.

Wheelchair propulsion necessitates the repetitive use of the upper limb joints and muscles which have been linked to repetitive injuries in manual wheelchair users [7]. A recent study, measuring shoulder muscle activity using EMG in one arm drive wheelchairs, has shown an increase in activity levels when propelling the Action3 wheelchair when compared to the NUW [16]. These changes in muscle activity may be linked to the incidence of repetitive injuries in wheelchair user. Repetitive injuries have also been associated with grip action during manual wheelchair propulsion. Early research surrounding hand function in wheelchair propulsion has focused on distinguishing between power grip and precision grip [17]. Whilst these positions have been explored extensively in terms of biomechanic, ergonomic, and functional differences, the findings are equivocal. Chao et al. [18] and Cooney and Chao [19] suggested that a strong pincer grip overloads forearm tendons more than a power grip. Conversely Edgren et al. [20] suggest a correlation between the tension of the long finger tendons and handgrip diameter suggesting that the optimization of the handrim diameter is important in reducing effort. This was further endorsed by Peebles and Norris [21] who suggested that grip force in all age groups can be affected by the handle diameter, position, and movement direction. The measurement of local force in the hand is an important aspect to be investigated in any hand tool evaluation process. Instrumented gloves with force sensor resistors (FSR) have been used to measure mechanical loads on the hand's surface during a prehensile activity and have been used for biomechanical evaluation studies [22, 23]. More recent work by Mastalerz et al. [24], using the *Grip* Force Measurement System, explored the maximal strength capabilities of the hand, regarding the grip handle diameter and position, among young men and women but failed to identify an optimal handle diameter generating the maximal handgrip force. Whilst these studies have provided valuable information on kinetics of grip function, the understanding of the interaction between hand surface and handrim of one arm drive wheelchairs remains unclear. Medola et al. [25] measured pressure in different components of the hand, including the thumb, palm, and finger tips, during manual wheelchair propulsion using a similar instrumented system to the *Grip*. The results showed that the hands hold the rim with the distal phalanges of the fingers, concentrating higher pressures on these areas. This could indicate increased loading in the flexor tendons of the forearm which may contribute to the development of overuse injury [19]. The research to date has only explored handrim grip force in standard wheelchairs and not one arm drive wheelchairs. The aim of this study was to explore the total and regional grip forces within the hand exerted at the hand/handrim interface in two different manual one arm drive wheelchairs.

Figure 1: The Neater Uni-wheelchair.

The research hypothesis was as follows. There will be differences in total and regional grip forces at the hand/handrim interface when propelling different manual one arm drive wheelchairs.

2. Methods

Ethical approval was sought and obtained from the University of Brighton Research Ethics Committee for the study.

Participants were recruited from the University of Brighton Campus using posters. The inclusion criteria were willingness to participate, no cardiac or respiratory disorder, no functional impairment, hand size to fit medium size vinyl glove, right-hand being dominant and being within the height and weight restrictions of 163–185 cm height and 54–90 kg weight. Exclusion criteria was inability to learn how to propel the wheelchair safely. Participants were provided with an information sheet prior to being recruited into the study to enable them to make an informed decision concerning their involvement. All participants who wished to take part completed a health declaration sheet and informed consent sheet.

The study was designed as a controlled, same subject study that measured grip force generated by each user during propulsion in two different manual one arm drive wheelchairs. The two wheelchairs, as described in the introduction, were

(1) an Action3 wheelchair with Neater Uni-wheelchair kits applied which included the differential in the rear wheel and the foot steering mechanism (Figure 1);

(2) an Action3 wheelchair with only the Neater Uni-footplate steering and a standard handrim for propulsion (Figure 2).

The drive wheels were the same size and diameter in both wheelchairs. There was no difference in weight between the wheelchairs.

Figure 2: The foot steered standard Action3 wheelchair.

3. Materials

3.1. The Grip Force Measurement System. The data being measured were grip force at the hand/handrim interface using the Tekscan *Grip* Force Measurement System. This is a portable interface force mapping system which records grip force distribution under the contact area. Tekscan force and pressure measurement systems have been demonstrated to generate reliable and valid data [26].

The *Grip* system includes a series of force transducers mounted on a flexible force plate which can mould to the shape of the hand. Each force transducer (sensel) consists of a unique piezoresistive material sandwiched between two pieces of flexible polyester, with printed silver conductors on each half [27]. Software is provided which can generate force data or force mapping of the contact area. The *Grip* has 18 sensing regions that are positioned over the fingers and palm (Figure 3). Gaps between the sensing areas allow the joints to move freely and not interfere with grip measurement. Each sensing region has multiple sensels for localized identification of contact points on the hand. There are a total of 349 sensels in the array and the sampling rate for the system is 30 Hz.

The study was conducted at an indoor circuit at the University of Brighton (Figure 5). All participants were given familiarisation training in the use of both the wheelchairs until they felt competent to undertake the trial. The steering for both wheelchairs involved the Neater Uni-wheelchair steering mechanism. Maneuvering the Neater Uni-wheelchair involved the use of the single rim which was attached to the rear wheel differential for propulsion. Maneuvering the foot steered Action3 involved propulsion only using the single handrim on one side of the wheelchair.

The total length of the driving course in the indoor circuit was 150 m. The distance was standardised for all participants and the time taken to complete each activity within the course was recorded. Participants were initially asked to drive across the gymnasium floor for 30 m, complete a 90° left turn, and continue for 10 m. A further 90° left hand turn took the user onto carpet and brush matting. The carpet and matting were 30 m long. At the end of the carpet the user made a 135° left hand turn into a series of corners consisting of three closely placed bollard markers which required tight 45° right- and left-hand turns. At the end of the corners, the user completed a 135° right-hand turn for a further 30 m of straight driving to take the user back to the start/finish line.

3.2. Procedure. Demographic data including age and gender were recorded for all subjects. The users hand was measured for size using a preinstrumented vinyl glove to which the Grip sensors had been attached (Figure 6) according to the manufacturer instructions [27].

The user's right hand was dusted with talcum powder prior to putting on a first tight fitting noninstrumented vinyl glove. This vinyl glove was also dusted with talcum powder and then inserted into the preinstrumented vinyl glove (Figure 6). The procedure of double gloving was for hygiene reasons. The use of talcum powder facilitated easy removal of the instrumented glove without damage to the sensors.

The preinstrumented glove was attached to the VersaTek cuffs (Figure 7) which processed the data and relayed it to the computer via USB connection.

The system was calibrated for each subject prior to data collection as recommended by the manufacturer [27]. Subjects were randomly allocated the wheelchairs using random numbers.

The participants were asked to drive each wheelchair round the course (Figure 5). Data was captured continuously throughout each circuit. Time taken to complete each activity within the circuit was recorded. The activities were defined as straight running, mats (simulating resistance), and corners.

The key time points were

(1) A-B: start and straight running to first bend;

(2) C-D: beginning of mats to end of mats and third bend;

(3) D-E: beginning of the corner to final bend.

The course was repeated once per wheelchair with a 30 minute gap, or however much time was necessary, for the users to feel recovered.

3.3. Data Processing. The force time data for each region was exported into Excel. A linear trapezoidal integration of force was performed for each region. The total grip force generated at the hand/handrim interface for each activity (straight running, corners, and mats) was recorded.

The raw grip force data was also manipulated using the *Grip* software to generate three regions of the hand consisting of thumb, digit, and palm fields. Figure 4 shows the position of the sensors. The thumb field consisted of two sensors, the palm field 4 sensors, and the finger field 3 sensors/finger, totaling 12 sensors.

Figure 8 shows an example of the grip forces generated in the finger, thumb, and palm field during straight running propulsion in the NUW.

For each field the data from all sensors were totaled. Totals were calculated for each of the three fields for each activity (straight running, mats, and corners) in each wheelchair.

3.4. Statistical Analysis. The data were investigated to explore the differences in forces in the different regions of the hand,

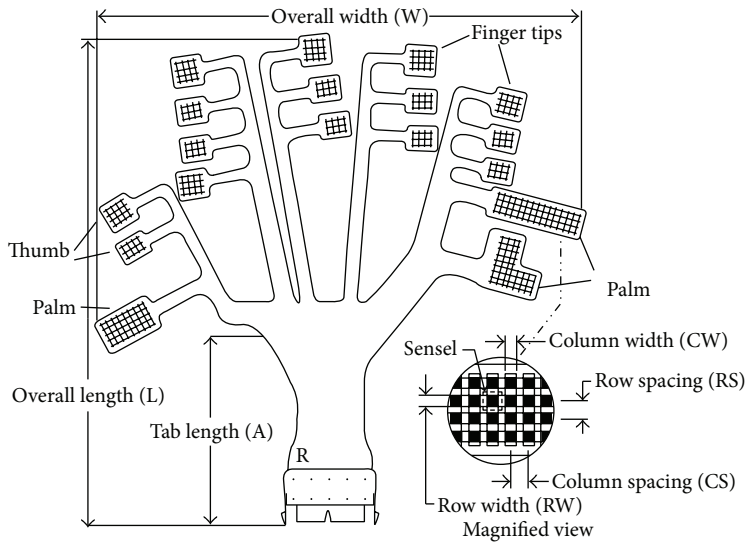

Figure 3: The *Grip* sensors prior to attachment to a vinyl glove [27].

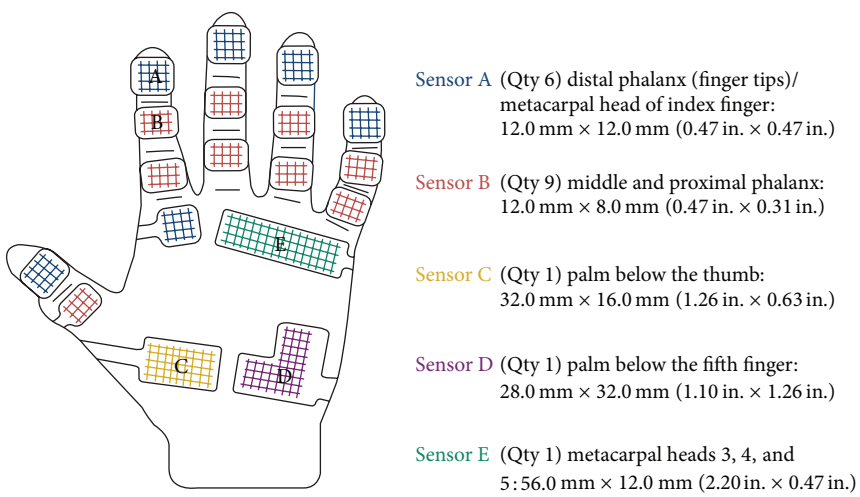

Figure 4: Location of the sensors after attachment to a glove [27].

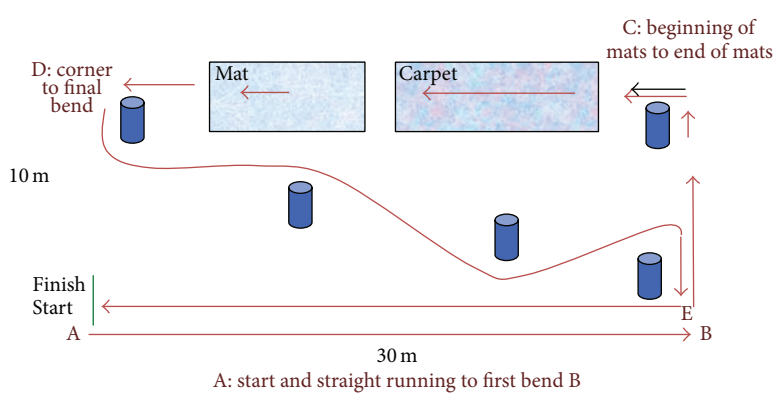

Figure 5: The driving course.

Figure 6: The preinstrumented glove showing the Grip attached to the vinyl glove [27]. Photographs courtesy of SATRA Technology Centre.

Figure 7: The VersaTek cuffs [27]. Photographs courtesy of SATRA Technology Centre.

over different surfaces between the two different wheelchairs across the whole sample. The data was not normally distributed and was logarithmically transformed using a log 10 manipulation to enable a three-way ANOVA with Tukey's post hoc test ($\alpha < 0.05$) to be undertaken to explore the forces across the regions of the hand. The three independent variables were type of wheelchair, activity, and region of the hand. The total forces were analysed using a two-way ANOVA to explore the differences between the wheelchairs.

The two independent variables were wheelchair and activity.

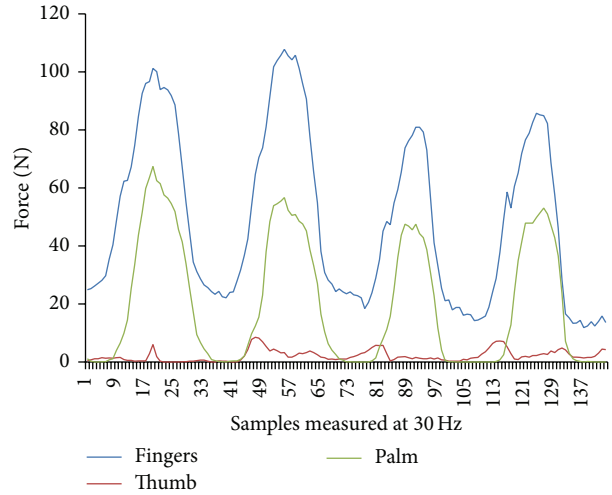

Figure 8: An example of raw data from the finger, thumb, and palm fields during straight running propulsion in the NUW.

4. Results

The gender distribution was 10 women and 7 men (Table 1).

There was no difference between the time taken to complete each activity within the circuit in the different wheelchairs ($P = 0.245$). The mean (SD) time taken to complete the straight running section of the course was 13.54 (3.86) secs in the Neater Uni-wheelchair and 12.94 (3.56) secs in the Action3 wheelchair. This would indicate that the velocity in straight running would be 0.71 (0.19) m/sec in the Neater Uni-wheelchair and 0.74 (0.18) m/sec in the Action3 wheelchair. This was also shown not to be significant ($P = 0.548$). The time taken to complete the corners section of the course was 11.76 (3.03) secs in the Neater Uni-wheelchair and 11.76 (3.93) secs in the Action3 wheelchair. For the mats section of the course the times were 19.12 (4.04) secs for the Neater Uni-wheelchair and 19.41 (4.64) secs for the Action3 wheelchair. These results were also not significant.

Total grip force and hand component grip forces are described in Table 2.

4.1. Total Handgrip Forces. The results of the 3-way ANOVA showed that there was a significant difference between the forces generated in propelling the different wheelchairs ($F = 5.016$, $df = 1$, and $P = 0.027$).

Post hoc t-tests indicated that there were significant differences between the wheelchairs in straight running ($t = -2.729$, $df = 16$, and $P = 0.015$). Greater total force was exerted when propelling the foot steered Action3 in straight running. There were no significant differences in force generation between the wheelchairs around corners ($t = -0.719$, $df = 16$, and $P = 0.483$) or over mats ($t = -1.463$, $df = 16$, and $P = 0.163$).

4.2. Grip Forces across the Different Regions of the Hand. The results of the 3-way ANOVA showed that there were significant differences in the force exerted between the

Table 1: To show demographic variables of the participants.

Male	Mean	SD	Female	Mean	SD
Age (yrs)	25.86	11.05	Age (yrs)	30.3	11.34
Height (cm)	183	9.70	Height (cm)	166.9	6.54
Weight (kg)	77.29	19.03	Weight (kg)	62.1	7.43

Table 2: Descriptive log transformed data to show mean and standard deviation of force measurement (N) per region per wheelchair for each activity.

Region	Wheelchair	Straight running mean (SD)	Corners mean (SD)	Mats mean (SD)
Fingers	Action3	2.81* (0.18)	2.90 (0.35)	2.76 (0.24)
	Neater	2.71 (0.17)	2.85 (0.17)	2.67 (0.23)
Thumb	Action3	2.22* (0.18)	2.26 (0.35)	2.09 (0.28)
	Neater	2.10 (0.19)	2.17 (0.18)	1.97 (0.23)
Palm	Action3	2.26 (0.31)	2.30 (0.45)	2.28 (0.31)
	Neater	2.30 (0.27)	2.33 (0.25)	2.21 (0.25)
Total	Action3	7.29* (0.56)	7.47 (1.09)	7.13 (0.78)
	Neater	7.12 (0.52)	7.34 (0.54)	6.85 (0.68)

Key
* signifies significant difference between the two wheelchairs.

regions ($F = 168.428$, $df = 2$, and $P < 0.001$) and the wheelchairs ($F = 4.782$, $df = 1$, and $P < 0.03$).

Post hoc t-test suggested that there were significant differences between the wheelchairs in force exerted through the fingers ($t = -2.634$, $P = 0.018$) and thumbs ($t = -3.301$, $P = 0.005$) in straight running. Greater forces were exerted in propelling the foot steered Action3 wheelchair.

There were no significant differences between the wheelchairs in any of the other activities.

Post hoc Tukey's test to explore differences between regions of the hand demonstrated differences between all regions with the fingers generating a greater force than the palm ($M = 0.5029$, 95% CI [0.4156, 0.5902], and $P < 0.001$) and thumb ($M = 0.648$, 95% CI [0.5607, 0.7353], and $P < 0.001$). The palm generated more force than the thumb ($M = 0.1452$, 95% CI [0.0579, 0.2325], and $P < 0.001$).

5. Discussion

The aim of this study was to measure and compare the total and regional handgrip forces measured at the hand/handrim interface during propulsion, in a sample of nondisabled participants using right-sided one arm drive wheelchairs. The objective of the study was to explore the grip forces generated when maneuvering different one arm drive wheelchairs in a controlled environment and around obstacles. The work is novel because grip forces have not been measured in any one arm drive wheelchairs.

The results indicated that users applied higher total grip forces and finger and thumb forces at the hand/handrim interface in straight running when using the foot steered Action3 wheelchair when compared to the Neater Uni-wheelchair. This difference may be explained through the action of the differential in the Neater Uni-wheelchair. The differential ensures that the force applied to the handrim delivers equal torque to both drive wheels of the wheelchair, thus ensuring that the wheelchair will be propelled more easily in a straight line. In the foot steered Action3 wheelchair, the force is applied to only one rear wheel. This results in a turning moment with the wheelchair tending to turn away from a straight path. To compensate for this, the front steering wheel would need to be maintained to apply an equal and opposite turning moment to keep the wheelchair path straight. Depending upon the nature of the surface, the propelling force acting on the turned steering front wheel could result in an increase in forward rolling resistance. This would be sufficient to explain the differences found in total force.

In light of the results from straight running, it was speculated that the results for the force during propulsion over the mats would also be lower in the Neater Uni-wheelchair. However, this was found to not be the case, although it was observed that during the data collection all the users struggled to manoeuvre each of the wheelchairs over the mats. This may suggest that if the differences in hand/handrim grip force were due to increased rolling resistance resulting from the compensatory turned position of the front steered wheel, this effect may have been masked by the greater effort required to propel the wheelchairs over the resistive surfaces. Further work could explore this in greater detail.

Early studies have investigated the influence of handle diameter on wheelchair propulsion rather than hand/handrim force [28]. Recent studies [24, 25] used the *Grip* measure tool to measure hand interface force in a variety of handle diameters. Their findings concur with earlier work [28] that a larger diameter handrim requires greater force for propulsion, reporting that the optimum diameter to develop the maximal handgrip force is between 20 and 30 mm. In this study the handrims in both the foot steered Action3 and the Neater Uni-wheelchair were the standard manufacturer handrims (20 mm) and within the optimum range.

The results from the two wheelchairs could also have been influenced by differences in the weights of the chairs, their rolling resistances, and the velocity with which they were

propelled. In this study the two wheelchairs were both of the same model and both had the same foot steering device attached. The only difference between the wheelchairs was the addition of the differential to the Neater Uni-wheelchair. The difference in weight between the wheelchairs was negligible (0.1 kg) which theoretically would suggest that there would be no difference in rolling resistance. It is, however, acknowledged that rolling resistance was not measured in the study. Moreover, the velocity of wheelchair propulsion was also not controlled in this study. However, as the time taken to complete the circuits did not vary significantly, it could be suggested that the velocity would also not have varied. Greater control of the velocity using a treadmill would eliminate this variable from the study; however this in turn would have precluded simulation of functional use of the wheelchair.

The results from the analysis of the regions of the hand also provided some interesting findings which supported those of the total grip force analysis. The results suggested that forces generated in the fingers and thumbs were greatest in the foot steered Action3 wheelchair in straight running. These findings concur with the results reported by Medola et al. [25] who showed that, when propelling a wheelchair, the hands hold the handrim with the distal phalanges of the fingers, concentrating higher force on these areas. The increase in grip forces generated by the fingers would indicate that a stronger grip was being applied when propelling the foot steered Action3 wheelchair. There was no significant difference in hand/handrim forces when propelling the wheelchairs over the mats or around corners.

The evidence suggests that, while it is possible to modify a standard Action3 wheelchair for use by people with use of one arm by attaching an independent steering mechanism, the grip force exerted at the hand/handrim interface will be significantly greater in straight running than the wheelchair with both components of the Neater Uni-wheelchair kit. Whilst this may be statistically significant, further work is required to establish if this has any clinical significance within a user population.

6. Conclusion

The measurements of grip forces have only previously been reported in standard manual wheelchairs and not in manual one arm drive wheelchairs. The findings from this novel study, whilst concurring with findings from the standard wheelchair literature, add to our understanding of handgrip forces exerted in one arm drive wheelchairs. The results suggest that the Action3 with foot steering generated greater grip forces which may infer a greater potential for repetitive strain injury in the upper limb. Modifying an Action3 wheelchair with the addition of a steering mechanism may be a potentially more economical alternative to the Neater Uni-wheelchair, but one which may not be so ergonomically efficient. Further work is required to explore whether the difference in grip force is of clinical significance in a disabled population.

Conflict of Interests

There is no conflict of interests between the participating parties. All contributors have reviewed and agreed the content of the paper.

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Research Article

Comparing the Activity Profiles of Wheelchair Rugby Using a Miniaturised Data Logger and Radio-Frequency Tracking System

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The current study assessed the validity and reliability of a miniaturised data logger (MDL) against a radio-frequency-based indoor tracking system (ITS) for quantifying key aspects of mobility performance during wheelchair rugby. Eleven international wheelchair rugby players were monitored by both devices during four wheelchair rugby matches. MDL data were averaged over both 1-second (MDL-1) and 5-second (MDL-5) intervals to calculate distance, mean, and peak speeds. The results revealed no significant differences between devices for the distance covered or mean speeds, although random errors of 10% and 12%, respectively, were identified in relation to the mean values. No significant differences in peak speed were revealed between ITS ($3.91 \pm 0.32\ \mathrm{m{\cdot}s^{-1}}$) and MDL-1 ($3.85 \pm 0.45\ \mathrm{m{\cdot}s^{-1}}$). Whereas peak speeds in MDL-5 ($2.75 \pm 0.29\ \mathrm{m{\cdot}s^{-1}}$) were significantly lower than ITS. Errors in peak speed led to large random errors in time and distance spent in speed zones relative to peak speed, especially in MDL-5. The current study revealed that MDL provide a reasonable representation of the distance and mean speed reported during wheelchair rugby. However, inaccuracy in the detection of peak speeds limits its use for monitoring performance and prescribing wheelchair rugby training programmes.

1. Introduction

The use of innovative assistive technology within the wheelchair court sports (basketball, rugby, and tennis) has increased dramatically over recent years and has been associated with improvements in athletic performance [1–3]. However, one ongoing challenge faced by researchers and practitioners is the ability to quantify the demands of these sports both accurately and practically in order to optimise how training is prescribed whilst minimising injury risk to athletes.

Miniaturised data loggers (MDL) were originally developed to determine the activity profiles of daily life wheelchair users [4] and have recently been implemented into sporting wheelchair environments [5–7]. The MDL are small, lightweight devices which attach near the axle of the main wheels, powered by long life batteries, enabling data to be collected and stored over periods of approximately 3 months [4]. This has practical implications that may benefit sport science practitioners, as devices could be attached to athletes' sports wheelchair to record the distances covered and speeds reached throughout training sessions over an extended period of time. Practitioners could then review each athlete's performance and modify training programmes accordingly over future training cycles, with minimal input required at individual training sessions.

Sindall et al. [8] recently validated the MDL for use on a court sports wheelchair against a motor-driven treadmill and a 1 Hz global positioning system (GPS). It was revealed that although the MDL provided an accurate and reliable representation of distance at low speeds ($<2.5\ \mathrm{m{\cdot}s^{-1}}$), coefficients of variations (% CV) as high as 19.9% CV were

reported at speeds in excess of 2.5 m·s^{-1} during controlled laboratory and field based trials [8]. However, when assessing the validity and reliability of a device, it is essential that the device is assessed during movements and intensities specific to the intended activity [9]. In the case of the wheelchair court sports, movements are multidirectional, intermittent activities, which include accelerating, turning, braking, and sprinting [10]. Sindall et al. [8] also attempted to validate the MDL during wheelchair tennis match-play, where significantly lower peak speeds, yet greater total distances and mean speeds, were reported by the MDL in relation to GPS. Unfortunately, the 1 Hz GPS used was unlikely to be sufficient for the assessment of these variables during competition, particularly when high speeds are being reached, questioning its suitability as a valid criterion measure [11–14].

Radio-frequency tracking systems have become increasingly popular for tracking athlete's movements in indoor sports, since they operate in a similar fashion to GPS with the main exception being the use of fixed base stations as sensors instead of satellites. Much like GPS, athletes are required to wear a small lightweight tag, making radio-frequency systems an unobtrusive and practical solution. Previous studies have also demonstrated that these systems can be highly precise and reliable in their assessment of location, distances, and speeds [15–18]. However, unpublished data from our laboratory has validated a radio-frequency-based indoor tracking system (ITS) specifically for use within the wheelchair court sports. It has suggested that at 8 Hz and 16 Hz sampling frequencies, the ITS provides a valid and reliable assessment of distance and mean speed with relative errors $\leq$ 0.7% observed during a range of tasks specific to the wheelchair court sports. Importantly, even at maximal speeds (>4 m·s^{-1}), random errors were <0.10 m·s^{-1}, with <2% CV using the ITS, further demonstrating its suitability for use with the wheelchair court sports. Unfortunately, from a practical perspective, the ITS requires greater set-up and calibration time and more practitioner input and are not as portable as MDL, making it slightly less attractive as a monitoring tool in an elite, applied environment.

An accurate quantification of peak speeds is highly desirable given that relative speed zones determined from peak speeds are emerging as a method for monitoring performance and optimising training strategies specific to each individual in team sports [19, 20]. A limitation associated with the MDL is that reed switches are currently positioned at 120° intervals and speed data is calculated as a function of the time and distance between successive reed switch activations, with the latter being determined by the athlete's wheel dimensions [4]. As a consequence, the MDL is not capable of reporting instantaneous speed. Subsequently previous research has averaged speed data over 5-second intervals, to correspond with heart rate data, in order to obtain a "mean" peak speed [6–8]. However, this analysis approach is likely to be too long to establish "true" peak speeds and alternatively averaging MDL data over shorter 1-second intervals may provide a more valid representation of peak speeds.

The aim of the current study was to determine the validity and reliability of the MDL, averaged over both 1-second and 5-second intervals in comparison to a radio-frequency ITS for the quantification of key aspects of wheelchair rugby mobility performance. A secondary aim was to investigate the times and distances spent in specified speed zones, to determine whether the MDL could be an appropriate device for monitoring performance and prescribing training programmes for wheelchair rugby players.

2. Methods

2.1. Participants. Eleven elite male wheelchair rugby players (age = 26 ± 6 years; body mass = 61.3 ± 10.5 kg) volunteered to participate in the current investigation. All participants were members of a national wheelchair rugby squad. Ethical clearance was approved by the university's local ethical advisory committee and all participants provided their written, informed consent prior to data collection.

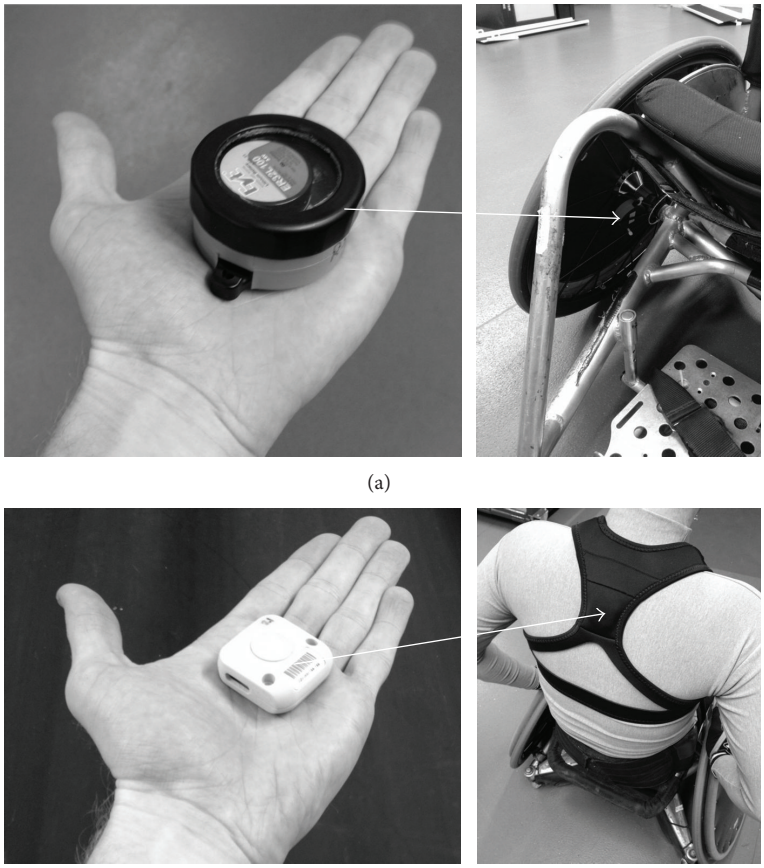
2.2. Equipment. Participants were tested in their own customised rugby-specific wheelchairs. These wheelchairs varied in mass (16.2 to 19.3 kg), wheel diameter (24 to 25 inch), and camber angle (16° to 19°). Tyre pressures were also self-selected and specific to each individual (110 to 150 psi).

2.2.1. Miniaturised Data Logger. Two magnetic reed switch MDL were used for each participant during data collection. The MDL, which weighs 96 g, was attached near the axle of both left and right main wheels (Figure 1(a)). As previously described by Tolerico et al. [4], each MDL is powered by a single 1/6D wafer-cell lithium battery. In brief, the MDL measures wheel rotation using three reed switches at 120° intervals, which are attached to a printed circuit board, with a magnet located at the bottom of a pendulum. Each time the wheel rotates, the magnet passes a reed switch and a time stamp is recorded to the nearest 0.10 second. Three reed activations in sequence relate to one wheel rotation. Using the dimensions of each participant's wheel, the following equation was used to calculate distance:

> Distance (m) = No. of reed switch activations · 1/3 wheel circumference [8].

Mean speed was simply calculated as the total distance covered divided by the total playing time. A customised MATLAB programme was used to compute these variables along with peak speed. Within this programme, all variables were analysed over both 1-second (MDL-1) and 5-second (MDL-5) intervals. During MDL-1 peak speed was calculated as the "average" speed for each time stamp activated in 1-second periods. Alternatively the peak speed calculated during MDL-5 referred to the "average" speed of each time stamp across 5-second intervals, as previously adopted by Sindall et al. [6–8].

2.2.2. Indoor Tracking System. The ITS (Ubisense, Cambridge, UK) is a wired, radio-frequency-based tracking system, which provides positioning data in real time. The ITS is comprised of six sensors that communicate wirelessly with small (40 × 40 × 10 mm), lightweight (25 g) tags. The



(a)

(b)

Figure 1: Illustration of (a) MDL and its positioning and attachment on the inside of a wheelchair rugby wheel and (b) a tag and its location in a GPS vest for the ITS.

six sensors are positioned high (approximately 4 m) around the perimeter of the court, with each sensor orientated with approximately 40° pitch and exactly 0° yaw. The tags, which are housed in a GPS vest worn by the participants (Figure 1(b)), emit ultrawideband radio-frequency signals. The angle-of-arrival and time-difference-of-arrival of these signals are detected by the sensors to determine an accurate tag location. All tags were set to record at 8 Hz, meaning that a location was obtained for each tag every 0.125 seconds. Raw data were filtered using a 3-pass sliding average filter with a window width proportional to the tag frequency.

2.3. Procedures. Data were collected from a total of four simulated wheelchair rugby matches, played over 4 × 8-minute quarters on separate days. Data were collected using the MDL and ITS from a minimum of three to a maximum of five participants at a time. Only data from full quarters were analysed, which gave a total of 42 quarters where data were collected from both devices. During each quarter the total distance covered, mean, and peak speeds were analysed from each device. In addition to this, the time spent and distance covered in the following five relative speed zones were analysed, which were derived using the peak speed (V_{max}) obtained by both devices [20]:

(i) <20% V_{max}—"very low,"

(ii) 20–50% V_{max}—"low,"

(iii) 51–80% V_{max}—"moderate,"

(iv) 81–95% V_{max}—"high,"

(v) >95% V_{max}—"very high".

The absolute time spent and distances covered in 8 arbitrary speed zones ranging from 0 to 4 $m \cdot s^{-1}$ at 0.5 $m \cdot s^{-1}$ intervals was also analysed, as previously defined by Sindall et al. [7].

2.4. Statistical Analysis. Data were analysed using the Statistical Package for Social Sciences (SPSS version 21.0). Mean ± SD were calculated for all performance variables collected from both the MDL and ITS. All data were checked for normality using Shapiro Wilk's tests. The mean differences between devices were then explored using either a paired samples t-test for normally distributed data or Wilcoxon's signed rank tests where assumptions of normality were violated. Significant differences ($P < 0.05$) identified whether significant systematic bias existed between MDL-1 and MDL-5 in relation to the ITS. To explore the reliability, the absolute differences of each performance parameter between the ITS and both MDL-1 and MDL-5 were first compared to the mean

Table 1: Differences in performance parameters assessed during full quarters of wheelchair rugby. All values are mean ± SD.

	ITS	MDL-1	MDL-5
Distance (m)	1403 ± 168	1399 ± 187	1401 ± 186
Mean speed ($m \cdot s^{-1}$)	1.26 ± 0.10	1.26 ± 0.13	1.26 ± 0.14
Peak speed ($m \cdot s^{-1}$)	3.91 ± 0.32	3.85 ± 0.45	2.75 ± 0.29[b,c]

Significant differences between devices are represented by
[a]ITS and MDL-1,
[b]ITS and MDL-5,
[c]MDL-1 and MDL-5.

Table 2: The 95% Limits of Agreement for the times spent and distances covered in arbitrary speed zones for both MDL-1 and MDL-5 compared to the ITS. The 95% Limits of Agreement are presented as systematic bias ± random error.

	Arbitrary speed zones					
	1	2	3	4	5	6
MDL-1						
Times (s)	20 ± 86*	6 ± 101	−107 ± 74*	94 ± 54*	1 ± 70	3 ± 49
Distances (m)	−27 ± 24*	21 ± 78*	−124 ± 84*	160 ± 86*	0 ± 167	0 ± 117
MDL-5						
Times (s)	−24 ± 77*	−48 ± 116*	55 ± 114*	53 ± 122*	3 ± 78	−22 ± 50*
Distances (m)	−22 ± 24*	−29 ± 89*	81 ± 135*	113 ± 132*	7 ± 152	−68 ± 130*

*denotes a significant systematic bias in relation to ITS. No statistical tests were performed for zones 7 and 8 due to the insufficient sample of athletes who registered speeds in these zones for each device.

values and checked for normality. A Pearsons correlation was performed on all normally distributed data, whereas Spearmans ranks were performed if conditions for normality were not satisfied to determine if heteroscedasticity was present. Low and insignificant correlations revealed that the current data were homoscedastic meaning that the 95% Limits of Agreement (95% LoA) could be calculated and reported in raw units. Bland-Altman plots were used to investigate the spread of the data with 95% LoA (presented as systematic bias ± random error) used to investigate the reliability of the MDL-1 and MDL-5 for distance covered, mean speed, peak speed, and the times spent and distances covered in arbitrary speed zones. The time and distance in speed zones relative to peak speed were compared between devices using paired sample t-tests. Statistical significance was accepted when $P < 0.05$.

3. Results

No significant differences in distance covered or mean speeds existed between devices (Table 1). Although no significant systematic bias was observed for these parameters between the ITS and both MDL-1 (distance: −3 m, $P = 0.767$; mean speed: 0.00 $m \cdot s^{-1}$, $P = 0.984$) and MDL-5 (distance: −2 m, $P = 0.875$; mean speed: 0.00 $m \cdot s^{-1}$, $P = 0.951$), random error was present. As shown by the Bland-Altman plots in Figure 2, random errors of ±135 m (MDL-1) and ±133 m (MDL-5) were revealed for distance covered, which compared to the mean value (1401 m) represented a relative random error of 10%. Random errors of ±0.15 $m \cdot s^{-1}$ were revealed for both MDL-1 and MDL-5, which equated to 12% random error.

Furthermore in Table 1, peak speeds were significantly lower in MDL-5 compared to ITS and MDL-1 ($P < 0.0005$). The significant systematic bias revealed for peak speed in MDL-5 (−1.16 $m \cdot s^{-1}$) was accompanied by random errors of ±0.66 $m \cdot s^{-1}$ (20%). Although no significant systematic bias was identified for MDL-1 (−0.06 $m \cdot s^{-1}$, $P = 0.496$) random errors of ±0.85 $m \cdot s^{-1}$ existed (Figure 2). This random error reflected 22% of the mean, peak speed value between MDL-1 and the criterion (3.88 $m \cdot s^{-1}$).

Differences in peak speed were shown to influence the time spent and distance covered in the 5 relative speed zones (Figure 3). MDL-5 significantly underestimated the time spent and distance covered in both "very low" and "low" speed zones compared to ITS and overestimated the times and distances in "moderate," "high," and "very high" zones ($P < 0.0005$). Significantly less time was spent in "very low" and "high" speed zones and greater time was spent in "low" speed zones for MDL-1 ($P \leq 0.035$). Distance covered was also significantly reduced in "very low," "high," and "very high" speed zones and yet increased in "low" zones for MDL-1 ($P < 0.0005$).

The mean time spent and distances covered in arbitrary speed zones independent of V_{max} are displayed for each device in Figure 4. MDL-5 demonstrated a significant systematic bias for both the times and distances in each speed zone except for speed zone 5 (Table 2). Less time and distance were reported in speed zones 1, 2, 6, 7, and 8, with greater times and distances revealed in zones 3 and 4 compared to the ITS (Figure 4). Alternatively, MDL-1 only displayed a significant systematic bias for the time spent in zones 1 ($P = 0.002$), 3 ($P < 0.0005$), and 4 ($P < 0.0005$) and the distances covered in zones 1 ($P < 0.0005$), 2 ($P = 0.001$), 3 ($P < 0.0005$), and

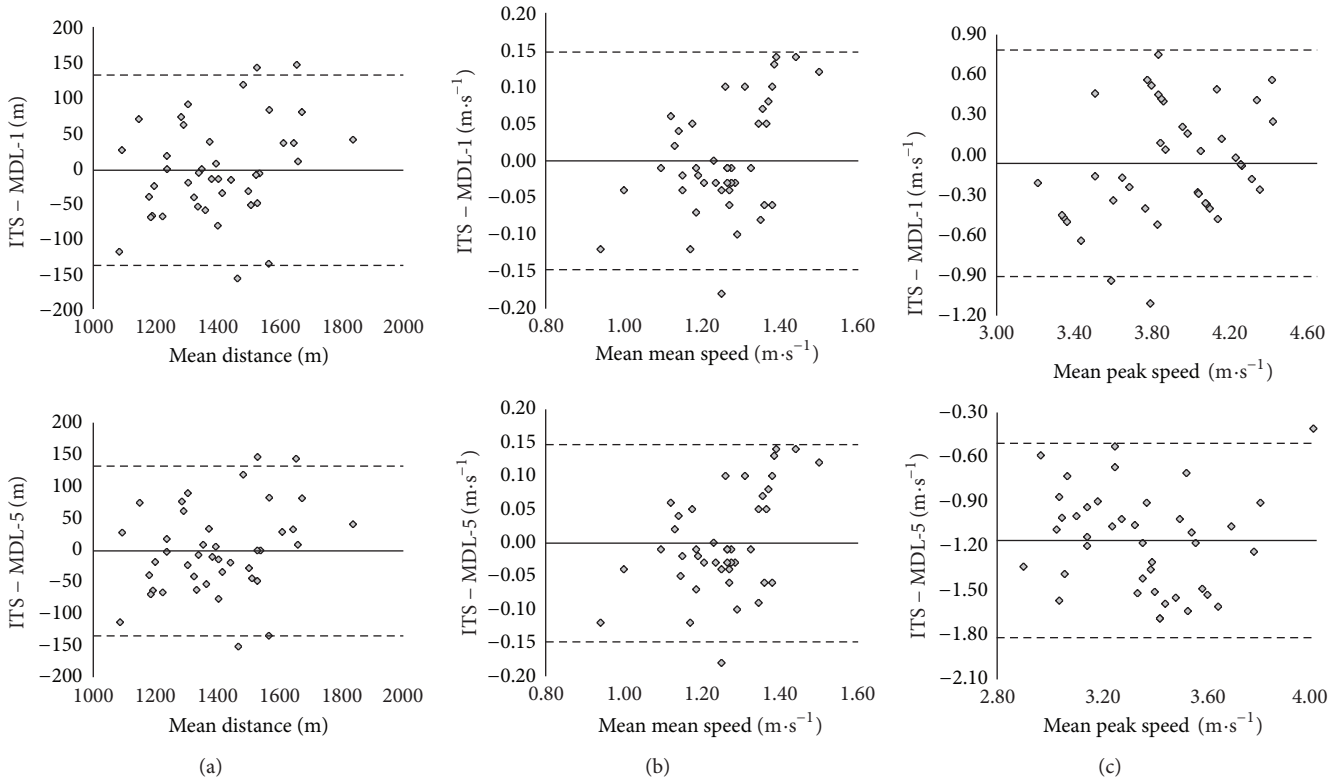


Figure 2: Bland-Altman plots demonstrating the 95% Limits of Agreement between the ITS and MDL-1 (top) and MDL-5 (bottom) for (a) distance covered, (b) mean speed, and (c) peak speed. Solid lines represent systematic bias. Dashed lines represent random error.

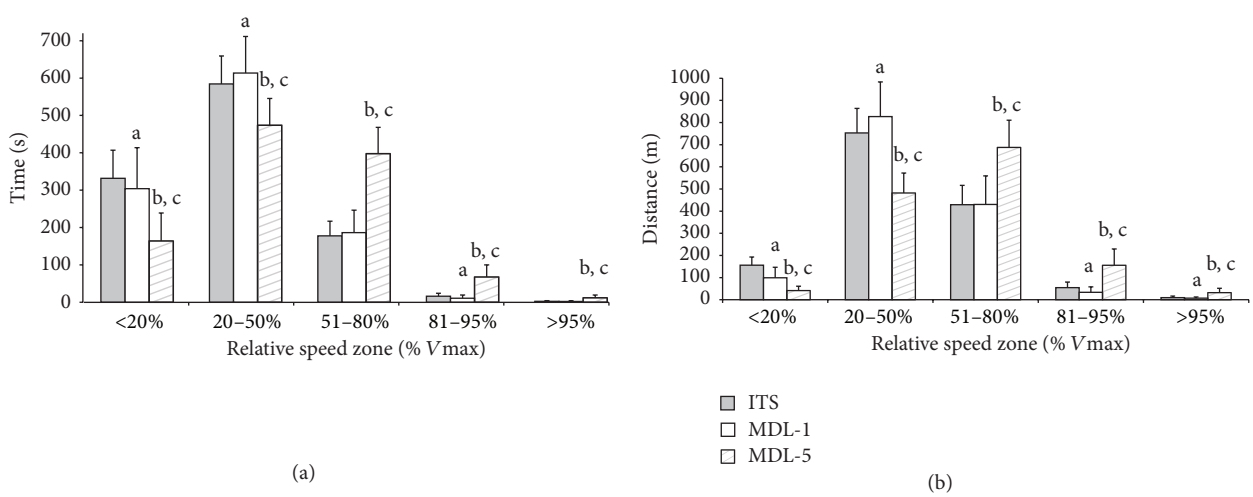


Figure 3: Time spent (a) and distance covered (b) in speed zones relative to the V_{max} of each device. All values are mean ± SD. Significant differences between devices are represented by [a]ITS and MDL-1, [b]ITS and MDL-5, and [c]MDL-1 and MDL-5.

4 ($P < 0.0005$). Figure 4 also illustrated the random error experienced between both the MDL-1 and MDL-5 with the ITS for the times and distances in each arbitrary speed zone, with a greater error reported in MDL-5. It was also revealed that the random error for both devices increased as a function of speed, with a sharp increase in error revealed for both times and distances between zones 4 and 5 (Figure 4). However random errors never dropped below 23% for times and 21% for distance in any of the speed zones.

4. Discussion

The results of the current study revealed that there was no significant differences between both MDL-1 and MDL-5 and the ITS for the distances covered or the mean speeds reached during competitive wheelchair rugby. No significant difference existed for the detection of peak speeds between MDL-1 and the ITS; however, a systematic bias was revealed for MDL-5, which significantly underestimated peak speeds.

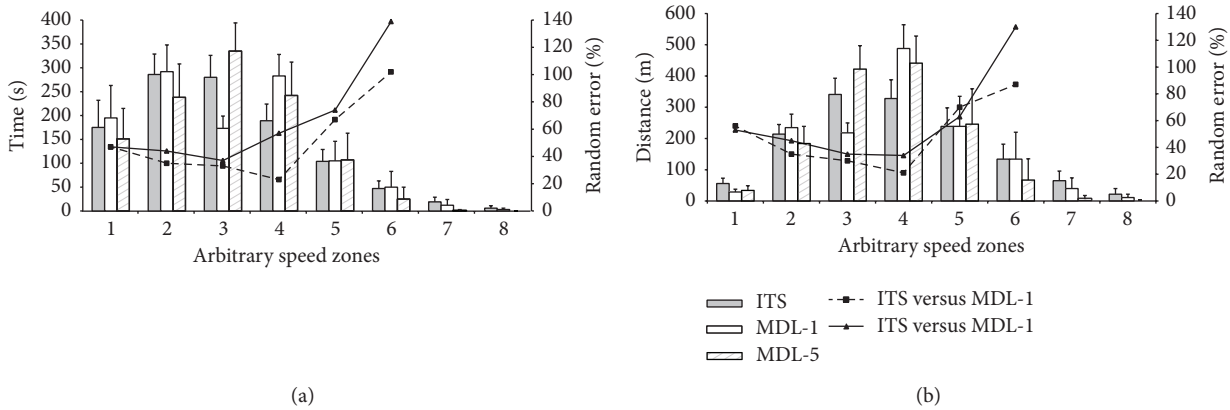


Figure 4: Mean ± SD values for the time spent (a) and distance covered (b) in 8 arbitrary speed zones by each device. The overlaying line graph represents the random error (reported as a % of the mean value). Note that random errors were not calculated for zones 7 and 8 due to insufficient data in these zones for MDL-5. Zones are defined as follows: 1 (0–0.5 m·s^{-1}), 2 (0.5–1.0 m·s^{-1}), 3 (1.0–1.5 m·s^{-1}), 4 (1.5–2.0 m·s^{-1}), 5 (2.0–2.5 m·s^{-1}), 6 (2.5–3.0 m·s^{-1}), 7 (3.0–3.5 m·s^{-1}), and 8 (3.5–4.0 m·s^{-1}).

Although significant systematic bias was not evident for MDL-1 for the assessment of distance, mean and peak speeds, random errors ranging from 10 to 22% still existed. Similar random errors 10–20% were also observed in MDL-5 for these parameters. These errors had a negative effect on the calculation of time spent and distance covered in both the relative and arbitrary speed zones, particularly for MDL-5, which questions the efficacy of using an MDL for the monitoring and prescription of wheelchair rugby training programmes.

The present findings suggest that the MDL is a suitable device for quantifying the distances covered and the mean speeds of sports wheelchair propulsion. These were the type of parameters that the MDL was initially developed to measure in daily life wheelchair users to gain a basic understanding of the amount of physical activity individuals were performing over relatively long time periods [4]. The current study revealed that during a more dynamic application, such as wheelchair rugby, the MDL still provided a reasonable representation of distances and mean speeds, whereby no significant systematic bias existed. Although random errors up to ±135 m and ±0.15 m·s^{-1} were observed for distance and mean speed, respectively, these errors never exceeded 12% in relation to the mean values. Therefore, it could be argued that sport science practitioners could use the MDL to quantify and monitor the total distance and mean speed during different wheelchair rugby training sessions, on the assumption that the data is interpreted with caution given the magnitude of random error present.

The selection of 1-second (MDL-1) or 5-second (MDL-5) analysis intervals was also not shown to be a key consideration when examining distance covered and mean speed, since no significant differences were observed. This finding is a derivative of how the MDL functions. Since the sum of time stamps counted and the overall time of the activity are all that is used to establish distance and mean speed, the effect that different analysis intervals has is negligible. As Sindall et al. [8] mentioned, any underestimations in distance (and subsequently mean speed) were the likely result of missed time stamps during wheel revolutions. Each time such an event occurs one-third of a wheel circumference is missed from the total distance covered. Although it is unclear as to exactly why this event occurs, this is a limitation currently associated with the MDL. Alternatively there are occasions when the distance measured by the MDL overestimated the distance and mean speed in relation to the criterion, hence the presence of random error. An explanation for this event could not be provided by Sindall et al. [8] during wheelchair tennis applications. However, during wheelchair rugby, where impacts with other wheelchairs form a key part of the sport, it is possible that additional time stamps from the same reed switch are activated as the result of a collision, when in reality the wheel is not revolving. Further investigations would be warranted to determine whether this was the cause of overestimations in data and if so amendments would need to be written into the data processing software. Sindall et al. [8] proposed that the future development of a potential six reed switch MDL may further improve the precision of distance and speed measurements. Such a development has since taken place with a six reed data logger combined with a gyroscope in a "wheel rotation monitor," although the accuracy of the reed switch device was not investigated [21]. A six reed switch MDL may reduce the frequency of underestimations in distance. For example, if a reed switch fails to register a time stamp, the distance and time before the next possible reed switch will be reduced, as these would now be positioned at 60° intervals as opposed to 120°. However, this development is unlikely to minimise the overestimations observed in distance and mean speed.

Although the use of 1-second or 5-second analysis intervals had no meaningful effect on the distance covered or the mean speed reached, it did have a significant bearing on the detection of peak speed. Systematic bias of −1.16 m·s^{-1} was identified for MDL-5, which demonstrated a significant underestimation of peak speed in relation to the ITS. The rationale for this finding can again be attributed to the way

in which MDL data is analysed. For MDL-5, the peak speed at each time stamp within a 5-second period is averaged to ultimately give a "mean" peak speed. This "mean" peak speed is always likely to underestimate the "true," instantaneous peak speed, since athletes would have to maintain the peak speed for 5 seconds when adopting the approach used by MDL-5. This was clearly unlikely given the intermittent nature of wheelchair rugby, where short duration bursts of high intensity activity are performed repeatedly [5, 10, 22]. This type of error was reduced when data was analysed at 1-second intervals, since no significant systematic bias was observed between the MDL-1 and ITS for the detection of peak speed. By averaging the speed of each time stamp over shorter intervals, mean underestimations of 0.06 $m \cdot s^{-1}$ were observed. Despite this reduction in systematic bias for MDL-1, random errors of $\pm 0.85\ m \cdot s^{-1}$ were still revealed. This finding demonstrated that although MDL-1 slightly underestimated the mean peak speeds, there were individual instances where MDL-1 actually overestimated the peak speed in order to account for the random errors. Again, the rationale for underestimations was clear since the peak speeds were subsequently speeds "averaged" over a 1-second time period. Overestimations in peak speed, which were not anticipated, were likely to be attributed to the way in which speed calculations are influenced by the wheel dimensions and the time stamps between reed switch activations. For example, an athlete with a 24-inch wheel, one reed switch activation, relates to a distance of approximately 0.62 m. If this event occurs over 0.10 seconds, a speed of 6.20 $m \cdot s^{-1}$ will be registered, whereas if it occurs over 0.20 seconds the speed is halved to 3.10 $m \cdot s^{-1}$. Therefore only fixed speeds of relatively large margins, which remain a function of wheel size, are obtainable. A peak speed of 6.20 $m \cdot s^{-1}$ seems highly unlikely based on previous research, which assessed the peak linear speeds of wheelchair rugby players during 15 m sprints [23]. Yet, if in reality the time difference between reed switch activations was in fact 0.14 seconds, the speed registered would have been 4.43 $m \cdot s^{-1}$, which is more feasible. Unfortunately, because the MDL can only register time stamps to the nearest tenth of a second, this activity gets severely rounded up (or down), which is why the designers advise against reporting instantaneous speeds. However, one or two of these "rounded up" speeds within a 1-second interval could easily account for the instances where peak speeds are overestimated in MDL-1. Regardless of the mechanisms responsible for the random error in peak speed, these errors equated to 22% and 20% for MDL-1 and MDL-5, respectively, implying that the use of an MDL, irrespective of analysis intervals, should not be used for the detection of peak speeds. This is due to the fact that large random errors between scientific equipment are a greater concern to researchers than systematic bias since the magnitude and direction of the error is not consistent and subsequently hard to control for [24].

Errors in the detection of peak speed leads to even greater errors when determining the time spent and distance covered in speed zones relative to peak speed. This approach has recently been adopted in able-bodied rugby union, whereby relative speed zones have been implemented to monitor and modify training programmes [19, 20]. The results of the current study suggest that the MDL is not an appropriate tool for similar use in wheelchair rugby. It was revealed that MDL-5 underestimated the time spent and distance covered in "very low" and "low" speed zones, yet it overestimated these parameters in "moderate," "high," and "very high" speed zones. Work-rest ratios have previously been defined as the time spent in ≥moderate speed zones (work) in relation to ≤low speed zones (rest), which can be used to plan and monitor the intensity of training sessions [25, 26]. If these principles were applied to the context of the current investigation, MDL-5 would increase the amount of work athletes would be prescribed and reduce their rest. The consequences of this overestimation in workload could potentially place athletes at a far greater risk of injury. Unlike MDL-5, no such pattern was revealed for MDL-1, whereby significant reductions in time and distance were recorded in "very low" and "high" speed zones, whereas increases were revealed in "low" speed zones. Subsequently, work-rest ratios based on this analysis would have appeared much closer to those recommended by the ITS. Despite this, these results must still be interpreted with caution. For instance, the absence of any significant difference between MDL-1 and the ITS at "very high" speed zones was most likely due to the minimal times and distances spent in these zones by athletes (2 ± 2 seconds). Also, given the similarities in total distance between devices, an underestimation of time/distance in one or more speed zones would be counteracted by overestimations in other speed zones. In light of these irregularities, it was clear that MDL averaged over 1 or 5-second would not be appropriate for monitoring athletic performance longitudinally during training sessions.

The errors observed in speed zones relative to peak speed were also inherent within arbitrary speed zones. The results revealed that random errors for the times and distances were relatively high (≥21%) across all speed zones, yet this error clearly increased between zones 4-5 (Figure 4). Zone 5 incorporated speeds ranging from 2.0 to 2.5 $m \cdot s^{-1}$ and therefore this increase in error is not too dissimilar to the 2.5 $m \cdot s^{-1}$ threshold that Sindall et al. [8] reported increases in error to occur with the MDL. Random errors were large for both MDL-1 and MDL-5, although these errors were clearly exacerbated in high speed zones in MDL-5. This was likely to be due to the fact that the "mean" peak speed revealed for MDL-5 was only 2.75 $m \cdot s^{-1}$, so very little activity was registered at zones in excess of this speed. As a result of the limited activity registered by MDL-5 at high speeds, comparisons could not be made between devices at the highest speed zones (zones 7 and 8). Consequently, greater activity was registered in the lower speed zones 3 and 4 (1.0–2.0 $m \cdot s^{-1}$). These findings further reiterate that in its current form, the MDL is suitable solely for the quantification of distance and mean speed. Whenever the time spent and distance covered in speed zones (relative or arbitrary) are of interest an MDL is not an appropriate tool due to its deficiencies at high speeds. As previously mentioned, future developments to the design of the MDL, which include 6 reed

switches and a greater frequency for recording time stamps, may increase the precision of the device to make it an effective tool in wheelchair sport settings.

5. Conclusions

The current study revealed that the MDL can be considered an acceptable tool for monitoring the distance covered and mean speed during wheelchair rugby applications. However, the MDL is currently not capable of accurately or reliably reporting the peak speeds produced by elite wheelchair rugby players. Consequently, it is not recommended that the MDL is used for the prescription or monitoring of training programmes, since it is also not capable of accurately determining the time spent and distance covered in certain speed zones.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

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