Department of Veterans Affairs

JRRD Volume 41, Number 3B, Pages 385–394
May/June 2004
Journal of Rehabilitation Research & Development

The biomechanics of wheelchair propulsion in individuals with and without upper-limb impairment

Margaret A. Finley, PhD, PT; Elizabeth K. Rasch, MS, PT; Randall E. Keyser, PhD; Mary M. Rodgers, PhD, PT
University of Maryland School of Medicine, Department of Physical Therapy and Rehabilitation Science, Baltimore, MD; Department of Veterans Affairs Maryland Health Care System, Research and Development Service, Baltimore, MD

Abstract—We used an instrumented wheelchair ergometer and 3D motion analysis system to collect joint kinematic and temporal data, as well as hand rim and joint kinetics, in 47 manual wheelchair users (MWCUs) (15 with upper-limb impairment and 32 without upper-limb impairment). The group with upper-limb impairment propelled with a higher stroke frequency and reduced hand-rim contact time, and smaller peak joint angles and joint excursion of the wrist, elbow, and shoulder during the contact phase. They also propelled with a reduced power output and reduced hand-rim propulsive and resultant forces, moments, and joint compressive forces. We concluded that these kinematic and kinetic strategies might be a mechanism for allowing MWCUs with upper-limb impairment to remain independent. Additionally, the reduced joint excursion and reduced magnitude of forces may protect them from the development of secondary upper-limb pathologies.

Key words: biomechanics, upper-limb impairment, wheelchair propulsion.

INTRODUCTION

Using data from the 1994–95 National Health Interview Survey Disability Supplement, LaPlante reported that over 1.6 million noninstitutionalized persons in the United States report use of a wheelchair (manual, electric, scooter) [1]. Conditions resulting in wheelchair use among adults (18–64 years of age) included multiple sclerosis (9.2% of all adults using any assistive device), paraplegia (7.0%), cerebrovascular accident (7.0%), and tetraplegia (5.0%), with osteoarthrosis, loss of lower limb, and cerebral palsy (CP) accounting for another 13.8 percent of the adult population [2]. Persons with multiple sclerosis, cerebrovascular accident, quadriplegia, CP, and rheumatoid arthritis may also have upper-limb impairments, including paralysis or paresis, sensory impairments, motor control deficits, pain, or restricted joint motion. Of the adults with one of the leading conditions associated with wheelchair or scooter use described by LaPlante, more than 50 percent have conditions that could result in upper-limb impairment [1]. In Scotland, Perks et al. reported that nearly 15 percent of a sample of 700 manual wheelchair users (MWCUs) had upper-limb impairment, indicating that a significant number reported difficulties with wheelchair propulsion due to impaired upper-limb function [3].

Abbreviations: ANCOVA = analysis of covariance, ANOVA = analysis of variance, CP = cerebral palsy, FEF = fraction of effective force, HSD = honestly significantly different, MCP = metacarpal, MWCU = manual wheelchair user, SB = spina bifida, SCI = spinal cord injury, SSXT = single-stage exercise test, VA = Department of Veterans Affairs.

This material was based on work supported by Department of Veterans Affairs Rehabilitation Research and Development Service, Merit Review Board project B2168RA.

Address all correspondence to Margaret A. Finley, University of Maryland School of Medicine, Department of Physical Therapy and Rehabilitation Science, 100 Penn Street, AHB, 115, Baltimore, MD 21201-1082; 410-706-0856; fax: 410-706-6387; email: mfinley@som.umaryland.edu.

Over the last decade, researchers have investigated many aspects of manual wheelchair propulsion, predominantly in individuals with paraplegia resulting from spinal cord injury (SCI). Only a few investigations addressing the biomechanical characteristics of MWCUs have included individuals with tetraplegia. Wheelchair propulsion temporal characteristics [4], upper-limb kinematics [5], hand-rim force application [6,7], and shoulder joint kinetics [8] have been studied in MWCUs with various levels of SCI, including those with cervical level injuries. These studies suggested that the biomechanics of wheelchair propulsion varied in relation to the subjects' levels of SCI [4–8]. Little information has been reported on functional abilities and propulsion characteristics of MWCUs with upper-limb impairments from conditions other than SCI. Only one study was found of the biomechanical characteristics of wheelchair propulsion in a group of individuals (n = 67) with a variety of conditions, noting reported differences among disability groups [9].

Curtis et al. found that a greater proportion of individuals with tetraplegia experienced shoulder pain and scored higher on the Wheelchair Users Shoulder Pain Index, as compared to paraplegic subjects [10]. Despite pain and upper-limb impairment, these individuals remained functionally independent in community and home mobility. Clearly, MWCUs have upper-limb impairment and continue to use a manual wheelchair independently. However, how this is accomplished from a biomechanical perspective is not known. Information regarding the biomechanics of manual wheelchair propulsion in this population will increase the overall knowledge base about performance of the task in wheelchair users and may provide insight into mechanisms of secondary pathologies. Therefore, it was hypothesized that biomechanical differences would exist between MWCUs with and without upper-limb impairment during wheelchair propulsion. This study compared the biomechanical characteristics of manual wheelchair propulsion in a diverse sample of MWCUs with and without upper-limb impairment.

METHODS

Subjects

Forty-seven MWCUs (15 with upper-limb impairment and 32 without upper-limb impairment) participated in the study. Inclusion criteria required use of a manual wheelchair for a minimum of 1 year prior to participation, and wheelchair use for the majority (over 50%) of home and community mobility. Prior to participation, wheelchair users were medically screened for participation. Individuals with upper-limb orthopedic disorders, systemic disease, ventilatory disorders, or pain were excluded from participation. In addition, a physical therapist with expertise in neurological rehabilitation repeated the upper-limb portion of the exam for the purpose of group classification. Subjects were classified with upper-limb impairment if paresis of the upper limbs (determined by manual muscle testing, score of ≤3/5), sensory deficits, and/or motor control deficits (determined by assessing rapid alternating movements and pointing accuracy) were noted. The presence of paresis, impaired sensation, or impaired motor control effectively identified all participants with upper-limb impairment. **Table 1** provides the demographic details of the groups. Medical conditions of the participants that necessitated wheelchair use were intentionally diverse to represent the variety present in the wheelchair user population.

Instrumentation

Following the attaining of informed consent and medical screening, all subjects completed a single-stage wheelchair ergometer exercise test to exhaustion. We performed the testing on a prototypical wheelchair ergometer (**Figure**). The ergometer has a 22-inch diameter hand rim, along with an adjustable seat height and seat width to match the individual's personal wheelchair. A chain and sprocket system connected the wheelchair axle to a flywheel at the rear of the ergometer. We created the system with a nylon belt to which known loads were applied for precise control of the resistance.

The wheelchair hand rims were instrumented with PY-6 force and torque transducers in the hubs (Bertec Corp, Worthington, OH). Bonded strain gauges measured hand rim forces and torques in 3D, with a maximal torque of 150 N•m and a maximum force measure in the plane-of-wheel of 3,500 N. Orientations of the right transducer coordinates were tangential (Fx, positive for forward), radial (Fy, positive for superior), and medial-lateral (Fz, positive for lateral). A potentiometer measured angular position of the wheel, transducer, and hand-rim assembly. The electrical signals from the transducer and potentiometer were collected with analog-to-digital converter and acquisition software (Peak Performance Technology, Englewood, CO). Hand-rim temporal, kinetic, and potentiometer data were collected at 360 samples/channel/second. Visual feedback of propulsion velocity was provided

Table 1.
Demographic characteristics of all participants.

Variable	Without UL Impairment (*n* = 32) Frequency or Mean ± SD		With UL Impairment (*n* = 15) Frequency or Mean ± SD	
Gender	26 males	6 females	10 males	5 females
Age (yr)	39.5 ± 9.9		37.6 ± 6.7	
Duration Wheelchair Use (yr)	10.0 ± 8.4		10.5 ± 8.2	
Height (cm)	170.0 ± 14.2		170.4 ± 14.6	
Weight (kg)	74.6 ± 15.8		66.4 ± 10.5	
Conditions of Participants Related to Wheelchair Use				
Cervical SCI	1 (incomplete)		6	
Thoracic SCI	17		0	
Lumbar SCI	3		0	
Spina Bifida	4		0	
Multiple Sclerosis	1		2	
Cerebral Palsy	1		3	
Other	5*		4†	

*Bergers disease, multitrauma (2), tropical spastic myelopathy, polio
†Arthrogryposis, spinal cerebellar atrophy, Guillian-Barre, brain abscess
SD = standard deviation
UL = upper-limb

Figure.
Instrumented wheelchair ergometer and 3D motion analysis system used for test setup.

by a bicycle speedometer (Cateye Cyclocomputer, Model CC-CD 100, Osaka, Japan) with a digital display attached to the right wheel and placed in view of the participant (accuracy = ±0.2 km/h, ±1 rpm). The relatively slow propulsion velocity of 3.4 km/h was selected because of the inclusion of individuals with upper-limb impairment and because it was within the ranges used in studies of subjects with cervical SCI [6,8,11].

We used three video cameras and the Motus video acquisition system (Peak Performance Technology, Engelwood, CO) to measure upper-limb and trunk movements, collected at 60 frames/s. The accuracy of the video acquisition system for angular measures was reported to be 1° at an angular speed of 300°/s [12]. Retroreflective markers were placed bilaterally on the dorsal surface of the fifth metacarpal (MCP) head, medial styloid process, lateral styloid process, radial head, acromion, and greater trochanter. The 12-marker system was used to calculate trunk flexion/extension (acromion and greater trochanter, measured against a stationary horizontal plane), shoulder flexion/extension (radial head, acromion, greater trochanter [hip] from sagittal view), shoulder abduction/adduction (same for flexion from frontal view), elbow flexion/extension (acromion, radial head, ulnar styloid), wrist flexion/extension (radial head, ulnar styloid, fifth MCP, frontal view), and supination/pronation (radial head, ulnar styloid, fifth MCP from the sagittal view) angles. The velocities and accelerations were calculated from these angles.

We used a 3D rigid-body segment model to determine upper-limb segment and joint kinematics and joint kinetics with a Newton-Euler inverse dynamics procedure, using custom-written routines [13]. Definitions of the global and local hand rim and segmental coordinate systems are described in detail elsewhere [14]. A second-order Butterworth digital filter was used to smooth the raw 3D coordinate data at a cutoff frequency of 6 Hz. Contact phase was defined as the entire time during which hand rim loading occurred. Input variables for the inverse dynamics program that calculated joint forces and moments included the motion vectors (angular displacement, angular velocity, angular and linear acceleration), hand rim forces and torques, and anthropometric data. The program then calculated the joint forces and moments of the wrist, elbow, and shoulder.

Exercise Tests

All subjects performed a single-stage exercise test (SSXT) to exhaustion. Resistance for the submaximal test was set at as 60 percent the peak load achieved on the individual's previous maximal exercise test. **Figure** shows a subject in the wheelchair ergometer, demonstrating the test setup. For the SSXT, subjects propelled the ergometer at 32 rpm (3.4 km/h). The test ended when the subject could no longer maintain the designated velocity, despite strong verbal encouragement. We collected propulsion variables (temporal, hand rim, kinematic) when the load was initially applied (fresh) and just prior to exhaustion (fatigue). The procedures and instrumentation used in this study have been shown to be reliable over repeated applications [15].

Data Reduction

Data from the right upper limb of each subject were used for all analyses. The kinematic and kinetic data were averaged over three cycles (contact-contact) for each condition (fresh, fatigued). Peak hand rim kinetics, joint kinetics, and joint kinematics during wheel contact were used for statistical analysis. Power output was calculated from the applied load, velocity, and the rolling distance of the flywheel. Resultant hand rim forces (F_{rslt}) and moments (M_{rslt}) were calculated. Additionally, the effective force was calculated as the ratio of the tangential force to the resultant force ($\mathrm{Fx}/F_{\mathrm{rslt}}$).

Data Analysis

Demographic characteristics (age, height, weight, duration of wheelchair use) of the groups were analyzed with a *t*-test ($p \leq 0.05$). Data were screened for distribution normality using the Wilks-Shapiro W statistic, and the majority of distributions (>80%) were found to be normal. Given the assumption of population normality rather than sample normality, two-factor analyses of variance (ANOVA) ($p \leq 0.05$) were used to determine if differences existed between the groups (with/without upper-limb impairment), and between states (fresh, fatigued) in temporal, kinematic, and joint kinetic variables. Pearson product moment correlations determined if relationships existed between the hand rim kinetic variables and power output. Since a relationship ($r^2 \geq 0.40$) existed between the hand rim variables and power output, a two-factor analysis of covariance (ANCOVA) was used to determine if differences existed between the groups and states for the hand rim and kinetic variables, with power output as the covariate. Interactions were analyzed post hoc with the use of a Tukey Honestly Significantly Different (HSD) test. Type I error threshold was held at $p = 0.05$.

RESULTS

No difference existed between the two groups for age, height, weight or duration of wheelchair use.

Hand Rim Temporal Characteristics and Kinetics

Regardless of fatigue state or power output, individuals with upper-limb impairment propelled with an increased stroke frequency and decreased contact time compared to those without upper-limb impairment (**Table 2**).

Collapsed across states, the group with upper-limb impairment demonstrated reduced power output relative to the unimpaired group. The tangential (Fx) and resultant (F_{rslt}) hand rim forces were reduced along with the propulsive moment (Mz) and resultant (M_{rslt}) moment in the group with upper-limb impairment after controlling for power output (**Table 3**).

Kinematics

Kinematically, minimal change was noted with fatigue. In the fatigued state, the wrist was in less radial deviation and had less joint excursion across both groups. Across states, the group with upper-limb impairment demonstrated smaller peak joint angles throughout the propulsion cycle in wrist extension, shoulder flexion, and abduction, with greater peak shoulder adduction angle compared to those without upper-limb impairment. (**Table 4**). Shoulder angles in the frontal plane are expressed as relative values, such that less shoulder abduction is also greater shoulder adduction. At the instant of wheel contact, the group with upper-limb impairment showed less wrist extension than the group without upper-limb impairment. At release, the group

Table 2.
Temporal differences between groups (collapsed across states).

Variable	Without UL Impairment (n = 32) Mean ± SD	With UL Impairment (n = 15) Mean ± SD
Stroke Frequency (c/s)*	1.2 ± 0.2	1.5 ± 0.3
Contact Time (s)*	0.5 ± 0.1	0.3 ± 0.1
Contact Time (% cycle)*	44.6 ± 6.7	37.1 ± 10.6

*Significant difference between groups ($p \leq 0.05$) SD = standard deviation UL = upper-limb

Table 3.
Comparison of peak hand rim kinetics between groups (collapsed across states).

Variable	Without UL Impairment (n = 32) Mean ± SD	With UL Impairment (n = 15) Mean ± SD
Power (W)*	57.2 ± 19.7	15.8 ± 10.9
Fx (N)*	68.3 ± 29.0	58.3 ± 30.3
Fy (N)	49.8 ± 30.5	34.6 ± 24.6
Fz (N)	37.6 ± 44.7	22.2 ± 14.6
F_{rslt} (N)*	102.4 ± 42.7	75.5 ± 33.2
Fx/F_{rslt} (%)	69.2 ± 21.2	77.3 ± 16.7
Mx (N•m)	5.5 ± 3.8	5.5 ± 5.0
My (N•m)	3.9 ± 2.4	3.3 ± 3.5
Mz (N•m)*	19.0 ± 7.7	13.1 ± 7.2
M_{rslt} (N•m)*	20.9 ± 7.0	15.1 ± 8.3

*Significant difference between groups after controlling for power output ($p \leq 0.05$) UL = upper-limb

Table 4.
Peak joint angles during propulsion cycle between groups (collapsed across states).

Variable	Without UL Impairment (n = 32) Mean ± SD	With UL Impairment (n = 15) Mean ± SD
Wrist Extension*	36.2° ± 14.9°	26.1° ± 20.6°
Wrist Flexion	9.0° ± 17.7°	11.4° ± 16.9°
Wrist Radial Deviation	5.2° ± 11.8°	1.0° ± 17.5°
Wrist Ulnar Deviation	30.1° ± 12.9°	25.9° ± 19.7°
Elbow Extension	152.5° ± 9.4°	151.6° ± 19.5°
Elbow Flexion	110.5° ± 11.3°	113.2° ± 19.4°
Shoulder Flexion*	5.7° ± 14.7°	–3.7° ± 17.6°
Shoulder Extension	48.0° ± 13.7°	45.1° ± 15.7°
Shoulder Adduction*†	19.5° ± 15.0°	9.4° ± 20.3°
Shoulder Abduction*	39.7° ± 10.6°	30.9° ± 13.2°
Trunk Extension	105.2° ± 12.5°	108.1° ± 12.2°
Trunk Flexion	93.5° ± 15.3°	97.5° ± 17.5°

*Significant difference between groups ($p \leq 0.05$)
†Shoulder adduction is in relative terms, such that smaller values indicate more adduction, larger indicate great abduction.
UL = upper-limb

with upper-limb impairment also had greater wrist extension and smaller shoulder abduction angles (**Table 5**). The total joint excursion during the contact phase of the propulsion cycle was less for wrist flexion/extension and increased at the elbow for the group with upper-limb impairment.

Table 5.
Joint angles at instant of wheel contact and at instant of wheel release between groups (collapsed across states).

Variable	Without Impairment (*n* = 32) Mean ± SD	With Impairment (*n* = 15) Mean ± SD
Contact		
Wrist Flexion (+) – Extension (–)*	–22.6° ± 18.0°	–11.4° ± 19.1°
Wrist Radial (–) – Ulnar (+) Deviation	10.7° ± 13.5°	8.3° ± 13.9°
Elbow Flexion (↓) – Extension (↑)	125.0° ± 12.0°	119.3° ± 20.6°
Shoulder Flexion (+) – Extension (–)	–44.0° ± 18.6°	–39.7° ± 18.5°
Shoulder Adduction (↓) – Abduction (↑)	30.4° ± 17.7°	24.5° ± 21.4°
Trunk Flexion (↓) – Extension (↑)	103.7° ± 12.9°	105.6° ± 12.6°
Release		
Wrist Flexion (+) – Extension (–)*	–1.1° ± 19.3°	–4.8° ± 20.6
Wrist Radial (–) – Ulnar (+) Deviation	17.5° ± 18.5°	18.9° ± 22.6°
Elbow Flexion (↓) – Extension (↑)	135.2° ± 15.2°	133.2° ± 34.2°
Shoulder Flexion (+) – Extension (–)	–7.8° ± 18.7°	–9.3° ± 34.1°
Shoulder Adduction (↓) – Abduction (↑)*	29.2° ± 9.5°	19.1° ± 17.0°
Trunk Flexion (↓) – Extension (↑)	96.4° ± 15.3°	100.3° ± 17.3°

*Significant difference between groups ($p \leq 0.05$) UL = upper-limb

Joint Kinetics

The group with upper-limb impairment demonstrated reduced compressive joint forces in the wrist, elbow, and shoulder joint and less elbow lateral shear than those without upper-limb impairment, regardless of state. There was no difference in the joint moments between the groups (**Table 6**).

DISCUSSION

The hypothesis was supported, in that differences were found in the propulsion biomechanics of individuals with upper-limb impairment relative to MWCUs without upper-limb impairment. The group with impairment demonstrated more frequent wheel contact of shorter duration with smaller joint range of motion. This may have resulted from decreased upper-limb strength or motor control deficits in the group with upper-limb impairment. Alternatively, the velocity employed for testing may have been challenging for participants with impaired motor control. Comparisons between studies are often difficult because of differences in testing procedures, units of measurement, equipment employed, and the characteristics of the sample studied. Kinematic findings are often difficult to compare because of variation in the variables reported, such as peak angles, angles at initial contact, or wheel release. The peak angles as well as the angles at wheel contact and release for the shoulder, elbow, and wrist found in the current study were all similar to those previously reported in the literature, with the exception of peak shoulder flexion, which was reduced in both the MWCU with and without upper-limb impairment [16–18]. The exception of peak shoulder flexion is possibly associated with the definition of motions or the velocity used in this study.

Previous investigations of propulsion mechanics in individuals with impaired upper limbs as a result of cervical SCI show velocity-dependent differences in temporal characteristics [4,8,9,11]. Prior studies allowed the subject to propel at self-selected speeds [4,8], or used different velocities in comparing individuals with different levels of tetraplegia (2 km/h and 3 km/h) [11]. The self-selected velocities of more impaired subjects were reported to be between 2.8 km/h [9] and 2.99 km/h [11].

Additionally, sprint tests have been used in comparing individuals with different levels of SCI, also at a self-determined maximal velocity [6,9]. In these earlier reports, the individuals with higher levels of SCI consistently achieved or selected lower velocities with decreased stroke frequency and increased contact time than those with lower levels of SCI. Results of the current study were different from these previous studies. In the current study, the group with upper-limb impairment

Table 6.
Joint kinetics variables between groups (collapsed across states).

Variable	Without UL Impairment (n = 32) Mean ± SD	With UL Impairment (n =15) Mean ± SD
Forces (N)		
Wrist Anterior Shear	39.9 ± 31.5	42.7 ± 25.8
Wrist Posterior Shear	10.6 ± 22.5	5.4 ± 12.4
Wrist Distraction	2.7 ± 5.9	5.3 ± 9.1
Wrist Compression*	72.0 ± 34.4	49.5 ± 35.2
Wrist Medial Shear	18.5 ± 40.3	6.5 ± 7.1
Wrist Lateral Shear	23.5 ± 19.1	17.3 ± 15.1
Elbow Anterior Shear	50.7 ± 29.1	46.6 ± 24.9
Elbow Posterior Shear	8.5 ±15.9	1.3 ± 4.1
Elbow Distraction	15.4 ± 10.5	16.8 ± 11.5
Elbow Compression*	50.7 ± 45.0	33.3 ± 33.4
Elbow Medial Shear	13.1 ± 29.4	6.3 ± 6.9
Elbow Lateral Shear*	42.1 ± 37.7	32.9 ± 25.0
Shoulder Anterior Shear	58.0 ± 39.2	53.1 ± 29.6
Shoulder Posterior Shear	6.9 ± 8.6	5.5 ± 6.5
Shoulder Distraction	51.9 ± 28.4	47.1 ± 20.5
Shoulder Compression*	17.9 ± 26.0	5.1 ± 13.5
Shoulder Medial Shear	24.6 ± 30.3	16.7 ± 8.8
Shoulder Lateral Shear	31.7 ± 35.7	23.2 ± 23.3
Moments (N•m)		
Wrist Extension	6.21 ± 7.1	6.1 ± 7.1
Wrist Flexion	11.9 ± 16.4	5.7 ± 5.7
Wrist Ulnar Deviation	35.8 ± 16.8	29.8 ± 15.5
Wrist Radial Deviation	0.02 ± 0.1	0.02 ± 0.1
Elbow Extension	36.2 ± 23.2	30.8 ± 20.4
Elbow Flexion	0.7 ± –3.0	0.92 ± 2.2
Shoulder Extension	1.2 ± 3.9	1.7 ± 4.7
Shoulder Flexion	52.1 ± 38.0	46.0 ± 31.9
Shoulder Abduction	13.2 ± 27.2	6.0 ± 7.9
Shoulder Adduction	35.2 ± 27.7	27.3 ± 26.7

*Significant difference between groups ($p \leq 0.05$) UL = upper-limb

demonstrated more frequent wheel contact with shorter duration and decreased joint range of motion than those without impairment. Previous work did not allow for the velocity difference that may have resulted in confounding group findings. The use of a uniform velocity for all subjects was chosen to provide a control within the testing protocol of this study, allowing comparisons of group differences to be determined. A designated propulsion velocity of 3.4 km/h was selected to optimize test performance in the group with upper-limb impairment. However, for some of the subjects with upper-limb impairment, this velocity may have provided too great a challenge, while being too slow for some of the subjects who did not have upper-limb impairment. Results of the current study agreed with the findings of Newsam et al. [5], who reported kinematic differences in wrist and forearm position at wheel contact between subjects with and without cervical SCI. Several of the subjects with upper-limb impairment had cervical SCI with limitations in wrist range of motion that may have contributed to some of the findings in the previous studies.

In addition to temporal and kinematic differences, individuals with upper-limb impairment showed reduced power output along with a reduced resultant force, propulsive moment (Mz), and overall resultant moment on the hand rim. Experimental results have shown that propulsion forces are not tangentially directed [19]. Veeger et al. introduced the term "fraction of effective force" (fraction of effective force, FEF = tangential force component/resultant force magnitude) as a measure of the effectiveness of force application [20]. In the current study, no difference was found between the groups in the ratio of propulsive force to resultant force (effective force), as has been reported by other researchers [7].

Using a sprint test, Woude et al. compared propulsion characteristics in wheelchair athletes of differing disability classifications (cervical SCI, thoracic SCI, lumbar-sacral SCI, CP, amputation, spina bifida [SB], polio) [9]. Similar to the current findings, they report that in the groups of athletes with the greatest upper-limb impairment (SCI cervical vertebra 5-6, SCI cervical vertebra 6-7, CP) resultant hand-rim forces were significantly lower as compared with higher-level classifications. Peak velocities and percentage of effective force achieved on the sprint test were also significantly lower in subjects with tetraplegia at cervical vertebra 5-6 (6.1 km/h, 38%) CP (7.9 km/h, 42%), compared to groups with paraplegia (thoracic vertebra 6-sacrial vertebra 1, polio, SB), (11.2 km/h, 56%), or lower-limb amputation (11.2 km/h, 55%). A highly variable and low effective force was also reported [9]. The effective force in the current study was not significantly different between groups, and was higher than the effective force reported by other investigators, although the FEF has been reported to be as high as 81 percent in submaximal work [19].

Of interest in the current study was that the percentage of effective force for the group with upper-limb impairment was found to be 77 percent, compared to 69 percent for those without upper-limb impairment. The effect-cost balance of wheelchair propulsion has been surmised to result in the pattern that forms the best coordination of the mechanical constraints of the wheelchair and the biomechanical abilities of the individual [19,21]. Although the magnitude of the hand-rim propulsive and resultant forces and moments were all reduced in those with upper-limb impairment, a larger percentage was applied in the propulsion direction. The term "mechanical efficiency" has been used to describe the direction of force application [23].

Historically, mechanical efficiency is the ratio of external energy production to consumed metabolic energy [10]. Medially directed forces have been considered to be ineffective contributions to propulsion, reducing efficiency [22,23,24]. No difference was found in the group with upper-limb impairment in the generation of medially directed forces. However, it has been suggested that individuals with impaired hand function and/or altered upper-limb strength generate increased medial forces to provide the needed friction for maintaining a grip on the hand rim [7]. The individuals with upper-limb impairment in the current study propelled with patterns of reduced joint excursion and reduced contact time with the hand rim, further constraining the user-wheelchair interface and possibly allowing for a larger percentage of tangential force application. Based on conclusions of a recent effect-cost balance study [22], and perhaps given the constraints imposed by an individuals disability in the interaction with the laboratory wheelchair ergometer, MWCUs with and without upper-limb impairment apply forces in the directions that result in their most efficient pattern, allowing them to complete the task. Thus, for those with upper-limb impairment to complete the task of propulsion, they must be able to develop the best cost-effect balance with their efforts, directing the majority of the force application toward forward motion of the wheelchair.

Reduced hand-rim force generation contributed to a reduction in the compressive forces (operationally defined as forces along the vertical axis, *y*), as well as the reduced shear in lateral elbow shear in the individuals with upper-limb impairment. The joint contact forces were not measured. The groups demonstrated no difference in the net joint moments generated during propulsion. Since net moments do not account for the coactivation at the joints, an underestimation of the actual muscle forces may result. However, net joint moments during four planes of motion of the glenohumeral joint have been shown to have a high correlation to glenohumeral joint compression (joint contact loading) forces [25]. We therefore suggest that net shoulder joint moments could be used to predict the mechanical load at the glenohumeral joint [25].

Although no direct relationship has been demonstrated regarding compression forces and glenohumeral joint damage, the inability of individuals with upper-limb impairment to generate these larger forces may be beneficial. The strategy of repetitive lower magnitude forces during the performance of propulsion and other activities of daily living may be a potential mechanism for the development of shoulder overuse pathologies.

The current study has shown that individuals with upper-limb impairment propel with smaller joint excursions, higher stroke frequency, lower power output, smaller propulsive forces, and with a greater percentage in the tangential force application. Although it appears that if only the FEF is considered, this group is more efficient in the current propulsion tasks, many activities of daily living and propulsion skills require the generation of larger forces and moments. The inability of individuals with upper-limb impairment to generate these larger forces, resulting in the need to produce repetitive, lower magnitude forces during the performance of propulsion and other activities of daily living, may be a potential mechanism for the development of shoulder overuse pathologies. Subjects with shoulder pain were excluded from the current study. Conclusions about potential relationships between upper-limb pains, the presence of upper-limb impairment (as defined in this study), and propulsion mechanics could not be made.

CONCLUSION

Although wheelchair users may have neuromuscular impairment of the upper limb, they are able to successfully complete the task of manual wheelchair propulsion, albeit with an overall reduction in biomechanical variables characteristic of MWCUs without upper-limb impairment. These adaptations may be a mechanism for allowing MWCUs to remain independent, and some of these adaptations ultimately may protect them from the development of secondary upper-limb pathologies. Future studies should evaluate differences in the occurrence of painful upper-limb conditions in MWCU with and without upper-limb impairment. In addition, the effect of training on the development of secondary upper-limb pathologies should be investigated, particularly in MWCUs with upper-limb impairments.

REFERENCES

1. LaPlante MP, Hendershot GE, Moss AJ. The prevalence of need for assistive technology devices and home accessibility features. Technol Disabil. 1997;6:17–28.
2. Kaye H, Kang T, LaPlante MP. Mobility device use in the United States. Disability Statistics Report No: 14. San Francisco (CA): National Institute on Disability and Rehabilitation Research; 2000 June.
3. Perks BA, Mackintosh R, Stewart CP, Bardsley GI. A survey of marginal wheelchair users. J Rehabil Res Dev. 1994; 31(4):297–302.
4. Newsam CJ, Mulroy SJ, Gronley JK, Bontrager EL, Perry J. Temporal-spatial characteristics of wheelchair propulsion. Am J Phys Med Rehabil. 1996;75(4):292–99.
5. Newsam CJ, Rao SS, Mulroy SJ, Gronley JK, Bontrager EL, Perry J. Three-dimensional upper-limb motion during manual wheelchair propulsion in men with different levels of spinal cord injury. Gait Posture. 1999;10(3):22–32.
6. Dallmeijer AJ, Kappe YJ, Veeger DH, Janssen TW, van der Woude LH. Anaerobic power output and propulsion technique in spinal cord injured subjects during wheelchair ergometry. J Rehabil Res Dev. 1994;31(2):120–28.
7. Dallmeijer AJ, van der Woude LH, Veeger HE, Hollander AP. Effectiveness of force application in manual wheelchair propulsion in persons with spinal cord injuries. Am J Phys Med Rehabil. 1998;77(3):213–21.
8. Kulig K, Newsam CJ, Mulroy SJ, Rao S, Gronley JK, Bontrager EL, et al. The effect of level of spinal cord injury on shoulder joint kinetics during manual wheelchair propulsion. Clin Biomech. 2001;16(9):51–74.
9. van der Woude LH, Bakker WH, Elkhuizen JW, Veeger HEJ, Gwinn T. Propulsion technique and anaerobic work capacity in elite wheelchair athletes: Cross-sectional analysis. Am J Phys Med Rehabil. 1998;77:222–34.
10. Curtis KA, Drysdale GA, Lanza RD, Kolber M, Vitolo RS, West R. Shoulder pain in wheelchair users with tetraplegia and paraplegia. Arch Phys Med Rehabil. 1999;80(4):453–57.
11. Dallmeijer AJ, Hopman MT, van As HH, van der Woude LH. Physical capacity and physical strain in persons with tetraplegia; the role of sport activity. Spinal Cord. 1996; 34(12):729–35.
12. Schrag D, Rodgers MM. Reliability and accuracy of the peak performance three-dimensional motion measurement system. In: Sixth Annual East Coast Clinical Gait Conference. 1990; Michigan State University; 1990.
13. Rodgers MM, Tummarakota S, Lieh J. Three-dimensional dynamic analysis of wheelchair propulsion. J Appl Biomech. 1998;14:80–92.

14. Rodgers MM, Keyser RE, Rasch, EK, Gorman PH, Russell PJ. Influence of training on biomechanics of wheelchair propulsion. J Rehabil Res Dev. 2001;38(5):505–11.
15. Finley MA, Rodgers MM, Rasch EK, Keyser RE, McQuade KJ. Reliability of biomechanical variables during wheelchair ergometry. J Rehabil Res Dev. 2002;39(1): 73–81.
16. Davis JL, Growney ES, Johnson ME, Iuliano BA, An KN. Three-dimensional kinematics of the shoulder complex during wheelchair propulsion: A technical report. J Rehabil Res Dev. 1998;35(1):61–72.
17. Rao SS, Bontrager EL, Gronley JK, Newsam CJ, Perry J. Three-dimensional kinematics of wheelchair propulsion. IEEE Trans Rehabil Eng. 1996;4(3):152–60.
18. Veeger HE, Meershoek LS, van der Woude LH, Langenhoff JM. Wrist motion in hand rim wheelchair propulsion. J Rehabil Res Dev. 1998;35(3):305–13.
19. Veeger HEJ. Biomechanics of manual wheelchair propulsion. In: van der Woude LH, Hopman MT, van Kemenade CH, editors. Biomedical aspects of manual wheelchair propulsion: The state of the art II. Amsterdam: IOS Press; 1999. p. 86–95.
20. Veeger HEJ, van der Woude LH, Rozendal, EP. Load on the upper extremity in manual wheelchair propulsion. J Electromyogr Kinesiol. 1991;1:270–80.
21. Rozendaal LA, Veeger HEJ, van der Woude LHV. The push force pattern in manual wheelchair propulsion as a balance between cost and effect. J Biomech. 2003;36(2): 239–48.
22. Veeger HEJ, van der Woude LH, Rozendaal RH. Effect of hand rim velocity on mechanical efficiency in wheelchair propulsion. Med Sci Sports Exerc. 1992;24:100–7.
23. Boninger ML, Cooper RA, Robertson RN, Shimada SD. Three-dimensional push rim forces during two speeds of wheelchair propulsion. Am J Phys Med Rehabil. 1997;76(5): 420–26.
24. Robertson RN, Boninger ML, Cooper RA, Shimada SD. Push rim forces and joint kinetics during wheelchair propulsion. Arch Phys Med Rehabil. 1996;77(9):856–64.
25. Praagman M, Stokdijk M, Veeger HEJ, Visser B. Predicting mechanical load of the glenohumeral joint using net joint moments. Clin Biomech. 2000;15:315–21.

Submitted for publication October 21, 2002. Accepted in revised form August 4, 2003.