Journal of Rehabilitation Research and Development
Vol. 40, No. 1, January/February 2003
Pages 27–38

Upper-limb fatigue-related joint power shifts in experienced wheelchair users and nonwheelchair users

Mary M. Rodgers, PhD, PT; Kevin J. McQuade, PhD, PT; Elizabeth K. Rasch, MS, PT; Randall E. Keyser, PhD; Margaret A. Finley, MA, PT

University of Maryland School of Medicine, Department of Physical Therapy, Baltimore, MD; Veterans Administration Medical Center, Baltimore, MD

Abstract—This paper evaluates power transfer or shifting across upper-limb segments, resulting from fatigue-inducing wheelchair propulsion. Nineteen manual wheelchair users (WCUs) and ten nonwheelchair users (NUs) participated in this study. Subjects propelled an instrumented wheelchair ergometer at a workload corresponding to 75% of the peak oxygen uptake attained during a maximal-graded exercise tolerance test. Subjects were required to propel the wheelchair for as long as they could at a constant velocity of 3 km/h (32 rpm). The test was terminated when subjects could no longer maintain the target velocity. Peak Performance video-capture system was used to determine upper-limb kinematics. Handrim forces and joint kinematics were used to calculate joint moments and power with the use of an inverse dynamics approach. Results showed that with fatigue, joint power shifts from the shoulder joint to the elbow and wrist joints. Implications for joint injury and propulsion efficiency are discussed.

Key words: biomechanics, fatigue, wheelchair.

INTRODUCTION

Between 1.4 and 1.6 million people (approximately 0.5% of the noninstitutionalized U.S. civilian population) use wheelchairs for locomotion [1,2]. Wheelchair use is increasing even when accounting for increases in population and age. Describing the biomechanics of wheelchair propulsion is a crucial first step to understanding how to prevent and treat overuse injuries that may result from long-term wheelchair propulsion. Toward this aim, biomechanical strategies and electromyographic activity of wheelchair propulsion have been extensively characterized in manual wheelchair users (WCUs) [3–15]. Experimental paradigms that perturb the conditions of wheelchair propulsion or compare subjects with differing attributes have been employed as approaches to recognizing potential pathomechanical adaptations [5,12,15–24].

A fatigue protocol is one model used to study the potential risks for overuse injuries. Rodgers et al. have shown that several biomechanical characteristics of manual wheelchair propulsion change with fatigue [25,26]. Comparisons of nonwheelchair users (NUs) to experienced WCUs have also been used [7,23]. Experienced WCUs appear to minimize peak handrim forces, prolong handrim impulse force, and prolong the duration of wheel contact. These responses may protect upper-limb joints from the trauma of repetitive forceful loading.

Abbreviations: GXT= graded exercise test, NU = nonwheelchair user, WCU = wheelchair user.

This material was based on work supported in part by the Department of Veterans Affairs (VA), Rehabilitation Research and Development Service, merit review grant B92-465A.

Address all correspondence and requests for reprints to Mary M. Rodgers, University of Maryland, School of Medicine, Department of Physical Therapy, 100 Penn Street, Baltimore, MD 21201; 410-706-5658; fax: 410-706-4903; mrodgers@umaryland.edu.

Calculation of joint power permits the estimation of upper-limb kinetic demands during wheelchair propulsion [25,27]. Data obtained empirically and from biomechanical modeling have shown that the shoulder is the primary source of joint power during wheelchair propulsion [25,11]. However, power shifts between upper-limb joints during various conditions of propulsion and with differing subjects have not been examined. This study evaluated power transfer across upper-limb segments because of fatigue-inducing wheelchair propulsion. We hypothesized that experienced WCUs will compensate for fatigue differently compared to NUs. Specifically, we hypothesized that fatigue causes the distribution of upper-limb power to be altered.

METHODS

Subjects

Nineteen manual WCUs and ten NUs participated in this study (**Table 1**). Of the WCUs, 3 were female and 16 were male. The majority (16) had spinal cord injuries. Age ranged from 21 to 68 yr and wheelchair use ranged from 3 to 38 yr. Of the NUs, three were female and seven were male. Age ranged from 23 to 47 yr. Prior to testing, the participants were given a medical examination by a physician familiar with the requirements for participation. Wheelchair users who used a manual wheelchair for at least 1 year before the study for the majority of home and community mobility were included in the study. Nonwheelchair users had no previous wheelchair use experience. Subjects in both groups were excluded if they had upper-limb disorders, ventilatory involvement, or systemic diseases or were taking medications that would alter exercise performance. Prior to testing, participants gave written, informed consent in accordance with the procedures approved by the Institutional Review Board.

Instrumentation

We conducted all exercise tests on a prototypical wheelchair ergometer (**Figure 1**) with a 28 cm diameter handrim and no camber in the wheels. Components of a stationary bicycle ergometer were used to provide frictional resistance to propulsion through a chain and sprocket system connected to the wheelchair axle at one end and to a flywheel at the other end. A nylon belt was used to create a pulley system to which known weights could be applied for precise control of resistance. The wheelchair ergometer was instrumented with PY6-4 six-component force/torque transducers (Bertec Corp., Worthington, Ohio) in its wheel hubs. The transducers used bonded strain gauges to measure handrim forces and moments in three dimensions (six channels). The rotating coordinate systems were transformed to the fixed laboratory frame with $+y$ vertical, $+x$ forward, and $+z$ lateral. The amplified electrical signals from the strain gauges and potentiometer were collected with an analog-to-digital converter and acquisition software (Peak Performance Technologies, Colorado Springs, Colorado) at 360 Hz. A bicycle speedometer with a digital display was attached to the right wheel of the chair and placed in view of the participant to provide visual feedback of propulsion velocity.

Shoulder, elbow, wrist, and trunk motions were measured with three Peak 3-D (three-dimensional) charge-coupled device cameras and a video acquisition system (Peak Performance Technologies, Colorado Springs, Colorado) at 60 frames/s. One camera was located 25 ft directly in front of the wheelchair ergometer, and the other two cameras were approximately 45° to each side of the center camera. Reflective markers were used to measure trunk, shoulder, elbow and wrist flexion/extension, shoulder abduction/adduction, and wrist radial/ulnar deviation (**Figure 1**). Markers were placed at the proximal shoulder, radial head, radial and ulnar styloids of the wrist, base of the fifth metacarpal of the hand, and at the hip joint. Joint marker linear displacements were measured and differentiated to obtain movement kinematics.

Model

We calculated joint forces, moments, and powers with a 3-D linked segment model [27]. Inputs to the model included 3-D angular displacement, velocity, acceleration and linear acceleration, handrim kinetic, and temporal and anthropometric data. This model used an inverse dynamics approach, employed the Newton-Euler method based on body coordinate systems, and assumed the arm to be three rigid segments (hand, forearm, and upper arm) connected by the wrist, elbow, and shoulder joints.

Displacement data collected during the motion analysis were differentiated to form velocity and acceleration vectors. Each marker's linear acceleration was used to interpolate the linear acceleration of the center of mass of each limb, which was transformed into respective body

Table 1.
Subject characteristics.

Gender	Age (yr)	Height (cm)	Weight (kg)	Arm Length (cm)	Trunk Length (cm)	Injury	Time in Wheelchair (yr)
Wheelchair Users							
M	21	180.5	72.4	47.0	47.0	SCI-T5	7
M	30	185.4	66.4	77.0	51.0	SCI-T3	12
M	32	157.5	56.8	73.0	51.0	Spinabifida	6
M	33	188.0	97.3	80.0	57.0	Bilateral PN	10
M	36	154.9	102.3	66.0	43.2	SCI-T12	10
F	37	121.9	52.3	60.0	32.0	SCI-T7	34
F	39	177.8	75.0	70.5	40.0	Multitrauma	12
M	40	167.5	100.0	73.0	47.0	SCI-T2	10
M	42	179.1	95.5	82.0	45.0	SCI-T12	11
M	44	178.8	71.2	68.5	46.5	SCI-T3	24
F	45	162.5	44.9	43.0	43.0	SCI-T7	24
M	49	187.0	127.0	50.5	48.0	SCI-L5	3
M	50	186.0	90.9	45.0	48.5	SCI-T5	28
M	52	180.3	66.8	74.0	42.0	SCI-T6	12
M	52	180.3	78.5	75.0	49.0	SCI-T5	26
M	54	182.9	77.3	73.0	51.0	Multitrauma	26
M	54	180.3	72.7	71.1	45.7	SCI-T5	5
M	58	178.0	76.3	46.0	42.0	SCI-T8	38
M	68	187.5	79.5	45.0	51.0	SCI-T4	21
Mean	44.0	174.5	79.1	64.2	46.3	—	16.8
SD	11.3	16.1	19.6	13.5	5.4	—	10.4
Nonwheelchair Users							
F	23	167.6	86.9	70.5	42.9	—	—
M	24	180.3	72.4	75.0	45.0	—	—
F	27	157.5	50.7	64.8	45.7	—	—
M	27	177.8	74.7	75.6	40.0	—	—
M	28	180.3	78.9	73.0	48.0	—	—
M	30	170.2	73.3	70.6	49.2	—	—
M	32	195.6	96.8	81.5	47.0	—	—
F	33	177.8	70.1	74.4	44.5	—	—
M	34	185.4	102.3	74.0	45.0	—	—
M	47	182.9	88.2	71.0	43.0	—	—
Mean	30.5	177.5	79.4	73.0	45.0	—	—
SD	6.9	10.5	14.8	4.3	2.7	—	—

PN = peripheral neuropathy, SCI = spinal cord injury, SD = standard deviation

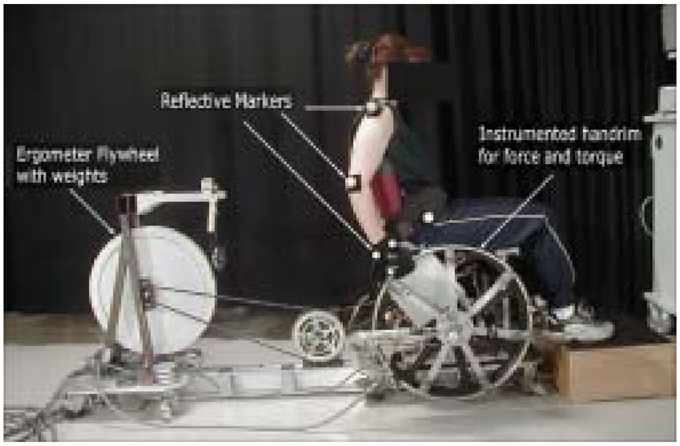

Figure 1.
Experimental setup for data collection.

coordinates. Contact forces and moments in 3-D were measured from the hub transducer, transformed from handrim to hand coordinates, and used as input to the hand equations. Actual moment applied by the hand to the wheelchair rim, using the fifth metacarpophalangeal joint as the location of the point of force application [28], was obtained by

$$M_T = M_m - r_w F,$$

where F is the measured contact force from hand to handrim, r_w is the radius of the handrim of the wheelchair, M_m is the measured moment from hand to handrim of the wheelchair, and M_T is the actual moment from hand to handrim of the wheelchair. Joint centers for the wrist, elbow, and shoulder joints were located relative to the markers at each respective location.

The inverse dynamics approach was used by the model to calculate joint kinetics for wrist flexion/extension, shoulder flexion/extension, elbow flexion/extension, wrist radial/ulnar deviation, and shoulder abduction/adduction. Joint power was calculated as the product of the net joint moment and angular velocity for each joint.

Test Protocol

WCUs and NUs completed a maximal graded exercise test (GXT) and a submaximal endurance test on the wheelchair ergometer. Seat width and backrest height of the ergometer were matched to the WCUs' own wheelchairs. Leg position approximated that of the WCUs seated in their own wheelchairs through the use of step stools of varying heights. For the NUs, the ergometer was adjusted to a seat width and backrest height that was comfortable. We achieved leg position with the thighs parallel to the floor by adjusting step stool height.

To acclimate to wheelchair propulsion, each NU had four 20 min practice sessions using the ergometer with minimal resistance. Practice sessions and propulsion tests were at least 24 h apart. The maximal GXT was used to establish resistance load for the fatigue test. For this test, subjects rested (6 min), propelled the wheelchair at a velocity of 3 km/h (32 rpm) without a load (3 min), then continued propelling while weight was incrementally added at a rate of 0.3 kg every 3 min. The test was terminated at volitional exhaustion, defined as the self-reported inability to maintain the target velocity. Subjects were monitored and encouraged to maintain the target velocity.

Two to seven days after completing the GXT, subjects completed an endurance (fatigue) test on the wheelchair ergometer. For the fatigue test, subjects propelled the wheelchair at 3 km/h without a load (3 min) and then continued propelling at the submaximal load until volitional exhaustion. The submaximal load corresponded to 75 percent of the peak oxygen uptake attained during the subject's GXT. Participants were encouraged to continue to exercise for as long as possible. Propulsion mechanics, including handrim kinetics, joint kinematics, and temporal characteristics, were collected for 6 s (three propulsion cycles) during the last 30 s of wheeling without a load, after 2.5 min of wheeling with a load (fresh), and just before cessation of the exercise (fatigued).

Data Analysis

We averaged kinetic, kinematic, and temporal data over three cycles (contact to contact) for two conditions (fresh and fatigued). Temporal variables included contact time (second and % cycle) and stroke frequency (cycles per second). Propulsion kinetics were characterized by peak handrim forces and moments during wheel contact and peak joint forces, moments, and powers. Joint position at the moment of wheel contact, maximal joint position and range of motion during wheel contact, and joint position at the moment of wheel release characterized upper-limb and trunk kinematics. We determined differences between peak shoulder, elbow, and wrist power occurring during wheel contact (positive work phase) to identify power shifts with fatigue.

Joint kinetics and kinematics, handrim kinetics, and propulsion temporal characteristics were compared between groups (WCU, NU) and during the fresh and fatigued states. Data were assessed for significant intragroup (first effect) and intergroup (second effect) differences with the use of two-way analyses of variance (ANOVA). The first independent variable was state, and the respective first effect variable levels were fresh versus fatigued states. A repeated measures procedure was used in this first effect. The second independent variable was group, and the respective second effect levels were WCUs and NUs. State-by-group interactions were also assessed by the statistical model. By using two-way ANOVAs for all comparisons, we maintained the experimentwise alpha level for type-I hypothetical error at the calculated probability of occurrence for each F-ratio. Therefore, a significance level of $p < 0.05$ was maintained for all comparisons. Scores were reported as means and standard deviations (SDs).

RESULTS

As a result of fatigue, stroke frequency (p = 0.02) and contact time as a percent of the propulsion cycle (p = 0.02) increased for all subjects (**Table 2**). There were no handrim kinetic changes with fatigue (**Table 3**). As shown in **Tables 4** and **5**, wrist radial (anterior) shear force (p = 0.03) and elbow extension moment (p = 0.04) decreased in both groups with fatigue. There were no changes in shoulder forces or moments because of fatigue (**Table 6**). Maximal trunk flexion angle during contact also increased by 5° ± 17° with fatigue in both groups (p = 0.02). No significant interactions of group and state were observed for the kinetic or kinematic data.

Significant group (NUs versus WCUs) main effects were found in the following temporal parameters. Stroke frequency was significantly higher (p = 0.005) and contact time significantly shorter (p = 0.009) for the WCUs (**Table 2**). Some significant kinematic differences were also found between the groups during wheel contact regardless of fatigue state. WCUs had greater maximum elbow flexion (WCUs 102° ± 7°, NUs 111° ± 10°, p = 0.005) than the NUs. NUs had greater maximum trunk flexion/extension range of motion (WCUs 10° ± 5°, NUs 33° ± 21°, p = 0.001), maximum shoulder adduction (WCUs 24° ± 11°, NUs 17° ± 5°, p = 0.025), and shoulder flexion/extension range of motion (WCUs 63° ± 16°, NUs 75° ± 11°, p = 0.024) than the WCUs.

Several kinetic differences between groups were also significant. Handrim downward (p = 0.03) and lateral (p = 0.04) forces were lower for the WCUs (**Table 3**). Elbow peak posterior shear force (p = 0.00l3) and lateral shear

Table 2.
Temporal characteristics.

State	Group: WCU	Group: NU	State Means ± SD
Stroke Frequency (cycles/s)			
Fresh	1.2 ± 0.32	1.0 ± 0.43	1.1 ± 0.38*
Fatigued	1.3 ± 0.32	1.1 ± 0.43	1.2 ± 0.38*
Group Means ± SD	1.2 ± 0.32†	1.0 ± 0.43†	—
Contact Time (s)			
Fresh	0.45 ± 0.16	0.54 ± 0.21	0.49 ± 0.21
Fatigued	0.45 ± 0.16	0.56 ± 0.21	0.50 ± 0.21
Group Means ± SD	0.45 ± 0.16†	0.55 ± 0.21†	—
Contact Time (% cycle)			
Fresh	40.7 ± 12.4	37.9 ± 17.8	39.3 ± 47.4*
Fatigued	43.7 ± 12.4	42.8 ± 17.8	43.5 ± 34.5*
Group Means ± SD	42.1 ± 43.6	41.0 ± 38.2	—

*Significant main effect for state, fresh versus fatigued ($p < 0.05$)
†Significant main effect for group, WCU versus NU ($p < 0.05$)
WCU = wheelchair user
NU = nonwheelchair user
Note: All data are means ± standard deviations (SDs).

Table 3.
Handrim kinetics.

State	Group		State Means ± SD
	WCU	NU	
Peak Forward Handrim Force (N)			
Fresh	68 ± 24	66 ± 18	68 ± 21
Fatigued	67 ± 21	60 ± 22	65 ± 21
Group Means ± SD	68 ± 23	63 ± 20	—
Peak Downward Handrim Force (N)			
Fresh	66 ± 19	102 ± 47	78 ± 33
Fatigued	75 ± 30	94 ± 47	81 ± 39
Group Means ± SD	70 ± 25*	98 ± 47*	—
Peak Lateral Handrim Force (N)			
Fresh	9 ± 10	23 ± 20	14 ± 15
Fatigued	14 ± 12	20 ± 17	16 ± 14
Group Means ± SD	11 ± 11*	22 ± 18*	—
Peak Propulsive (forward) Handrim Moment (N•m)			
Fresh	21 ± 7	27 ± 9	23 ± 8
Fatigued	21 ± 6	24 ± 9	22 ± 7
Group Means ± SD	21 ± 6	26 ± 9	—

*Significant main effect for group, WCU versus NU ($p < 0.05$)
WCU = wheelchair user
NU = nonwheelchair user
Note: All data are means ± standard deviations (SDs).

force (p = 0.005) were less in the WCUs (**Table 5**). As shown in **Table 6**, shoulder peak compression force (p = 0.02), lateral shear force (p = 0.004), and adduction moment (p = 0.05) were less in the WCUs. Net joint power outputs were not statistically different between the groups.

The main effect of fatigue (for both groups) is shown in the change in power distribution (**Figure 2(a)** and **(b)**). **Tables 4** to **6** show the absolute power data for the joints. For both groups, shoulder flexion power decreased significantly (p = 0.01) with fatigue from 61 percent of the total power exerted in the upper limb to 50 percent after fatigue. The elbow extension and wrist ulnar deviation percent of total upper-limb power increased by 4 and 7 percent, respectively with fatigue. The power shifts were significant from the fresh to fatigued states between the shoulder and elbow (fresh 86 W ± 78 W, fatigued 51 W ± 69 W, p = 0.005) and between the shoulder and wrist (fresh 68 W ± 91 W, fatigued 33 W ± 88 W, p = 0.04).

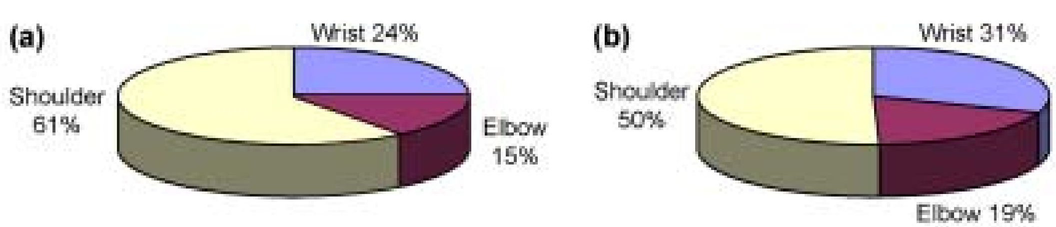

Figure 2.
(a) Upper-limb power distribution during nonfatigued propulsion for all subjects (n = 29). **(b)** Upper-limb power distribution during fatigued propulsion for all subjects (n = 29).

Table 4.
Wrist joint kinetics.

State	Group WCU	Group NU	State Means ± SD
Peak Anterior Shear Force (N)			
Fresh	45 ± 32	49 ± 38	46 ± 21*
Fatigued	37 ± 27	42 ± 38	39 ± 21*
Group Means ± SD	41 ± 21	46 ± 21	—
Peak Compression Force (N)			
Fresh	75 ± 32	103 ± 81	85 ± 38
Fatigued	82 ± 48	92 ± 86	86 ± 43
Group Means ± SD	79 ± 32	98 ± 48	—
Peak Lateral Shear Force (N)			
Fresh	24 ± 27	38 ± 48	29 ± 21
Fatigued	25 ± 32	38 ± 32	30 ± 21
Group Means ± SD	25 ± 21	38 ± 21	—
Peak Flexion Moment (N•m)			
Fresh	5 ± 5	11 ± 16	7 ± 5
Fatigued	8 ± 11	9 ± 16	8 ± 5
Group Means ± SD	6 ± 5	10 ± 5	—
Peak Ulnar Deviation Moment (N•m)			
Fresh	40 ± 16	44 ± 21	41 ± 11
Fatigued	38 ± 16	38 ± 21	38 ± 11
Group Means ± SD	39 ± 11	41 ± 11	—
Peak Ulnar Deviation Power (N•m)			
Fresh	33 ± 37	69 ± 78	46 ± 57
Fatigued	45 ± 43	54 ± 47	49 ± 45
Group Means ± SD	40 ± 40	6 ± 62	—

*Significant main effect for state, fresh versus fatigued ($p < 0.05$)
WCU = wheelchair user
NU = nonwheelchair user
Note: All data are means ± standard deviations (SDs).

DISCUSSION

As the population of WCUs increases, it is important for society as a whole to address the challenges this group faces in maintaining their functional independence. If optimal health can be maintained for these individuals, then the spiraling health costs and detrimental individual effects on WCUs associated with secondary conditions can be contained. Scientists studying the biomechanics of wheelchair propulsion are faced with the dilemma of linking their results to understanding how to prevent and treat overuse injuries that may result from long-term wheelchair propulsion. The use of the fatigue condition is an attempt to provide a perturbation that mimics conditions present with overuse.

This study provided data to evaluate how the transfer of power across upper-limb segments changes during fatigue-inducing wheelchair propulsion. We hypothesized that experienced WCUs compensate for fatigue differently than NUs. We found group differences in both states for the temporal and handrim kinetic parameters of—

1. Stroke frequency; higher for WCUs.
2. Contact time; shorter for WCUs.

Table 5.
Elbow joint kinetics.

State	Group		State Means ± SD
	WCU	NU	
Peak Posterior Shear Force (N)			
Fresh	4 ± 5	26 ± 48	11 ± 16
Fatigued	7 ± 11	24 ± 54	13 ± 21
Group Means ± SD	5 ± 5*	25 ± 32*	—
Peak Compression Force (N)			
Fresh	62 ± 38	80 ± 65	68 ± 32
Fatigued	62 ± 48	67 ± 70	64 ± 38
Group Means ± SD	62 ± 32	74 ± 38	—
Peak Lateral Shear Force (N)			
Fresh	34 ± 32	72 ± 48	47 ± 27
Fatigued	40 ± 48	70 ± 54	50 ± 32
Group Means ± SD	37 ± 32*	71 ± 32*	—
Peak Extension Moment (N•m)			
Fresh	44 ± 21	40 ± 21	42 ± 16†
Fatigued	39 ± 16	35 ± 21	38 ± 11†
Group Means ± SD	41 ± 16	37 ± 11	—
Peak Extension Power (W)			
Fresh	28 ± 25	28 ± 22	28 ± 24
Fatigued	32 ± 24	28 ± 19	30 ± 21
Group Means ± SD	30 ± 25	28 ± 21	—

*Significant main effect for group, WCU versus NU ($p < 0.05$)
†Significant main effect for state, fresh versus fatigued ($p < 0.05$)
WCU = wheelchair user
NU = nonwheelchair user
Note: All data are means ± standard deviations (SDs).

3. Handrim downward and lateral forces; lower for WCUs.
4. Shoulder peak compression force, lateral shear force, and adduction moment; less in WCUs.
5. Elbow peak posterior and lateral shear forces; less in the WCUs.

These pattern differences demonstrated the more rapid, shorter strokes and the diminished propulsive handrim forces of the WCUs compared to the NUs. However, these differences were present regardless of whether the subjects were fresh or fatigued.

The first hypothesis was found to be true regardless of fatigue state for the kinematic measures:

1. Maximum elbow flexion; WCUs had more.
2. Maximum trunk flexion/extension range of motion; NUs had more.
3. Maximum shoulder adduction; NUs had more.
4. Shoulder flexion/extension range of motion; NUs had more.

These differences demonstrated that the inexperienced able-bodied subjects used their trunk and upper limbs through greater ranges. Intergroup differences in control of the trunk musculature may have been an important contributor to their differing kinematic patterns.

We also hypothesized that fatigue causes the distribution of upper-limb power to be altered. We found this hypothesis to be true for both experienced and inexperienced WCUs. The results of this analysis suggested that as WCUs and NUs became fatigued, the proportion of the total power generated at the wrist increased with respect to the power generated at the shoulder. The increase in this relationship did not account for the total decrease at

Table 6.
Shoulder joint kinetics.

State	Group		State Means ± SD
	WCU	NU	
Peak Anterior Shear Force (N)			
Fresh	64 ± 32	63 ± 48	64 ± 27
Fatigued	63 ± 38	58 ± 54	61 ± 32
Group Means ± SD	63 ± 27	61 ± 32	—
Peak Compression Force (N)			
Fresh	12 ± 21	43 ± 70	22 ± 27
Fatigued	20 ± 32	35 ± 65	25 ± 32
Group Means ± SD	16 ± 21*	39 ± 38*	—
Peak Lateral Shear Force (N)			
Fresh	24 ± 27	61 ± 48	37 ± 27
Fatigued	31 ± 43	57 ± 59	40 ± 32
Group Means ± SD	28 ± 27*	59 ± 32*	—
Peak Adduction Moment (N•m)			
Fresh	31 ± 38	54 ± 38	39 ± 27
Fatigued	32 ± 43	54 ± 38	40 ± 27
Group Means ± SD	32 ± 32*	54 ± 21*	—
Peak Flexion Moment (N•m)			
Fresh	65 ± 27	63 ± 32	64 ± 21
Fatigued	60 ± 32	56 ± 38	59 ± 21
Group Means ± SD	63 ± 21	59 ± 21	—
Peak Flexion Power (W)			
Fresh	114 ± 83	112 ± 71	113 ± 79†
Fatigued	85 ± 68	74 ± 66	81 ± 67†
Group Means ± SD	100 ± 75	59 ± 21	—

*Significant main effect for group, WCU versus NU ($p < 0.05$)
†Significant main effect for state, fresh versus fatigued ($p < 0.05$)
WCU = wheelchair user
NU = nonwheelchair user
Note: All data are means ± standard deviations (SDs).

the shoulder. Reasons for the unexplained power difference may be related to increased movement at the trunk in the fatigued state. Even though the proportion of total power shift that was attributed to the change in the relationship between the shoulder and the wrist was small, the finding may still be of biological consequence because the wrist joint is a small, relatively unstable joint and the muscles that surround it are of a small mass.

The NUs' mean trunk excursion was over three times that of the WCUs'. Although trunk kinetic data were not available (a limitation of the model used), one could speculate that by using more trunk motion, NUs generated power from the trunk, which contributed to the total power output that was not available to WCUs because of their disabilities (reflected in **Table 1**). Since most WCUs may not be able to use trunk muscles to help generate propulsion power needed as fatigue occurs, they may have an increased susceptibility to injury at the wrist as the shoulder muscles fatigue.

Joint power generated during gait and cycling has been assessed, but no investigators have looked specifically at joint power transfer during wheelchair propulsion [29,30]. Van Ingen et al found that during cycling, the total power generated (calculated as the product of pedal moment and pedal crank angular velocity) was similar to the sum of joint power calculated from the hip, knee, and ankle [30]. An analogous comparison of the wheelchair wheel to the cycle crank, using the product of the peak wheel rotation moment and wheel angular velocity, showed that the sum of the joint powers was as much as double the power generated at the wheel. Unlike the results of Van Ingen for the bicycle crank, our results suggest that significant intersegmental power was dissipated or inefficiently transferred to the wheel.

A number of issues are relevant to the interpretation of data from this study. The experimental setup incorporated a prototypical stationary wheelchair ergometer. Although width and seat height were adjusted to the subject's own wheelchair, no camber was present. Although absence of camber may have limited the similarity to the user's own wheelchair, this setup ensured consistency and experimental control for data collection. The model used for this study included a number of assumptions and limitations. One of the assumptions was the location of the point of application of force to the handrim at the fifth metacarpophalangeal joint. As Sabick and associates have shown, this assumption resulted in no more error than other methods that have been used [28]. Another limitation of the model was the lack of trunk kinetic calculations. Although the sample size was comparable to other work in the field, it is possible that some propulsion styles and anthropometrics may not be reflected in this sample.

CONCLUSION

In summary, our results demonstrated joint power shifts from the shoulder joint to the elbow and wrist joints with fatigue in both WCUs and NUs. We found that for some kinematic measures, experienced WCUs compensated for fatigue differently than NUs. Although power shifting may be a necessary strategy to maintain a constant work output with fatigue, it may increase the risk of injury to upper-limb joints. The mechanism underlying the direct association between joint power shifting and risk of injury remains unknown and warrants further investigation.

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. Russell JN, Hendershot GE, LeClere F. National Center for Health Statistics, Trends and differential use of assistive technology devices: United States, 1994. Vital Health Stat Adv Data 1997:292.
3. Shimada SD, Robertson RN, Bonninger ML, Cooper RA. Kinematic characterization of wheelchair propulsion. J Rehabil Res Dev 1998;35(2):210–18.
4. 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.
5. 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.
6. Dallmeijer AJ, van der Woude LHV, Veeger HEJ, Hollander AP. Effectiveness of force application in manual wheelchair propulsion in persons with spinal cord injuries. Am J Phys Med Rehabil 1998;77:213–21.
7. Robertson RN, Boninger ML, Cooper RA, Shimada SD. Pushrim forces and joint kinetics during wheelchair propulsion. Arch Phys Med Rehabil 1996;77:856–64.
8. Robertson RN, Cooper RA. Kinetic characteristics of wheelchair propulsion utilizing the SMART^{Wheel}. American Society of Biomechanics; 1993 Oct 21–23; University of Iowa; Iowa City, Iowa.
9. Tanaka O. The analysis of wheelchair driving motion: Torque to handrim and pattern of muscle activities in upper extemity [translated from Japanese]. Sogo Rehabil 1982;10:251–57.
10. Theisen D, Francaux M, Fayt A, Sturbois X. A new procedure to determine external power output during handrim wheelchair propulsion on a roller ergometer: A reliability study. Int J Sports Med 1996;17:564–71.
11. van der Helm FCT, Veeger HEJ. Quasi-static analysis of muscle forces in the shoulder mechanism during wheelchair propulsion. J Biomech 1996;29(1):39–52.
12. van der Woude LHV, 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.
13. Kulig K, Rao SS, Mulroy SJ, Newsam CJ, Gronley JK, Bontrager EL, et al. Shoulder joint kinetics during the push phase of wheelchair propulsion. Clin Orthop 1998; 354:132–43.
14. Mulroy SJ, Gronley JK, Newsam CJ, Perry J. Electromyographic activity of shoulder muscles during wheelchair propulsion by paraplegic persons. Arch Phys Med Rehabil 1996;77:187–93.

15. Boninger ML, Cooper RA, Robertson RN, Shimada SD. Three-dimentional pushrim forces during two speeds of wheelchair propulsion. Am J Phys Med Rehabil 1997; 76:420–26.
16. Ruggles DL, Cahalan T, An K-N. Biomechanics of wheelchair propulsion by able-bodied subjects. Arch Phys Med Rehabil 1994;75:540–44.
17. Brown DD, Knowlton RG, Hamil J, Schneider TL, Hetzler RK. Physiological and biomechanical differences between wheelchair-dependent and able-bodied subjects during wheelchair ergometry. Eur J Appl Physiol Occ Physiol 1990;60:179–82.
18. Rudins A, Laskowski ER, Growney ES, Cahalan TD, An K-N. Kinematics of the elbow during wheelchair propulsion: A comparison of two wheelchairs and two stroking techniques. Arch Phys Med Rehabil 1997;78:1204–10.
19. van der Woude LHV, Hendrich KMM, Veeger HEJ, van Ingen Schenau GJ, Rozendal RH, de Groot G, et al. Manual wheelchair propulsion: Effects of power output on physiology and technique. Med Sci Sports Exerc 1988; 20(1):70–78.
20. Veeger HEJ, van der Woude LHV, Rozendal RH. Wheelchair propulsion technique at different speeds. Scand J Rehabil Med 1989;21:197–203.
21. Vanlandewijck YC, Spaepen AJ, Lysens RJ. Wheelchair propulsion efficiency: Movement pattern adaptations to speed changes. Med Sci Sports Exerc 1994;26:1373–81.
22. Bednarczyk JH, Sanderson DJ. Kinematics of wheelchair propulsion in adults and children with spinal cord injury. Arch Phys Med Rehabil 1994;75:1327–34.
23. Veeger HEJ, Lute EMC, Roeleveld K, van der Woude LHV. Differences in performance between trained and untrained subjects during a 30-s sprint test in a wheelchair ergometer. Eur J Appl Physiol Occ Physiol 1992; 64:158–64.
24. Dallmeijer AJ, Kappe YJ, Veeger HEJ, Janssen TWJ, van der Woude LHV. Anaerobic power output and propulsion technique in spinal cord injured subjects during wheelchair ergometry. J Rehabil Res Dev 1994;31(2):120–28.
25. Rodgers MM, Gayle GW, Figoni SF, Kobayashi M, Lieh J, Glaser RM. Biomechanics of wheelchair propulsion during fatigue. Arch Phys Med Rehabil 1994;75:85–93.
26. Rodgers MM, Keyser RE, Gardner ER, Russell PJ, Gorman PH. Influence of trunk flexion on biomechanics of wheelchair propulsion. J Rehabil Res Dev 2000;37(3): 283–95.
27. Rodgers MM, Tummarakota S, Lieh J. Three-dimensional dynamic analysis of wheelchair propulsion. J Appl Biomech 1998;14:80–92.
28. Sabick MB, Zhao KD, An K-N. A comparison of methods to compute the point of force application in handrim wheelchair propulsion: A technical note. J Rehabil Res Dev 2001 Jan–Feb;38(1):57–68.
29. Winter DA. Moments of force and mechanical power in jogging. J Biomech 1983;16:91–97.
30. Van Ingen S, Van Woensel W, Boots P, Snackers R, De Groot G. Determination and interpretation of mechanical power in human movement: Application to ergometer cycling. Eur J Appl Physiol Occ Physiol 1990;61:11–19.

Submitted for publication May 2, 2001. Accepted in revised form October 7, 2002.