

## Reliability of biomechanical variables during wheelchair ergometry testing

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**Abstract**—Wheelchair ergometer testing is used to characterize wheelchair propulsion mechanics. The reliability of kinematic and kinetic measures has not been investigated for wheelchair ergometer testing. In this study, test-retest reliability of biomechanical measurements on a wheelchair ergometer was determined during a submaximal endurance test. Ten nondisabled subjects (seven male, three female), inexperienced in wheelchair propulsion, completed three separate submaximal fatigue tests. An instrumented wheelchair ergometer was used to measure handrim kinetics while three-dimensional kinematic data were collected. Analysis of variance was used to determine if measurement differences existed across the tests. Intraclass correlation coefficients (ICC) were calculated to determine the reliability of the measurements. The majority of handrim and temporal variables were found to be reliable. Joint kinematic variables were less reliable, especially those involving wrist movements in the fatigued state. It was concluded that most biomechanical variables obtained during wheelchair ergometry were reliable.

**Key words:** *biomechanics, test-retest reliability, wheelchair propulsion.*

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### INTRODUCTION

In an effort to identify pathomechanics, researchers study wheelchair propulsion mechanics, using instrumented wheelchair ergometers of various designs (1–5). These devices simulate manual wheelchair propulsion and allow measurement of kinematic and kinetic features of the task. Kinematic patterns and temporal characteristics of wheelchair propulsion have been reported more frequently (1,2,6–10); however, only a limited number of studies have reported the reliability of these measures (2,11,12). Even fewer researchers include handrim kinetics in their studies of wheelchair propulsion. Various research laboratories employ a variety of systems and methods in conducting these experiments (3,4,11–14). No previous reports have included internal validity measures of handrim kinetics.

The purpose of this study was to determine the reliability of biomechanical measurements during repeated wheelchair ergometer exercise tests to fatigue. Nondisabled individuals, inexperienced in wheelchair propulsion, were chosen as subjects in an attempt to maximize potential test-retest variability.

## METHODS

### Subjects

Ten nonwheelchair users (seven male, three female; height =  $178 \pm 11$  cm; weight =  $79 \pm 15$  kg; age =  $31 \pm 7$  yr) served as study participants. Participants were excluded if they presented with upper body orthopedic disorders, systemic diseases that would contraindicate participation by limiting upper body exercise performance, or medication that would impede or enhance exercise performance. None of the participants had used a manual wheelchair for primary mobility in the past or had participated in an upper-body aerobic exercise-training program in the past year. Before participation, written informed consent was obtained in accordance with procedures approved by the Institutional Review Board.

### Instrumentation

A prototypical wheelchair ergometer described in detail elsewhere was used to simulate propulsion mechanics (**Figure**) (15). Handrim forces and torques were measured in three dimensions (3D) by PY6-4 six-component force/torque transducers with bonded



**Figure.** Wheelchair ergometer with subject, including marker placement. Axis coordinate system ( $x$ ,  $y$ ,  $z$ ) of the transducer embedded in the wheel hub is shown.

strain gauges (Bertec Corp., Worthington, OH) mounted in the wheel hubs of the wheelchair ergometer. Orientation of the force coordinates of the right transducer was tangential ( $F_x$ , forward+), radial ( $F_y$ , up+), and medial-lateral ( $F_z$ , out of the wheel+), respectively (**Figure**). The transducers had a maximum torque capacity of 150 Nm for the  $M_z$  component (largest moment during the task of propulsion) and maximum plane-of-wheel force ( $F_x$  and  $F_y$ ) capacity of 3500 N, corresponding to a gain setting of unity (1.0) and a full-scale output of  $\pm 10$  V.

Angular positions of the wheel, transducer, and handrim assembly were measured by a potentiometer. A 12-bit analog to digital converter and acquisition software (Peak Performance Technologies, Colorado Springs, CO) were used to collect the amplified electrical signals from the strain gauges and potentiometer. A sample rate of 360 Hz was used for collecting handrim kinetic, temporal, and potentiometer data. A digital display bicycle speedometer attached to the right wheel of the chair and placed in view of the participant provided information on propulsion velocity.

Three Peak 3D Charge Coupled Device (CCD) cameras and a video acquisition system (Peak Performance Technologies, Colorado Springs, CO) were used to measure upper extremity and trunk movement. Kinematic data were collected at 60 frames/s. Dynamic accuracy of this system was reported to be  $1^\circ$  at speeds up to  $300^\circ/\text{s}$  (16). The cameras were located at a distance of 25 ft from the wheelchair, with one camera directly in front of the wheelchair and the other two cameras placed approximately  $45^\circ$  to each side of the center camera. A 12-point reflective marker system (**Figure**) was used to measure trunk, shoulder, elbow, wrist flexion/extension, shoulder abduction/adduction, and wrist radial/ulnar deviations. Markers were placed on the greater trochanter, acromion process, lateral epicondyle, radial styloid, ulnar styloid and head of the fifth metacarpal, bilaterally. One researcher measured and differentiated joint marker linear displacements to obtain movement kinematics.

### Procedures

Procedures used in this investigation have been described in detail, elsewhere (17). Briefly, to become accustomed to wheelchair propulsion and to adopt a consistent propulsion style, subjects completed four 20-min sessions of wheelchair propulsion, with minimal resistance, before the exercise tests and at least 24 h apart. Subjects

then completed a peak exercise test by propelling the wheelchair at a speed of 3 km/h as weight was added incrementally to the carriage. While target velocity was maintained, 0.3-kg weights were added every 3 min. The addition of weight continued over time until the subjects were unable to maintain the required target velocity. Verbal encouragement was provided to maintain the required velocity. The peak test was performed to establish the resistive load for the subsequent fatigue tests.

Resistance for the fatigue tests corresponded to that which elicited 75 percent of the peak oxygen consumption ( $\text{VO}_2$ ) attained during the peak test. The peak test was performed 2 to 7 days before initiating fatigue tests. Three constant work rate fatigue tests were completed at least 48 h, but not more than 1 wk, apart. These fatigue tests were performed at a similar time of day, within a range of  $\pm 2$  h. Subjects completed a propulsion interval for 3 min without weight on the carriage, followed by application of the load. Subjects propelled the ergometer until they were unable to sustain the target velocity (3 km/h). Data were collected when the load was initially applied (fresh) and again when the subjects demonstrated that they could not longer maintain the designated velocity (fatigued state). The test ended following collection of these data. As in the peak test, subjects were verbally encouraged to maintain the required propulsion velocity.

### Statistical Analysis

Three propulsion cycles (contact-to-contact) were averaged to depict handrim kinetic, temporal, and joint kinematic data. Dependent variables in this study in the fresh and fatigued states included peak right handrim forces ( $F_x$ ,  $F_y$ ,  $F_z^+$ ,  $F_z^-$ ) and moments ( $M_x^+$ ,  $M_x^-$ ,  $M_y^+$ ,  $M_y^-$ ,  $M_z$ ), mathematically calculated resultant handrim

forces ( $F_{\text{res}}$ ) and moments ( $M_{\text{res}}$ ), stroke frequency (SF), contact time (in seconds and expressed as a percent of the contact cycle), peak angles throughout the entire contact cycle, and maximal joint angles measured at specific events of initial contact, during contact phase, and at release.

The independent variable was test number, identified as test 1, test 2, or test 3. Data were analyzed for significant differences between tests with the use of a repeated measures one-way analysis of variance (ANOVA). Type I error was accepted at  $p \leq 0.05$  for determination of a significant  $F$ -ratio. Tukey post hoc analysis was performed with significant  $F$ -ratios. The three tests were collapsed and fresh and fatigued states were compared again with the use of one-way repeated measures ANOVA. The mean squares for subjects ( $MS_s$ ) and within ( $MS_w$ ) for each dependent variable was used to calculate test-retest and fresh-fatigue intraclass correlation coefficients ( $R$ ) (18). Acceptable intraclass correlation was  $R \geq 0.60$ .

## RESULTS

The means and standard deviations, and reliability coefficients of temporal, kinetic, and kinematic variables are presented in **Tables 1–6**. Temporal variable correlation yielded strong  $R$  values, with no significant difference found in SF. However, differences were found in contact time, expressed in seconds and as a percent of the contact cycle, in the fresh state (**Table 1**). Significant differences were found for  $F_{\text{res}}$  in fatigue ( $p < 0.01$ ) (**Table 2**).  $R$  values were strong for all handrim variables except  $F_x$  ( $R = 0.59$ ). Joint kinematic results are shown in **Tables 3–6**. When expressed as peak angles that occurred throughout

**Table 1.**  
Temporal variables during fresh and fatigued states across tests.

Variable	State	Test 1 (mean $\pm$ SD)	Test 2 (mean $\pm$ SD)	Test 3 (mean $\pm$ SD)	ICC ( $R$ )	Grand Mean $\pm$ SD	( $R^2$ )*
Stroke frequency (cycles/s)	fresh	0.96 $\pm$ 0.14	1.00 $\pm$ 0.11	0.99 $\pm$ 0.11	0.82	0.99 $\pm$ 0.12	0.67
	fatigued	1.07 $\pm$ 0.20	0.99 $\pm$ 0.21	1.07 $\pm$ 0.22	0.84	1.04 $\pm$ 0.21	0.71
Contact time (s)	fresh <sup>†‡</sup>	0.55 $\pm$ 0.10	0.54 $\pm$ 0.10	0.62 $\pm$ 0.85	0.85	0.57 $\pm$ 0.10	0.72
	fatigued	0.56 $\pm$ 0.14	0.62 $\pm$ 0.17	0.55 $\pm$ 0.18	0.82	0.57 $\pm$ 0.16	0.67
Contact time (% cycle)	fresh <sup>‡</sup>	39.03 $\pm$ 7.57	39.84 $\pm$ 4.62	45.47 $\pm$ 4.95	0.66	41.45 $\pm$ 5.71	0.44
	fatigued	42.88 $\pm$ 6.17	43.30 $\pm$ 7.20	42.46 $\pm$ 9.76	0.88	42.88 $\pm$ 7.71	0.77

ICC = intraclass correlation <sup>†</sup>Significant difference between tests 2 and 3  
\*Coefficient of determination <sup>‡</sup>Significant difference between tests 1 and 3

**Table 2.**Handrim forces ( $N$ ) and moments ( $N \cdot m$ ) during fresh and fatigued states, across tests.

Variable	State	Test 1 (mean $\pm$ SD)	Test 2 (mean $\pm$ SD)	Test 3 (mean $\pm$ SD)	ICC ( $R$ )	Grand Mean $\pm$ SD	( $R^2$ )*
$F_{x+}$	fresh	64.7 $\pm$ 18.4	64.4 $\pm$ 29.1	59.1 $\pm$ 18.2	0.82	62.7 $\pm$ 21.9	0.67
	fatigued	54.9 $\pm$ 16.8	53.9 $\pm$ 20.4	44.5 $\pm$ 9.9	0.59	51.1 $\pm$ 15.7	0.35
$F_{y-}$	fresh	-109.0 $\pm$ 41.5	-101.6 $\pm$ 40.1	-90.7 $\pm$ 37.6	0.77	-100.5 $\pm$ 39.8	0.59
	fatigued	-97.9 $\pm$ 46.5	-111.7 $\pm$ 42.0	-85.8 $\pm$ 32.5	0.89	-98.5 $\pm$ 40.3	0.79
$F_{z-}$	fresh	-15.9 $\pm$ 17.4	-17.2 $\pm$ 18.6	-18.6 $\pm$ 15.6	0.86	-17.2 $\pm$ 17.2	0.74
	fatigued	-15.1 $\pm$ 16.0	-17.5 $\pm$ 21.3	-21.9 $\pm$ 16.3	0.81	-18.2 $\pm$ 17.8	0.66
$F_{z+}$	fresh	22.8 $\pm$ 19.5	19.0 $\pm$ 16.7	16.7 $\pm$ 14.8	0.87	19.5 $\pm$ 17.0	0.76
	fatigued	19.1 $\pm$ 16.1	23.8 $\pm$ 15.8	17.0 $\pm$ 13.1	0.91	20.0 $\pm$ 15.0	0.83
$M_{x+}$	fresh	2.6 $\pm$ 3.3	2.5 $\pm$ 2.9	2.9 $\pm$ 2.6	0.88	2.7 $\pm$ 2.9	0.77
	fatigued	2.7 $\pm$ 2.9	3.8 $\pm$ 2.8	2.3 $\pm$ 2.3	0.93	2.9 $\pm$ 2.7	0.86
$M_{x-}$	fresh	-2.6 $\pm$ 3.1	-2.7 $\pm$ 2.9	-3.1 $\pm$ 2.9	0.85	-2.8 $\pm$ 3.0	0.72
	fatigued	-3.3 $\pm$ 3.4	-2.6 $\pm$ 3.8	-3.7 $\pm$ 3.5	0.87	-3.2 $\pm$ 3.6	0.76
$M_{y+}$	fresh	2.1 $\pm$ 3.8	2.6 $\pm$ 4.1	3.0 $\pm$ 3.2	0.93	2.6 $\pm$ 3.7	0.86
	fatigued	2.3 $\pm$ 3.3	1.9 $\pm$ 3.6	2.6 $\pm$ 3.4	0.85	2.2 $\pm$ 3.4	0.72
$M_{y-}$	fresh	-2.9 $\pm$ 1.8	-2.8 $\pm$ 2.0	-3.1 $\pm$ 1.5	0.91	-2.9 $\pm$ 1.8	0.83
	fatigued	-2.6 $\pm$ 1.5	-3.6 $\pm$ 1.7	-2.8 $\pm$ 1.2	0.86	-3.0 $\pm$ 1.5	0.74
$M_z$	fresh	-27.4 $\pm$ 8.6	-26.4 $\pm$ 9.0	-25.2 $\pm$ 6.9	0.81	-26.4 $\pm$ 8.2	0.66
	fatigued	-24.0 $\pm$ 8.6	-26.0 $\pm$ 8.3	-22.2 $\pm$ 6.5	0.85	-24.1 $\pm$ 7.8	0.72
$F_{res}$	fresh†	132.5 $\pm$ 43.2	127.3 $\pm$ 42.6	114.3 $\pm$ 36.3	0.83	124.7 $\pm$ 40.7	0.69
	fatigued	118.5 $\pm$ 43.9	131.8 $\pm$ 39.6	103.3 $\pm$ 28.9	0.91	117.9 $\pm$ 37.6	0.83
$M_{res}$	fresh	28.5 $\pm$ 8.6	27.8 $\pm$ 8.1	26.4 $\pm$ 6.4	0.80	27.6 $\pm$ 7.7	0.64
	fatigued	25.3 $\pm$ 8.2	27.5 $\pm$ 7.9	23.7 $\pm$ 5.9	0.87	25.5 $\pm$ 7.3	0.76

ICC = Intraclass correlation

\*Coefficient of determination

†Significant difference between test 2 and test 3

the entire cycle (**Table 3**), the only significant difference between tests was found in wrist radial deviation in fresh and fatigue. The  $R$  values were weak for wrist extension (fresh, fatigue), wrist flexion (fatigue), wrist radial deviation (fatigue), shoulder flexion (fatigue), shoulder abduction (fatigue), shoulder adduction (fresh, fatigue), trunk flexion (fresh), and trunk extension (fatigue). When the joint kinematic variables were reported as angles measured at specific events, only wrist radial deviation during contact (fresh) was significant between tests (**Table 4**). Most  $R$  values indicating a strong correlation for the angles reported during the entire cycle were also significant for the specific events of contact and/or release, primarily in fresh. **Tables 5 and 6** display kinematic variables at contact and the angular displacement at the specific events of release, in each state as well as across tests.

Significant differences were found in shoulder abduction and trunk excursion (flexion to extension range), and

$F_x$  between the fresh and fatigued states. Good reliability was found for all temporal and kinetic variables and the majority of kinematic variables (**Table 7**).

## DISCUSSION

Wheelchair locomotion is studied in a manner similar to bipedal upright gait. Phases of stance in upright gait are equated to the contact time while the swing phase is synonymous with the release during wheelchair propulsion. Stance phase of gait comprises 60 percent of the entire cycle, and the remaining 40 percent is attributed to swing phase (19). Just as with upright gait, (20,21) the influence of propulsion speed on temporal variables has been reported for wheelchair propulsion (3). Boninger et al. (5) reported that stroke time, push time, peak force

**Table 3.**

Peak angular displacement of entire cycle (contact-to-contact), fresh and fatigued, across tests.

Variable (°)	State	Test 1 (mean ± SD)	Test 2 (mean ± SD)	Test 3 (mean ± SD)	ICC (R)	Grand Mean ± SD	(R <sup>2</sup> )*
Wrist extension	fresh	-35.7 ± 8.3	-34.0 ± 9.4	-36.8 ± 11.1	0.39	-35.5 ± 9.6	0.15
	fatigued	-29.6 ± 13.0	-35.7 ± 16.5	-40.8 ± 10.8	0.03	-35.4 ± 13.4	0.001
Wrist flexion	fresh	30.8 ± 13.5	26.4 ± 10.3	32.0 ± 17.7	0.74	29.7 ± 13.8	0.55
	fatigued	32.8 ± 14.4	37.2 ± 21.9	18.7 ± 14.7	0.26	29.6 ± 17.0	0.07
Wrist radial deviation	fresh†‡	-8.3 ± 9.2	0.1 ± 8.1	-7.8 ± 10.0	0.87	-5.3 ± 9.1	0.76
	fatigued†	1.5 ± 11.8	-15.6 ± 17.8	-5.0 ± 11.0	0.39	-6.3 ± 13.5	0.15
Wrist ulnar deviation	fresh	35.9 ± 8.5	36.7 ± 11.6	40.8 ± 9.6	0.80	37.8 ± 9.9	0.64
	fatigued	32.8 ± 7.1	41.8 ± 17.4	36.4 ± 13.3	0.63	37.0 ± 12.6	0.40
Elbow flexion	fresh	112.4 ± 11.0	109.4 ± 11.3	110.1 ± 6.3	0.79	110.6 ± 9.5	0.62
	fatigued	110.5 ± 8.9	108.8 ± 11.2	109.3 ± 5.7	0.82	109.5 ± 8.6	0.67
Elbow extension	fresh	158.8 ± 5.1	162.2 ± 5.7	160.8 ± 7.6	0.75	160.6 ± 6.1	0.56
	fatigued	158.5 ± 9.2	158.8 ± 9.4	159.1 ± 7.2	0.90	158.8 ± 8.6	0.81
Shoulder extension	fresh	-54.3 ± 5.6	-55.1 ± 4.3	-53.6 ± 4.1	0.77	-54.3 ± 4.7	0.59
	fatigued	-50.7 ± 6.5	-52.9 ± 12.7	-54.0 ± 5.2	0.67	-52.5 ± 8.1	0.45
Shoulder flexion	fresh	30.2 ± 7.4	25.1 ± 9.8	29.6 ± 10.6	0.81	28.3 ± 9.3	0.66
	fatigued	33.2 ± 11.8	31.3 ± 14.	29.7 ± 16.6	0.55	31.4 ± 14.1	0.30
Shoulder abduction	fresh	41.2 ± 7.0	40.1 ± 7.2	42.6 ± 5.2	0.65	41.3 ± 6.5	0.42
	fatigued	43.3 ± 6.4	44.0 ± 8.2	44.0 ± 7.5	0.14	43.8 ± 7.4	0.02
Shoulder adduction	fresh	10.4 ± 8.0	13.7 ± 8.4	16.1 ± 7.7	0.41	13.4 ± 8.0	0.17
	fatigued	17.1 ± 4.9	14.1 ± 12.7	12.9 ± 11.2	0.57	14.7 ± 9.6	0.32
Trunk flexion	fresh	67.2 ± 15.7	68.9 ± 13.7	66.5 ± 14.5	0.59	67.5 ± 14.5	0.35
	fatigued	57.6 ± 19.1	55.2 ± 20.5	62.7 ± 22.4	0.74	58.5 ± 20.7	0.55
Trunk extension	fresh	102.9 ± 3.9	100.3 ± 4.8	101.4 ± 4.2	0.63	101.5 ± 4.3	0.40
	fatigued	101.5 ± 3.1	104.5 ± 6.8	104.1 ± 8.6	0.48	103.4 ± 6.2	0.23

ICC = Intraclass correlation

†Significant difference between test 1 and test 2

\*Coefficient of Determination

‡Significant difference between 2 and test 3

tangential to the handrim, and peak moment applied radial to the handrim were statistically stable at various speeds of propulsion. Measures of the temporal characteristics of wheelchair propulsion found in the current study are within the ranges reported in previous studies that employed similar velocities (2,3,6,8). Although various kinematic models have been used, the peak ranges of motion in the current study were similar to those in other studies (2,9,10,12,22). Veeger, van der Woude, and Rozendal (3) compared propulsion force measurements of nondisabled individuals during submaximal exercise tests on a wheelchair ergometer to measurements obtained while propelling a wheelchair on a motor-driven treadmill. The authors reported that the wheelchair ergo-

meter is a valid tool for simulating wheelchair propulsion (3). The mean resultant force reported in this current study ( $124.7 \pm 40.7$  N) was similar to that reported by Veeger, van der Woude, and Rozendal (3) ( $133.4 \pm 23.4$  N) at similar velocities. Resultant forces and moments reported by other authors differ (5,13). The differences in resultant forces and moments may be attributed to different testing speeds (5). Robertson et al. (1996) reported differences in tangential forces, depending on the subject's level of experience in wheelchair propulsion. Additionally, there is a variety of definitions for kinetic variables used by different researchers, which complicates comparisons across studies (5,23). Forces and moments measured with two different prototypical

**Table 4.**

Peak angular displacement during handrim contact, during fresh and fatigued states, across tests.

Variable (°)	State	Test 1 (mean ± SD)	Test 2 (mean ± SD)	Test 3 (mean ± SD)	ICC (R)	Grand Mean ± SD	(R <sup>2</sup> )*
Wrist extension	fresh	-31.1 ± 10.2	-31.4 ± 11.0	-34.9 ± 9.9	0.06	-32.5 ± 10.4	0.00
	fatigued	-27.6 ± 13.4	-34.4 ± 16.5	-38.2 ± 9.8	0.19	-33.4 ± 13.2	0.04
Wrist flexion	fresh	25.8 ± 15.4	21.5 ± 15.0	29.8 ± 17.3	0.76	25.7 ± 15.9	0.58
	fatigued	30.9 ± 14.3	29.7 ± 17.8	16.6 ± 15.7	0.44	25.8 ± 15.9	0.19
Wrist radial deviation	fresh†‡	-3.0 ± 12.8	7.2 ± 12.0	-2.7 ± 12.8	0.87	0.5 ± 11.9	0.76
	fatigued	6.1 ± 9.7	-7.3 ± 20.8	0.1 ± 11.1	0.57	-0.4 ± 13.9	0.32
Wrist ulnar deviation	fresh	33.0 ± 11.1	35.8 ± 12.0	36.9 ± 14.8	0.83	35.2 ± 12.6	0.69
	fatigued	31.4 ± 7.8	31.2 ± 18.1	36.3 ± 13.2	0.62	33.0 ± 13.0	0.38
Elbow flexion	fresh	113.5 ± 11.8	110.3 ± 11.2	96.5 ± 34.7	0.12	106.8 ± 19.2	0.01
	fatigued	113.3 ± 10.4	110.6 ± 11.1	111.6 ± 5.5	0.85	111.8 ± 9.0	0.72
Elbow extension	fresh	153. ± 5.7	156.5 ± 7.3	157.9 ± 7.1	0.73	155.9 ± 6.7	0.53
	fatigued	152.0 ± 10.9	153.5 ± 10.2	155.0 ± 5.3	0.76	153.5 ± 8.7	0.58
Shoulder extension	fresh	-49.4 ± 7.3	-52.5 ± 4.8	-50.8 ± 3.9	0.60	-50.9 ± 5.3	0.36
	fatigued	-49.1 ± 7.0	-51.1 ± 12.2	-50.9 ± 5.4	0.74	-50.4 ± 8.2	0.55
Shoulder flexion	fresh	26.1 ± 9.2	21.1 ± 9.4	26.4 ± 10.4	0.85	24.5 ± 9.7	0.72
	fatigued	26.9 ± 12.0	26.8 ± 12.8	25.4 ± 17.9	0.53	26.3 ± 14.2	0.28
Shoulder abduction	fresh	38.8 ± 5.7	38.0 ± 7.1	39.3 ± 5.1	0.70	38.7 ± 6.0	0.49
	fatigued	40.4 ± 6.6	39.4 ± 7.3	41.4 ± 7.1	0.32	40.4 ± 6.3	0.10
Shoulder adduction	fresh	15.8 ± 7.3	15.6 ± 9.7	14.3 ± 9.3	0.64	15.2 ± 8.8	0.41
	fatigued	17.9 ± 5.7	16.6 ± 13.3	15.6 ± 8.7	0.56	16.7 ± 9.2	0.31
Trunk flexion	fresh	69.9 ± 14.8	69.9 ± 13.5	62.0 ± 25.1	0.68	67.3 ± 17.8	0.46
	fatigued	60.4 ± 18.5	58.0 ± 19.4	63.9 ± 22.6	0.78	60.8 ± 20.2	0.61
Trunk extension	fresh	101.2 ± 4.5	99.4 ± 5.4	99.9 ± 3.7	0.45	100.2 ± 4.6	0.20
	fatigued	100.6 ± 3.4	103.4 ± 6.8	102.9 ± 9.0	0.43	102.3 ± 6.4	0.18

ICC = Intraclass correlation †Significant difference between test 1 and test 2  
 \*Coefficient of determination ‡Significant difference between test 2 and test 3

instrumented wheelchair handrims have been reported to be valid (3,4). In the current study, these variables, with the exception of a few components, were found to be reproducible. The propulsive moments ( $M_z$ ) were shown to be highly reliable in both fresh and fatigued states, with intraclass correlation coefficients of 0.81 and 0.85, respectively. Conversely, the resultant force was found to differ across trials in the fatigued state. Post hoc analysis revealed that the differences were between Tests 2 and 3. The impact of the fatigue state during a novel task may have contributed to these findings.

Studying experienced wheelchair users across multiple propulsion cycles, Rao (2) revealed high coefficient of multiple correlation values and low root mean square

errors, indicating consistent intercycle kinematic patterns within individual subjects. Some variability in joint patterns was present between subjects. Variance was attributed to marker placement as well as differences in propulsion style. Joint kinematic measurements in the current study demonstrated some inconsistencies when expressed as peak angles across the entire cycle and as the peak angle occurring at contact, and release, as well as throughout contact. Significant differences were found in radial deviation across the entire cycle and poor correlation ( $R = 0.39$ ) was found in the fatigued state. Additionally, poor correlations ( $R < 0.60$ ) were found in wrist flexion and extension in both methods of reporting. The process of digitizing also creates a possible source of

**Table 5.**  
Kinematic variables at contact.

Variable (°)	State	Test 1 (mean ± SD)	Test 2 (mean ± SD)	Test 3 (mean ± SD)	ICC (R)	Grand Mean ± SD	(R <sup>2</sup> )*
Wrist flexion (+)/extension (-)	fresh	-19.0 ± 11.9	-15.5 ± 16.6	-23.0 ± 10.3	0.57	-19.2 ± 12.9	0.32
	fatigued	-20.1 ± 11.7	-3.9 ± 23.8	-19.8 ± 14.9	0.49	-14.6 ± 16.8	0.24
Wrist radial (-)/ulnar deviation (+)	fresh	11.2 ± 5.9	10.7 ± 10.2	5.8 ± 5.8	0.78	9.2 ± 7.3	0.61
	fatigued	11.8 ± 7.1	4.0 ± 14.0	7.7 ± 9.5	0.69	7.8 ± 10.2	0.48
Elbow flexion/extension (increasing)	fresh	127.1 ± 11.9	128.4 ± 8.3	126.1 ± 12.5	0.42	127.2 ± 10.9	0.18
	fatigued	124.5 ± 7.0	128.7 ± 12.7	124.4 ± 7.8	0.55	125.9 ± 9.16	0.30
Shoulder flexion (+)/extension (-)	fresh	-47.2 ± 7.3	-50.8 ± 5.1	-47.4 ± 5.6	0.61	-48.5 ± 5.9	0.37
	fatigued	-46.4 ± 6.8	-49.0 ± 13.4	-48.7 ± 4.8	0.58	-48.0 ± 8.3	0.34
Shoulder abduction (+)/adduction (-)	fresh	22.9 ± 11.6	19.7 ± 12.8	19.3 ± 7.8	0.74	20.7 ± 10.7	0.55
	fatigued	24.7 ± 9.3	23.7 ± 18.6	25.3 ± 11.9	0.58	24.6 ± 13.2	0.34
Trunk flexion (increasing)/extension	fresh	99.5 ± 4.3	98.0 ± 5.3	95.1 ± 8.8	0.18	97.5 ± 6.1	0.03
	fatigued	98.1 ± 3.4	96.9 ± 18.7	100.7 ± 8.7	0.34	98.6 ± 10.3	0.12
ICC = Intraclass correlation		*Coefficient of determination					

error. During wheelchair propulsion, wrist measurements are especially susceptible to marker dropout as the subject rotates through the propulsion cycle, resulting in less than three visible markers in a minimum of two cameras. Loss of markers or marker collision creates the need for marker interpolation, resulting in additional variance among repeated kinematic measurements. Additional camera views or perhaps the use of markers applied to a stick at the wrist may reduce this problem.

Fatigue has been reported to increase peak handrim forces, increase trunk flexion and decrease wrist deviation motions (14). However, the reliability of these effects was not reported previously. Good reliability of the temporal, kinetic, and the majority of kinematic variables at fatigue was found in this study. The weak reliability coefficients were found predominately in wrist angles throughout the entire propulsion cycle and/or at contact. This variability may be associated with subjects' attempts to adapt their propulsion styles as they fatigue.

**Table 6.**  
Angular displacement at time of release, fresh and fatigued states, across tests.

Variable (°)	State	Test 1 (mean ± SD)	Test 2 (mean ± SD)	Test 3 (mean ± SD)	ICC (R)	Grand Mean ± SD	(R <sup>2</sup> )*
Wrist flexion (+)/extension (-)	fresh	9.2 ± 15.5	9.2 ± 25.4	16.3 ± 18.0	0.37	11.6 ± 19.6	0.14
	fatigued	23.2 ± 17.3	8.6 ± 33.4	2.6 ± 26.0	0.05	11.4 ± 25.6	0.00
Wrist radial (-)/ulnar deviation (+)	fresh	19.3 ± 14.0	27.1 ± 9.7	22.3 ± 19.7	0.86	22.9 ± 14.5	0.74
	fatigued	26.1 ± 7.8	17.6 ± 25.2	26.3 ± 15.6	0.64	23.3 ± 16.2	0.41
Elbow flexion/extension (increasing)	fresh	145.5 ± 9.7	153.5 ± 8.0	148.0 ± 18.7	0.28	149.0 ± 12.1	0.08
	fatigued	149.2 ± 10.3	146.5 ± 13.0	148.1 ± 8.9	0.56	148.0 ± 10.8	0.31
Shoulder flexion (+)/extension (-)	fresh	19.6 ± 13.3	19.2 ± 9.5	21.9 ± 10.2	0.81	20.2 ± 11.0	0.66
	fatigued	24.7 ± 12.7	20.8 ± 15.4	22.3 ± 19.5	0.51	22.6 ± 15.9	0.26
Shoulder abduction (+)/adduction (-)	fresh	29.1 ± 7.5	30.0 ± 6.5	31.4 ± 3.3	0.40	30.2 ± 5.8	0.16
	fatigued	31.0 ± 7.9	30.6 ± 6.8	30.3 ± 9.0	0.31	30.6 ± 7.9	0.10
Trunk flexion (increasing)/extension	fresh	73.2 ± 14.0	71.4 ± 13.0	66.7 ± 17.2	0.69	70.4 ± 15.0	0.48
	fatigued	62.9 ± 19.0	65.7 ± 22.1	66.2 ± 22.7	0.65	64.9 ± 21.5	0.42
ICC = Intraclass correlation		*Coefficient of determination					

**Table 7.**  
Reliability coefficients for changes in variables from fresh to fatigued states.

Temporal Variables	ICC (R)	Kinematic Variables			
		Entire Cycle	ICC (R)	During contact	ICC (R)
Stroke frequency (cycles/s)	0.69	wrist extension	0.48	wrist extension	0.10
Contact time (s)	0.85	wrist flexion	0.71	wrist flexion	0.75
Contact time (% cycle)	0.90	wrist radial deviation	0.50	wrist radial deviation	0.23
		wrist ulnar deviation	0.98	wrist ulnar deviation	0.84
		elbow flexion	0.96	elbow flexion	0.34
		elbow extension	0.91	elbow extension	0.78
		shoulder extension	0.71	shoulder extension	0.78
		shoulder flexion	0.88	shoulder flexion	0.91
		shoulder abduction	0.92	shoulder abduction	0.85
		shoulder adduction	0.80	shoulder adduction	0.74
		trunk flexion	0.85	trunk flexion	0.84
		trunk extension	0.57	trunk extension	0.69
Handrim Variables Force, Moments	ICC (R)	At Contact	ICC (R)	At Release	ICC (R)
$F_{x+}$	0.92	wrist flexion/extension	0.59	wrist flexion/extension	0.63
$F_{y-}$	0.80	wrist radial/ulnar deviation	0.31	wrist radial/ulnar deviation	0.75
$F_{z-}$	0.88	elbow flexion/extension	0.61	elbow flexion/extension	0.84
$F_{z+}$	0.78	shoulder flexion/extension	0.87	shoulder flexion/extension	0.91
$M_{x+}$	0.91	shoulder abduction/ adduction	0.74	shoulder abduction/ adduction	0.79
$M_{x-}$	0.74	trunk flexion/extension	0.76	trunk flexion/extension	0.74
$M_{y+}$	0.75				
$M_{y-}$	0.80				
$M_z$	0.89				
$F_{res}$	0.79				
$M_{res}$	0.79				

ICC = Intraclass correlation

As stated previously, the velocity selected for this study was lower than in some other studies (2,3,6,8,24). The magnitudes of the variance for force and temporal variables have been reported to increase with increasing velocity (3,8). Therefore, it is possible that the choice of a faster propulsion velocity may have resulted in greater intrasubject variance, but the consistency of this variance is unlikely to change between tests.

Because of the novel nature of the task, subjects may have been variable in their individual performances in the more challenging fatigued state. Although the subjects were given practice sessions to acclimate to the task, a fatigued state was not attained during those acclimation sessions. More vigorous and physically demanding exposure may be required for inexperienced individuals to develop consistent strategies to complete the task in a fatigued condition. The use of study subjects

experienced in wheelchair propulsion might have eliminated this variability.

## CONCLUSION

The majority of the biomechanical variables measured (temporal characteristics, handrim kinetics, and joint kinematics) for nonwheelchair users were reliable. Fatigue was found to increase variance, predominately in the wrist kinematics. Experiments that include participants not experienced in wheelchair propulsion must consider the lack of experience as a possible source of bias. Inexperienced individuals may require greater acclimation time and intensity for kinematic repeatability to be achieved, or consideration should be given to using experienced wheelchair users.

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