Influence of trunk flexion on biomechanics of wheelchair propulsion

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Abstract—. Propulsion styles that are characterized by high stresses may influence the susceptibility of manual wheelchair users (MWCU) to upper limb injury. An experimental, cross-sectional study was designed to compare physiological and bio-mechanical characteristics of wheelchair propulsion in two groups of MWCU—a trunk flexion propulsion style group (FG) and a non-trunk flexion propulsion style group (NFG)—across fresh and fatigued states. Data on joint kinetics and kinematics, handrim kinetics, propulsion temporal characteristics, and electromyography were collected at the fresh and fatigued states, and oxygen uptake was collected continuously, to characterize wheelchair propulsion performance of 19 MWCU during a sub-maximal exercise test to exhaustion. The FG was characterized by a more flexed trunk position accompanied by greater shoulder flexion and elbow extension ($p < 0.05$), which was accentuated with fatigue when compared to the NFG. When fatigued, marked decreases ($p < 0.05$) in key propulsion muscle activity were observed in the FG, but not in the NFG. Temporally, the

FG decreased contact time on the handrim by 1 percent of the propulsion cycle when fatigued, in contrast to the NFG who increased contact time by 7 percent ($p < 0.05$). Results suggest that a trunk flexion style of wheelchair propulsion may lead to potentially debilitating upper limb injury, since these individuals appear to be compensating for peripheral muscle fatigue.

Key Words: *biomechanics, fatigue, kinematics, kinetics, propulsion style, wheelchair propulsion.*

INTRODUCTION

Chronic health problems due to long-term use of manual wheelchairs significantly impair the independence of individuals who depend upon wheelchairs for locomotion. Manual wheelchair users (MWCU) are likely to experience upper limb pain (1–4) which limits physical performance (4,5) and quality of life (6–8). In different study samples, sixty-four to 73 percent of MWCU with spinal cord injuries (SCI) have reported shoulder or wrist joint pain (1,2,4,9) and 67 percent of MWCU have upper limb mononeuropathies defined by strict electrodiagnostic criteria (3). Gellman and coworkers (9) and Pentland et al. (5) reported increases in the

prevalence of upper limb pain as a function of time since sustaining SCI. Rodgers, et al. (10) suggested the frequency and duration of arm use may compound the risk for overuse injury in MWCU. Others have reported mechanical inefficiency during wheelchair propulsion (11–13) and activities of daily living (14,15), suggesting propulsion mechanics may be a pathogenic mechanism for these injuries.

Rapid application of propulsion force (16,17), loading of joints while in extreme positions (18–21), high strength requirements (21–23), and metabolic demands (14,15,24,25) appear to be important contributors to upper limb injury in MWCU. Propulsion style, therefore, seems implicated as a possible determinant of the condition. Individual variants in propulsion style have been noted (26,27), with the amount of trunk flexion being widely variable among MWCU (27,28) and increasing with propulsion speed (29,30). Important differences in upper limb kinematics (28), propulsion temporal characteristics (28), and energy expenditure (30) have been identified based on varying degrees of trunk flexion.

We tested the hypothesis that a propulsion style incorporating excessive or exaggerated trunk flexion would result in biomechanical and physiological characteristics thought to incur painful upper limb injury, as compared to a style that does not incorporate trunk flexion. We predicted that a trunk flexion style of propulsion, specifically, would result in increased oxygen uptake (VO^2), and potentially harmful joint kinematics, kinetics, muscle activation patterns, and propulsion temporal characteristics. The aim of this study was to determine the degree to which theorized mechanisms of upper limb injury differ as a result of trunk flexion and non-trunk flexion propulsion styles in the MWCU.

METHODS

Subjects

Nineteen non-athlete MWCUs participated in this study (**Table 1**). Three were female and the majority of participants had spinal cord injuries. Ages ranged from

Table 1.
Subject characteristics.

Sex	Age	Injury	Control	Time	Hgt	Wgt	Arm	Trunk
M	21	T5	absent	7	180.5	72.4	47.0	47.0
M	30	T3 incomplete	absent	12	185.4	66.4	77.0	51.0
M	32	spina bifida	present	6	157.5	56.8	73.0	51.0
M	33	bilat TTS*	present	10	188.0	97.3	80.0	57.0
M	36	T12	present	10	154.9	102.3	66.0	43.2
F	37	T7	absent	34	121.9	52.3	60.0	32.0
F	39	multi-trauma	present	12	177.8	75.0	70.5	40.0
M	40	T2	absent	10	167.5	100.0	73.0	47.0
M	42	T12 incomplete	present	11	179.1	95.5	82.0	45.0
M	44	T3	absent	24	178.8	71.2	68.5	46.5
F	45	T7	absent	24	162.5	44.9	43.0	43.0
M	49	L5 imcomplete	present	3	187.0	127.0	50.5	48.0
M	50	T5	absent	28	186.0	90.9	45.0	48.5
M	52	T6	absent	12	180.3	66.8	74.0	42.0
M	52	T5	absent	26	180.3	78.5	75.0	49.0
M	54	multi-trauma	present	26	182.9	77.3	73.0	51.0
M	54	T5 incomplete	present	5	180.3	72.7	71.1	45.7
M	58	T8	absent	38	178.0	76.3	46.0	42.0
M	68	T4 incomplete	present	21	187.5	79.5	45.0	51.0
Mean	**44.0**			**16.8**	**174.5**	**79.1**	**64.2**	**46.3**
SD	**11.3**			**10.4**	**16.1**	**19.6**	**13.5**	**5.4**

Age in years; Control=trunk control; Time=time in wheelchair, in years; Hgt=height in cm; Wgt=weight in kg; Arm=arm length in cm; Trunk=trunk length in cm; *TTS=tarsal tunnel syndrome.

21–68 years and durations of wheelchair use ranged from 3–38 years. Self-perceived fitness ratings were in the medium range, and none were actively participating on a sport team or in recreational sports. Before testing, a medical examination was completed by a physician familiar with the requirements for study participation. Inclusion criteria included use of a manual wheelchair for at least one year prior to the study, wheelchair use for the majority of home and community mobility, and absence of upper limb involvement, ventilatory involvement, or systemic diseases that would preclude or limit exercise testing. Before testing, written, informed consents were obtained in accordance with the procedures approved by the Institutional Review Board.

Instrumentation

The wheelchair measurement system included the instrumented wheelchair ergometer, three Peak three-dimensional (3-D) Charged Coupled Device (CCD) cameras and video acquisition system (Peak Performance Technologies, Colorado Springs, CO), an eight channel surface Electromyographic (EMG) telemetry system (Noraxon USA Inc, Scottsdale, AZ), and a metabolic cart (Cardio 2, Medical Graphics Corp, St. Paul, MN). All exercise tests were conducted on the wheelchair ergometer (**Figure 1**) with a 22-in diameter handrim. Components of a stationary bicycle ergometer were used to provide frictional propulsion resistance through a chain and sprocket system connected to the wheelchair axle at one end and to a flywheel at the other end (**Figure 2**).

Figure 1.
A manual wheelchair user prepared for data acquisition. Note instrumented wheelchair ergometer and locations of reflective markers used for joint kinematic data acquisition.
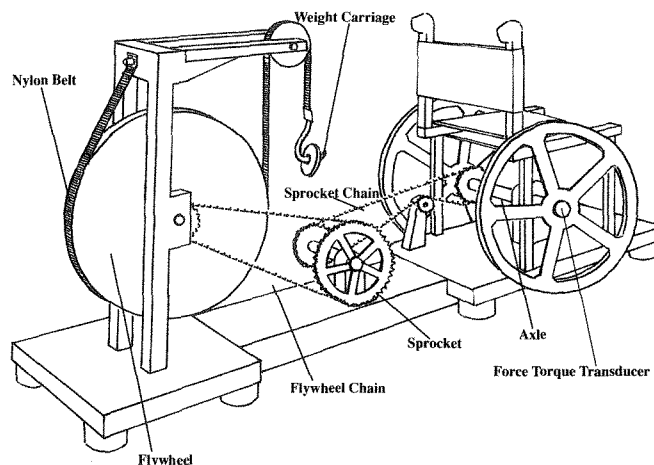

Figure 2.
Schematic diagram of the wheelchair ergometer showing the method used to apply resistance to the handrim.

Torque applied by the subject to the handrims of the wheels propelled the flywheel. A nylon belt was used to create a pulley system to which known weights could be applied, enabling precise control of resistance to the flywheel that was then transmitted to the wheels.

The wheels of the wheelchair ergometer had no camber, and the backrest was positioned perpendicular to the floor. Subjects used their own wheelchair cushions in the wheelchair ergometer seat. Wheelchair ergometer seat width and height were adjusted to match as closely as possible the wheelchair seat of each individual. Subjects were positioned in the wheelchair with their feet on a stepstool the height of which corresponded to that required to keep the thighs parallel to the floor.

The wheelchair ergometer was instrumented with a PY6-4 six-component force/torque transducer (Bertec Corp., Worthington, OH) in its wheel hub. The Bertec transducer used bonded strain gauges to measure handrim forces and moments in 3-D (six channels). It had a maximum torque (M_z) capacity of 150 N-m and a maximum plane-of-wheel force (F_x and F_y) capacity of 3,500 N. These capacities corresponded to a gain setting of unity, and the maximum usable range was set by user-selected gains. The full-scale output of the transducer was ±10 V. Since the transducer rotated with the wheel, connections were enabled using a slip ring. A potentiometer monitored the angular positions of the wheel, transducer, and handrim assembly. The amplified electrical signals from the force transducer and potentiometer were collected using an analog-to-digital converter and acquisition soft-

ware (Peak Performance Technologies, Colorado Springs, CO). Handrim kinetic, temporal, and potentiometer data were collected 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. The speedometer (Cateye Cyclocomputer, Model CC-CD 100, Osaka, Japan) had an accuracy of ±0.2 km/hr (±1rpm) and range of 0–100km/hr. Wheelchair propulsion velocity was maintained at 32 rpm (3 km/hr) for all testing. The relatively slow propulsion velocity of 3 km/hr was selected because of the inclusion of elderly individuals in the sample and because of the difficulty of maintaining higher speeds at high resistance loads.

Kinematic data were collected at 60 frames/s using the video cameras and data acquisition system. Calibration was performed by videotaping a precisely calibrated cube with 24 known coordinate points and digitizing each point location. Direct linear transformation computations were performed using the Cartesian (X-Y-Z) coordinate system, with camera lens distortion corrected using the Peak software package. System accuracy of the Peak Performance Measurement System has been reported by the company to exhibit a 0.5 mm average error when calibrating in static conditions. Our own dynamic evaluations have determined the angular calculations to be within 1° at speeds up to $300^{\circ}/s$. Cameras were positioned so that all markers could be viewed in at least two cameras at any time during the experiment. One camera was located directly in front of the wheelchair ergometer at a distance of 25 feet and the other two cameras were approximately 45° to each side of the center camera. Spherical retroreflective markers were placed on the dorsal surface of the fifth metacarpal head, medial styloid process, lateral styloid process, radial head, acromion, and greater trochanter. Joint marker displacements were recorded using the camera, and joint angles (trunk, shoulder, elbow and wrist flexion/extension, shoulder abduction/adduction, and wrist radial/ulnar deviation), velocities and accelerations were calculated using the Peak Performance software.

Joint kinetics were calculated using a 3-D, linked segment model (31). 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. Definitions for the global and local (handrim, hand, forearm, upper arm, and trunk) coordinate systems have been

described in detail (**Figure 3**). Displacement data collected using the video cameras were differentiated to form velocity and acceleration vectors. The linear acceleration of each marker was used to interpolate the linear acceleration of the center of mass for each limb segment. These estimates were then transformed into respective body coordinates. The forces and moments measured from the hub transducer were transformed from handrim to hand coordinates, using the radius of the handrim of the wheelchair, and were used as the input to the hand equations. Raw data were smoothed using a Butterworth low-pass filter with a cutoff frequency of 6 Hz. Contact phase was defined as the entire time of handrim loading. These
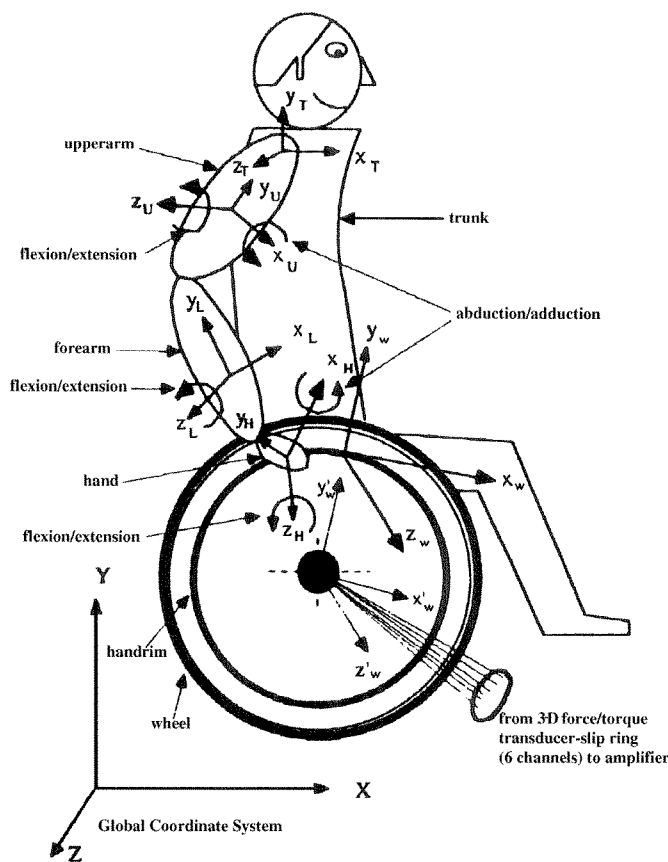

Figure 3.
Global and local coordinate systems for the wheelchair propulsion model. Body segments (T, trunk; u, upper arm; L, lower arm or forearm; H, hand) and segmental motions as a function of moments about local axes (flexion/extension, abduction/adduction, ulnar/radial deviation) are indicated. Note: x'y'z' are coordinates for the hub transducer; x'w-y'w-z'w coordinates are parallel to xw-yw-zw coordinates; xyz coordinates with a w subscript refer to wheel coordinates. (Reprinted by permission from Mary M. Rodgers, Srinivas Tummarakota, and Junghsen Lieh, 1998, "Three dimensional dynamic analysis of wheelchair propulsion," *Journal of Applied Biomechanics* 14(1):81.)

motion vectors, forces, and torques, together with anthropometric data, were the input variables to a program that computes the forces and moments of the wrist using an inverse dynamics process. The recursive program then determines the joint forces and moments of the elbow and shoulder.

EMG data were collected from an eight-channel telemetry system at 960 Hz (Noraxon USA Inc, Scottsdale, AZ). Pairs of surface electrodes were placed on the right side of each subject, over motor point locations for the pectoralis major, anterior and posterior deltoids, middle trapezius, biceps and triceps brachii, flexor carpi ulnaris and extensor carpi radialis muscles. Signals were processed by computing the mathematical root-mean square (RMS) value of the EMG signal (32). The signal was then normalized to the EMG level recorded during maximal isometric voluntary contraction (MVC) for each muscle using standardized manual muscle test positions (33).

Heart rate was monitored by a telemetered pulse rate monitor that included a transmitter, attached by a belt to the thorax, and a receiver with a digital display. Cardiorespiratory measurements were made using breath-by-breath open circuit spirometry. The metabolic cart system included rapid response oxygen (zirconium cell) and carbon dioxide (infrared cell) analyzers, and a pneumotachometer, all interfaced with a microcomputer. Cardiorespiratory measurements were averaged over 30 s for reporting purposes.

Exercise Tests

A maximal graded exercise test (GXT) on the wheelchair ergometer was used to establish resistance load for the fatigue test. For the GXT, subjects rested (6 min), propelled the wheelchair at a velocity of 3 km/hr (32 rpm) without a load (3 min), then continued while weight was incrementally added at a rate of 0.3 kg every 3 min, to increase power output. 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 designated velocity. Heart rate was recorded at the end of each 3-min stage and cardiorespiratory information was continuously recorded as 30-s averages.

Two to seven days following the GXT, subjects completed the fatigue test. Load corresponded to 75 percent of the peak VO_2 that occurred during the GXT. Since resistance load used for the fatigue test was based on the

individual's performance on the GXT, resistance loads varied from individual to individual. For the fatigue test, subjects rested (6 min), propelled the wheelchair at 3 km/hr without a load (3 min), then continued propelling at the sub-maximal load until volitional exhaustion. Subjects were monitored and encouraged to maintain the designated velocity. Heart rate was recorded at the end of each 3-min stage and cardiorespiratory information was continuously recorded as 30-s averages. Propulsion mechanics, including handrim kinetics, joint kinematics, EMG, and temporal characteristics, were collected for 6 s (3 propulsion cycles) under three conditions: during the last 30 s of wheeling without a load, after 2.5 min of wheeling with a load (fresh), and just before exhaustion (fatigued).

Strength Tests

Strength of the dominant upper limb was tested using two techniques. Handgrip strength was tested in all subjects using a Baseline hydraulic hand dynamometer. Three maximal grips were averaged to represent grip strength. Eleven of the subjects were also tested using a Kincom isokinetic dynamometer (60º/s) to test other upper limb joint motions which included shoulder flexion/extension (0–55º), shoulder internal/external rotation (0–55º), elbow flexion/extension (5–70º), wrist flexion/extension (0–45º), and wrist radial/ulnar deviation (0–45º). The dynamometer speed of 60º/s corresponded to the joint angle velocities occurring during wheelchair propulsion at a velocity of 3 km/hr. Three maximal voluntary contractions were collected concentrically and eccentrically. Mean peak moment was used to characterize strength for each motion.

Groups

Subjects were divided into two groups based on trunk position (upright sitting position=90º relative to the horizontal plane). Trunk angles greater than 90º were defined as trunk extension and those less than 90º were defined as trunk flexion. Subjects who demonstrated more than 10º of trunk flexion (in the 0–80º range) during contact when fresh or fatigued, and those whose trunk flexion increased more than 10º from the fresh to the fatigued state were included in the flexion group (FG). All others were included in the non-flexion group (NFG). Based on the group assignment criteria, 9 subjects were assigned to the FG and 10 to the NFG. The FG included four who had increased trunk flexion angle, one who had

increased flexion range from fresh to fatigued state, and four who met both criteria. Anthropometric characteristics of subjects are shown in **Table 1**. Wheelchair seat widths ranged from 15–19 in (38–48 cm) with a mean of 16.4±1.8 in (41.7±4.6 cm).

Data Analysis

EMG peak amplitude was determined from the average of three propulsion cycles for each trial. Timing of muscle activity and integrated EMG (total activity or area under the curve) were determined from a representative cycle for each trial and related to propulsion and recovery phases. Propulsion phase (contact time) was defined as the entire time of handrim loading, with recovery phase being defined as the time when the hand was not in contact with the handrim. All EMG data were normalized to the maximum isometric voluntary contraction of each muscle (percent MVC). Kinetic and kinematic data were averaged over three cycles (contact to contact) for each condition (fresh and fatigued). The decision to average three cycles was based on the low cycle-to-cycle variability achieved, attributed to the fact that wheelchair propulsion is a constrained activity which was also performed at a set velocity. Any cycles that were aberrant for any reason (i.e., miscontact or slippage on the wheel) were not included in the analysis.

Joint kinetics and kinematics, handrim kinetics, propulsion temporal characteristics, and EMG data were compared between groups during the fresh and fatigued states. Oxygen uptake (VO_2) during the fatigue test was characterized by absolute VO_2 (ml/min), oxygen uptake divided by maximal oxygen uptake (VO_2/VO_{2max}), and oxygen uptake divided by power output (VO_2/resistance load). These variables were compared at times corresponding to one-third, two-thirds, and completion of the test for both groups. Comparisons were performed using Analysis of Variance (ANOVA), with first effect repeated (fresh and fatigued states) and second effect by group (FG and NFG). Interaction effects were further examined through post-hoc analysis. Type-I error threshold was held at or below 5 percent ($p \leq 0.05$).

Subject characteristics, including age, height, weight, length of time in a wheelchair, and upper limb strength were compared between groups using a two-sample t-test. Trunk control was classified as present or absent for all subjects. Those with motor-complete spinal cord injuries above the twelfth thoracic vertebra (T12) were classified with absent trunk control. Those with motor incomplete spinal cord injuries, complete injuries below T12, or other types of injuries were classified with trunk control. Gender and

trunk control were compared between groups using a two-way frequency analysis. Statistical significance was established when the type-I error probability was not observed above five percent ($p \leq 0.05$).

RESULTS

Trunk Flexion Versus Non-trunk Flexion Groups

No significant inter-group differences in age, height, weight, length of time as a wheelchair user, gender, or trunk control were observed. Upper limb strength (grip and isokinetic) was not significantly different between groups, with the exception of the concentric shoulder extension moment, which was greater ($p<0.04$) in the NFG (67±4 N-m) than the FG (41±18 N-m). Failure of differences in the means to reach the 0.05 threshold for significance across all muscle groups appeared to be the result of the small number of subjects completing isokinetic dynamometer tests (NFG=3 subjects; FG=8 subjects).

Kinematic differences are depicted for representative subjects from each group in **Figure 4**. Significant inter-group differences in trunk position were found (**Table 2**). Inter-group kinematic differences at the shoulder and
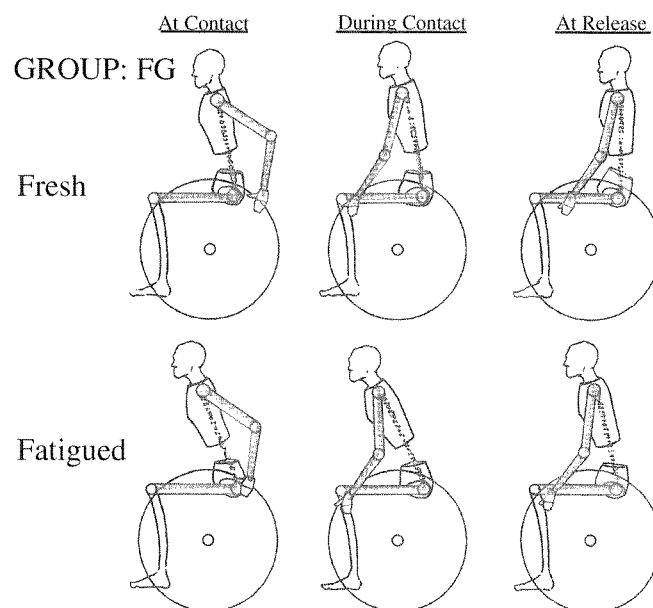
Figure 4.
Representative subjects from the flexion group (FG) and non-flexion group (NFG) depicting joint kinematic differences. Greater trunk flexion, along with associated differences in shoulder and elbow angles, can be seen in the FG compared to the NFG at contact, during contact and at release during the fresh and fatigued states.

Table 2.
Trunk flexion kinematics.

		FG	NFG	
At contact				*State*
Fresh		91±11	97±7	*94±9*
Fatigued		83±12 *	99±9*	*91±11*
	Group	*87±12*	*98±8*	
During contact				*State*
Fresh		81±11	92±6*	*87±9*
Fatigued		75±10 *	93±8*	*84±9*
	Group	*78±11*	*93±7*	
At release				*State*
Fresh		85±11	94±6*	*90±9*
Fatigued		78±9 *	94±8*	*86±9*
	Group	*82±10*	*94±7*	
ROM (during contact)				*State*
Fresh		12±6	7±5	*10±6*
Fatigued		13±8	7±4	*10±6*
	Group	*13±7*	*7±5#*	

Joint angles in degrees; FG=trunk flexion; NFG=non-trunk flexion; ROM=range of motion; *=significantly different interaction (p<0.05); #=significantly different main effect (p<0.05); scores are means±SD.

elbow were also observed (**Tables 3 and 4**, respectively). During contact (p<0.006) and at release (p<0.004), the FG demonstrated greater shoulder flexion than the NFG. Elbow extension was greater for the FG during contact (p<0.013) and at release (p<0.031). No significant differences were found in the wrist kinematics (**Table 5**).

Resistance applied to the wheelchair (FG=1,500±1,071 g, NFG=1,350±852 g) and the length of time the wheelchair was propelled (FG=24.8±15.9 min, NFG=29.9±26.4 min) were similar between groups. VO_{2max} during the fatigue test was also similar between groups (FG=1,087±456 ml/min, NFG=1,080±319 ml/min). There were no significant inter-group differences in VO^2, VO_2/VO_{2max}, and VO_2/load values for each group at the beginning third, middle third, and end of the fatigue test (**Table 6**). Only two inter-group differences in joint kinetics were apparent. The FG had 7 N less (p<0.022) posterior force (F_x) and 29 N more (p<0.046) medial force (F_z) at the elbow during contact compared to the NFG.

Group differences in muscle activity patterns were seen in three muscles. The FG demonstrated 20 percent earlier cessation of flexor carpi ulnaris (p<0.001) and pectoralis major (p<0.031) activity. The total biceps activity was greater (p<0.034) for FG compared to NFG (FG=18 percent, NFG=10 percent).

Fresh Versus Fatigued State

Fatigue main effects were seen in some of the variables. Both groups demonstrated more shoulder flexion (p<0.047 during contact, p<0.018 at release) when fatigued. Both groups applied more inward force (F_z) to the handrim (p<0.03) when fatigued than when fresh (**Table 7**). Both groups demonstrated less (p<0.024) wrist flexion moment when fresh (−5±5 N-m) than when fatigued (−8±7 N-m), and less (p<0.022) radioulnar shear force (F_x) when fatigued (45±25 N fresh, 37±21 N fatigued). Fatigue altered muscle activity patterns in three muscles. With fatigue, both groups demonstrated significant decreases in peak amplitude of the biceps (p<0.006) and pectoralis major muscles (p< 0.025). Finally, fatigue-induced changes in both groups included decreased total biceps (p<0.006) and pectoralis major (P<0.04) activity, and earlier onset (p<0.02) and peak activity of the triceps (p<0.01).

Interactions

Interaction effects were seen for several variables (p<0.05). In the FG, trunk flexion increased 7–10% more with fatigue than in the NFG. When fatigued, shoulder flexion increased at contact by 6% in the FG, but not in the NFG. Propulsion temporal characteristics are shown in **Table 8**. The FG decreased contact time on the handrim by 1 percent of the propulsion cycle when fatigued (0.04 s decrease), in contrast to NFG who increased (p<0.033) contact time by 7 percent of the cycle (0.04 s increase) as they fatigued (**Figure 5**).

Interaction effects were also observed in muscle activity patterns. When fatigued, FG demonstrated significant decreases in peak amplitude of the flexor carpi ulnaris and triceps muscles, which did not occur in NFG (**Figure 6**). Although only significant for the flexor carpi ulnaris and triceps, the FG demonstrated declines in peak amplitude with fatigue across all muscle groups tested, while the NFG demonstrated increases with fatigue in all but three muscle groups.

DISCUSSION

Trunk Flexion Versus Non-trunk Flexion Groups

Inter-group propulsion style differences were characterized by variations in shoulder flexion and elbow extension. A pattern of greater shoulder flexion and elbow extension that accompanied trunk flexion was seen in the FG. Consistent with the observations of others

Table 3.
Shoulder kinematics.

		Flexion				Abduction		
		FG	NFG			FG	NFG	
At contact				*State*				*State*
Fresh		-42±12	-47±10	*-45±11*		41±8	44±11	*43±10*
Fatigued		-36±15 *	48±9*	*-42±12*		38±8	44±11	*41±10*
	Group	*-39±14*	*-48±10*		*Group*	*40±8*	*44±11*	
During contact				*State*				*State*
Fresh		23±11	8±12	*16±12*		43±9	47±9	*45±9*
Fatigued		25±8	13±10	*19±9 #*		41±8	46±8	*44±8*
	Group	*24±10*	*11±11#*		*Group*	*42±9*	*47±9*	
At release				*State*				*State*
Fresh		20±11	5±12	*13±12*		22±9	28±7	*25±8*
Fatigued		23±7	10±9	*17±8 #*		24±7	29±10	*27±9*
	Group	*22±9*	*8±11#*		*Group*	*23±8*	*29±9*	
ROM(during contact)				*State*				*State*
Fresh		67±9	58±21	*63±15*		23±10	20±7	*22±9*
Fatigued		63±14	63±15	*63±15*		19±9	21±7	*20±8*
	Group	*65±12*	*61±18*		*Group*	*21±10*	*21±7*	

Joint angles in degrees; FG=trunk flexion; NFG=non-trunk flexion; ROM=range of motion; *=significantly different interaction (p<0.05); #=significantly different main effect (p<0.05); scores are means±SD.

Table 4.
Elbow extension kinematics.

		FG	NFG	
		FG	NFG	
At contact				*State*
Fresh		121±8	112±7	*117±8*
Fatigued		115±13	111±5	*113±9*
	Group	*118±11*	*112±6*	
During contact				*State*
Fresh		153±8	142±9	*148±9*
Fatigued		152±6	145±9	*149±8*
	Group	*153±7*	*144±9*	
At release				*State*
Fresh		150±9	139±10	*145±10*
Fatigued		149±7	142±10	*146±9*
	Group	*150±8*	*141±10*	
ROM (during contact)				*State*
Fresh		48±11	40±13	*44±12*
Fatigued		48±10	45±10	*47±10*
	Group	*48±11*	*43±12*	

Joint angles in degrees; FG=trunk flexion; NFG=non-trunk flexion; ROM=range of motion; #=significantly different main effect (p<0.05); scores are means±SD.

(28,29), this pattern of movement allowed the FG to rely on trunk excursion to generate the translational forces necessary for wheelchair propulsion. In contrast, the NFG made contact with the wheel in a position of greater shoulder extension than the FG. This pattern did not change with fatigue. Trunk excursion was not characteristic of the NFG style, which relied on force generation from sources other than those translated through the trunk, such as muscular power (34), for propulsion. The FG also had larger medial elbow joint forces, which may be related to the greater elbow extension.

Inter-group propulsion style differences did not appear to be attributed to differences in the degree of volitional trunk control. Subjects with inadequate volitional trunk control would not be expected to adopt a trunk flexion propulsion style. Subjects in this study with the most impaired trunk control had spinal cord injuries between T2-T4 (n=4), with the majority at the T5 level and below or with injuries not involving the trunk (n=15). Most of the subjects in the current study appeared to have sufficient trunk control to allow some degree of forward lean. Other investigators have noted

Table 5.
Wrist kinematics.

		Extension				Deviation		
		FG	NFG			FG	NFG	
At contact				*State*				*State*
Fresh		-24±16	-23±19	*-24±18*		5±12	7±15	*6±14*
Fatigued		-28±14	-20±20	*-24±17*		6±7	3±16	*5±12*
	Group	*-26±15*	*-22±20*		*Group*	*6±10*	*5±16*	
During contact				*State*				*State*
Fresh		-37±12	-28±21	*-33±17*		37±12	29±22	*33±17*
Fatigued		-35±10	-26±19	*-31±15*		34±13	27±23	*31±18*
	Group	*-36±11*	*-27±20*		*Group*	*36±13*	*28±23*	
At release				*State*				*State*
Fresh		6±12	10±16	*8±14*		24±8	15±19	*20±14*
Fatigued		5±13	10±19	*8±16*		24±7	14±17	*19±12*
	Group	*6±13*	*10±18*		*Group*	*24±8*	*15±18*	
ROM(during contact)				*State*				*State*
Fresh		66±16	49±22	*58±19*		38±17	31±16	*35±17*
Fatigued		59±20	51±21	*55±21*		31±14	30±22	*31±18*
	Group	*63±18*	*50±22*		*Group*	*35±16*	*31±19*	

Joint angles in degrees; FG=trunk flexion; NFG=non-trunk flexion; Extension=wrist extension; Deviation=wrist ulnar deviation; ROM=range of motion; scores are means±SD.

Table 6.
Oxygen uptake.

	VO_2 (ml/min)		VO_2/VO_2max		VO_2/load	
	FG	NFG	FG	NFG	FG	NFG
1/3 of test	1035±431	846±459	96±12	78±30	83±32	72±36
2/3 of test	1027±440	964±276	94±5	91±15	82±32	91±44
At exhaustion	1015±419	979±295	94±5	91±8	83±36	93±45

FG=trunk flexion; NFG=non-trunk flexion; scores are means±SD.

Table 7.
Handrim kinetics.

Peak Forces (N)

	FG	NFG		FG	NFG		FG	NFG	
	Fx (tangential)		*State*	**Fy (radial)**		*State*	**Fz (medial)**		*State*
Fresh	70±32	67±14	*69±23*	-67±21	-65±19	*-66±20*	11±14	7±6	*9±10*
Fatigued	67±25	68±18	*68±22*	-75±37	-74±26	*-75±32*	15±15	13±9	*14±12*
									#
Group	*69±29*	*68±16*		*-71±29*	*-70±23*		*13±15*	*10±8*	

Peak Moments (N-m)

	Mx			My			Mz*		
Fresh	-4±4	-3±1	*-4±3*	-3±2	-3±2	*-3±2*	-22±9	-20±5	*-21±7*
Fatigued	-3±4	-3±3	*-3±4*	-2±2	-2±2	*-2±2*	-21±18	-21±5	*-21±7*
Group	*-4±4*	*-3±2*		*-3±2*	*-3±2*		*-22±9*	*-21±5*	

FG=trunk flexion; NFG=non-trunk flexion; # significantly different main effect (p<0.05); scores are means±SD; *=propulsive moment.

Table 8.
Propulsion temporal characteristics.

	Stroke Frequency		Contact (sec)		Contact (% cycle)	
	FG	NFG	FG	NFG	FG	NFG
Fresh	1.1±0.3	1.2±0.2	0.49±0.08	0.41±0.11*	41.7±12.2	39.2±7.9
Fatigued	1.2±0.2	1.3±0.2	0.45±0.08	0.45±0.07	40.9±6.5	46.4±4.7 *

Stroke frequency in cycles per second; FG=trunk flexion; NFG=non-trunk flexion; *=significantly different interaction (p<0.05); scores are means±SD.
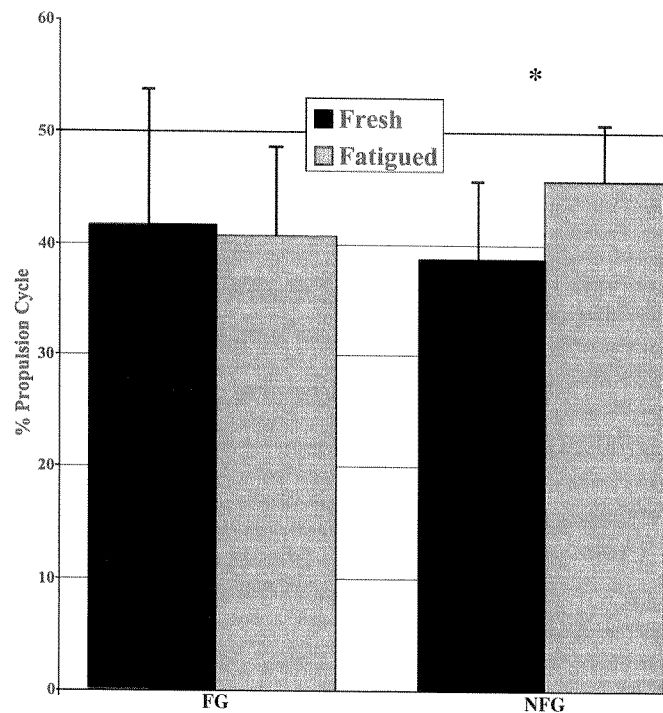
Figure 5.
Significant inter-group differences (mean±sd) in handrim contact time with fatigue. In contrast to the non-flexion group (NFG), the flexion group (FG) demonstrated decreased contact time with fatigue.
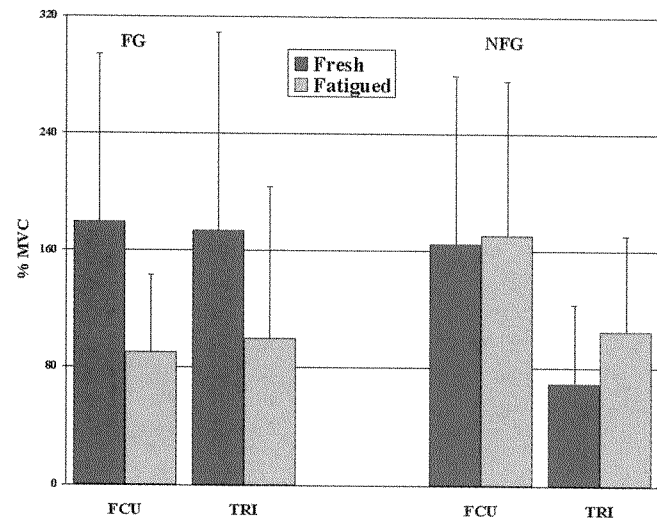

Figure 6.
Mean (±sd) EMG peak magnitude of the flexor carpi ulnaris (FCU) and triceps (TRI) muscles at the fresh and fatigued states for both groups. EMG peak magnitude is normalized to maximum isometric voluntary contraction (percent MVC). Note decreases in EMG activity with fatigue in the FG not evident in the NFG that were statistically significant.

propulsion similarities for other characteristics of MWCU with high and low paraplegia (26,35), suggesting that the difference in trunk control did not affect wheelchair propulsion ability. The degree of volitional trunk control was not measured for this study. Trunk control was classified by the level and type of injury, and from the medical screening examination. Information regarding the extent to which subjects are able to lean forward in the wheelchair and return to the starting position would be valuable in understanding the relationship between trunk control and propulsion style.

Fresh Versus Fatigued State

The trunk flexion movement pattern appeared to have been compensatory for muscle fatigue, since it was more pronounced in the fatigued state. The inter-group similarity in load, test duration, and VO_2 supported the premise that the trunk flexion style of propulsion was adopted to compensate for muscle fatigue rather than for aerobic economy. The increase in inward handrim force (F_z) may be a compensation for the fatiguing of the muscles used in grasping the handrim. Dallmeijer et al. (26) found that subjects with cervical level spinal cord injuries

(C4-C8) who lacked grasping ability used the inward handrim force for additional friction during propulsion.

The fatigue model was used in this study in an attempt to simulate the conditions that have been suggested to contribute to overuse injuries. It is clear that repetitive activities are associated with increased risk of joint degenerative changes. These changes may result from damage to the static and dynamic controlling mechanisms of joints (ligaments, joint capsules, menisci, and muscles). Work by Pugh (36) has shown that muscular control of joint movement is directly related to shock absorption. Based on this work, the fatigued muscle would appear to be an inadequate shock absorber, contributing to the probability of joint degradation. The changes in muscle activity patterns and biomechanical variables with fatigue that we have observed in our studies seem to support the use of the fatigue model to identify those MWCU most susceptible to overuse injuries. Since subjects had to be pain-free to participate in this study, experience in the wheelchair could not be correlated with current overuse injury. The only way to directly relate our measurements to overuse injury would be to follow our subjects over time to see what characteristics were consistent in those who developed overuse injuries. Longitudinal studies are very difficult, but are definitely warranted.

Interactions

Differences in the magnitude and timing of muscle activity with fatigue were observed according to propulsion style. During local muscle fatigue, EMG amplitude is known to increase due to increased motor unit recruitment, among other factors (37). In the current study, EMG peak amplitude of muscles critical to propulsion increased in the NFG from the fresh to fatigued state. However, there was a decline in peak amplitude of all muscles tested in FG, particularly in the flexor carpi ulnaris and triceps muscles. This decline may be interpreted as the recruitment of fewer motor units in upper limb muscles tested at the point of exhaustion. A fatigue-mediated reduction in motor unit recruitment appears to have been enabled by the employment of a compensatory movement strategy in the FG (i.e., trunk flexion, to generate propulsion moment). Such a strategy may not have been necessary for the NFG, who did not demonstrate the degree of fatigue shown by FG, suggesting better fitness of key propulsion muscles in the NFG. Extremely large skeletal muscle fiber areas in the triceps muscles of wheelchair athletes have been reported and attributed to hypertrophy from constant use during wheeling and other daily activi-

ties (38). Dynamic muscular strength of the triceps has been identified as an important determinant of handrim impulse generation (39), highlighting the important role of triceps activity during wheelchair propulsion and supporting the notion that triceps fatigue may trigger compensatory propulsion strategies.

The muscular fatigue demonstrated by the FG may place them at higher risk for injury. Muscle damage, as evidenced by elevated plasma creatine kinase, lactate dehydrogenase, and myoglobin content, has been noted in able-bodied individuals following progressive, submaximal exercise on a wheelchair treadmill (40). The reduction in contact time that accompanies fatigue may also place those with the trunk flexion style of propulsion at greater risk for injury. Faster application of handrim forces, and therefore joint forces, would be expected with decreases in contact time. Rapid loading of the arms during propulsion has been associated with certain overuse injuries among MWCU, such as carpal tunnel syndrome (17). Conversely, longer contact times and lower peak forces applied to the handrim are characteristic of experienced MWCU, versus inexperienced non-wheelchair users, thereby lessening potentially harmful joint forces (16). The longer contact times occurring with fatigue in the NFG may be a benefit of their propulsion style in terms of injury risk reduction. Further investigation of propulsion pathomechanics may identify additional factors contributing to injury. Further research is necessary to determine if overuse injuries can be prevented by strength and endurance training of muscles critical to propulsion (e.g., triceps) in those who demonstrate the trunk flexion propulsion style, since they appear to be compensating for peripheral muscle fatigue.

CONCLUSIONS

A trunk flexion style of wheelchair propulsion did not result in increased VO_2 compared to a non-trunk flexion style of propulsion. However, the differences observed for kinematics, kinetics, muscle activation patterns, and propulsion temporal characteristics supported our hypothesis that a trunk flexion style of wheelchair propulsion may, in fact, lead to painful upper limb injury which limits physical performance and quality of life. Definitive studies on pathomechanical mechanisms of upper limb injuries in MWCU may support these findings and facilitate interventions for the resulting impairment and disability. Implications from the current study are

that strength and endurance training of the muscles critical to propulsion (e.g., triceps) may be important in MWCU who demonstrate the trunk flexion style, since they appear to be compensating for peripheral muscle fatigue.

ACKNOWLDEGEMENTS

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