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The effect of seat position on wheelchair propulsion biomechanics

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Abstract—This study examined the effect of seat position on handrim biomechanics. Thirteen experienced users propelled a wheelchair over a smooth level floor at a self-selected speed. Kinetic and temporal-distance data were collected with the use of an instrumented rim and a motion analysis system. A custom-designed axle was used to change the seat position. We used repeated measures analysis of variance to evaluate if differences existed in the temporal-distance and kinetic data with change in seat position. Results showed that a shorter distance between the axle and shoulder (low seat height) improved the push time and push angle temporal variables ($p < 0.0001$). Tangential force output did not change with seat position. Axial and radial forces were highest in the lowest seat position ($p < 0.001$). Propulsion efficiency as measured by the fraction of effective force did not significantly change with seat position.

Key words: biomechanics, forces, moments, propulsion, seat position, wheelchair.

INTRODUCTION

Upper-limb pain and dysfunction are common among people who use manual wheelchairs for mobility. For example, surveys involving as many as 450 wheelchair-based individuals find that as many as 73 percent report some degree of chronic upper-limb pain, which they attribute primarily to wheelchair propulsion and transfers [1]. Individuals with paraplegia are particularly active wheelchair users and, when questioned, report prevalence of shoulder, elbow, and wrist/hand pain between 30 percent and 65 percent, with the shoulder the most common site of involvement [1–10].

These issues have not gone unnoticed, and the biomechanical analysis of manual wheelchair propulsion has become increasingly common over the last decade. The aim of this past research has been typically twofold: first, to improve propulsion mechanical efficiency (ratio of power output to oxygen cost), and second, to gain better understanding of the wheelchair-user interface to address the musculoskeletal problems associated with wheelchair use. A number of studies have examined upper-limb kinematics [11–19], forces exerted on the wheelchair handrim [20–23], and muscle activity during propulsion [24–28]. Researchers have investigated variables such as handrim size, wheel camber, rim tube diameter, and seat position to determine how changes in wheelchair configuration affect the energy cost and mechanical efficiency of wheelchair propulsion [24,27,29–41].

Abbreviations: EMG = electromyographic, FEF = fraction of effective force, GCVSPL = generalized cross-validation spline smoothing routine.

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Seat position, in particular, has been studied for almost 30 years. Engel et al., for example, performed a comparative study of three different seat positions on a fixed wheelchair simulator with lever propulsion in the 1970s and found that the most efficient configuration involved posterior placement of the seat unit relative to the rear wheels [42]. Brubaker also noted that propulsion efficiency is significantly affected by the user's position relative to the axle and advocated a posterior seat position to decrease rolling resistance and increase propulsion efficiency [40]. Van der Woude et al. studied the effect of seat height on energy consumption and kinematics in nine nonwheelchair subjects on a motor-driven wheelchair treadmill [30]. These investigators found that changes in seat height had significant effects on a wide range of variables, including gross mechanical efficiency, oxygen cost, hand range of motion during pushing ("push range"), push phase duration, and arm and trunk motion. These authors concluded that an elbow flexion angle of 60° to 80° when the subject was sitting at rest with the hand resting on the top of the wheel was ideal for daily use and basketball wheelchairs. Hughes et al. tested the effect of seat position on trunk and upper-limb kinematics using both handrim and lever propulsion in nine able-bodied and six spinal cord injured persons using a wheelchair-simulator data-acquisition system [29]. The data supported a strong relationship between seat position and joint motion during propulsion for only handrim propulsion. These investigators concluded that an ergonomically optimal seat position exists for each user.

Masse et al. collected kinematic and electromyographic (EMG) activity from five men with paraplegia in six different seat positions using a racing wheelchair and rollers [24]. Three horizontal and two vertical seating height conditions were studied, and an efficient wheelchair pushing pattern was defined as one in which EMG activity and pushing frequency were low and the recovery phase was prolonged. The investigators found that the lower seat positions were the most efficient in that they corresponded with lower EMG activity and pushing frequency, as well as smoother upper-limb motion. Specifically, the backward-low position was associated with the lowest overall level of EMG activity and middle-low with the lowest pushing frequency. Boninger et al. studied 40 individuals with paraplegia on a dynamometer as they accelerated from a dead stop to maximum speed as well as at two different steady-state speeds [41]. Subjects used their own wheelchairs in their normal configurations. Biomechanical variables included position of the axle relative to the shoulder, propulsion frequency, peak and rate of rise of the resultant forces and moment not acting in the plane of the wheel, start angle, and push range. These investigators found that an increase in the vertical distance between the shoulder and axle was associated with a decrease in the push angle. Similarly, positions with the axle forward relative to the shoulder were associated with increased push angles, lower propulsion frequencies, and slower rise of the propulsive forces.

In summary, shorter vertical distances between the axle and the shoulder and a more forward axle position have been correlated with improvements in wheelchair propulsion biomechanics. Richter investigated the relationship between seat position and propulsion biomechanics with a quasistatic two-dimensional wheelchair propulsion model [37]. Model inputs were length of the user's arms, shoulder position, handrim size, and handrim force profile. Output variables included joint kinematics, push angle, push frequency, and joint torques. Decreases in the axle to shoulder distances were associated with increased push angles, decreased push frequency, decreased shoulder torque, and increased elbow extension torque. The authors suggested that future research investigating the role of seat position on propulsion biomechanics should include both kinematics and kinetics of the upper limb.

This research provides a good basic understanding of the relationship between wheelchair configuration and propulsion. This study furthered our knowledge of the wheelchair-user interface by refining our understanding of the effect of seat position on handrim biomechanics. We hypothesized that lower seat heights and posterior seat positions would be associated with a more efficient stroke as defined by decreased stroke frequency, increased push angle, prolongation of the push phase, higher fraction of effective force (FEF), and a larger propulsion moment. Our study is unique in that subjects did not use a treadmill or dynamometer to simulate propulsion but rather propelled over a smooth floor in a motion analysis laboratory. While we do not know of research that determines if this would be a factor in affecting the variables we studied, it is interesting that gait research has demonstrated some discrepancies with regard to EMG activity and ground reaction forces when treadmill walking is compared to overground walking [43–45]. Therefore, the current study may provide a more realistic reflection of the actual conditions for wheelchair propulsion than is currently available in the literature. Also, we know of no other studies that have examined both timing and force application variables as they relate to change in seat position.

METHODS

Subjects

The Mayo Institutional Review Board reviewed the protocol. Following approval, wheelchair users were identified from our rehabilitation unit patient database and sent a letter describing the study and requesting their participation. Additional recruitment was accomplished via word of mouth and advertisements posted on bulletin boards in local accessible housing units. Inclusion criteria were use of a manual wheelchair as the primary mode of mobility, an injury level between T5 and L3, and at least 6 months experience in wheelchair propulsion. Exclusion criteria were current or chronic upper-limb pain, a history of a significant upper-limb injury, involvement in competitive sports with a specific training program, or employment in an occupation that required repetitive use of the upper limbs in an elevated position. All subjects signed a consent form prior to data collection. We interviewed each subject to familiarize them with the protocol and to determine if the inclusion and exclusion criteria were satisfied. Kinematic and kinetic data were collected in a motion analysis laboratory. A description of the data collection system follows.

Kinematic System

A 10-camera Expertvision™ System (Motion Analysis Corporation, Santa Rosa, CA) was used to collect the 3D trajectory data of markers placed on the left upper limb and wheel at a sampling frequency of 60 Hz. Thirteen 25 mm reflective markers were placed at the following bony landmarks: lateral side of the 2nd metacarpal head, medial side of the 5th metacarpal head, radial and ulnar styloid processes, medial and lateral epicondyles of the elbow, acromion process, spinous process of seventh cervical vertebrae, sternal notch, and xiphoid process. We fixed three additional markers mounted to a rigid frame to the humerus to define a local coordinate system aligned with the long axis of the humerus (**Figure 1**). To permit monitoring of its location and orientation, we mounted five additional markers on the wheelchair wheel. The motion capture system cameras were adjusted so that each marker could be seen by at least two cameras throughout

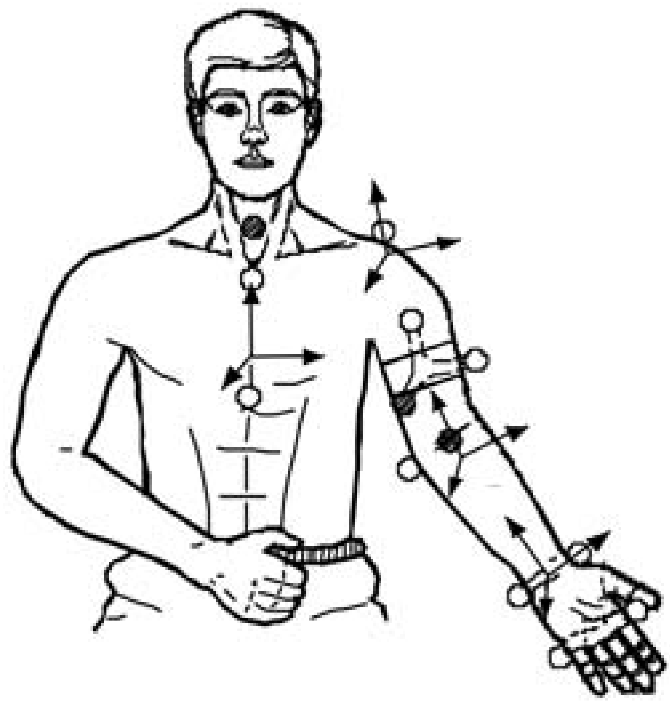

Figure 1.
Locations of 13 reflective markers applied to upper limb and orientations of trunk, upper arm, forearm, and hand coordinate systems.

at least one full propulsion cycle in a 5.0 m long × 3.0 m wide × 2.0 m high view volume that was calibrated so that the 3D coordinates of each marker could be collected in the laboratory coordinate system.

Kinetic System

A wheelchair wheel and hand rim instrumented with a six-component load cell (model 45E15A-MAYO1, JR3, Inc., Woodland, CA), developed and validated in this laboratory, was used to collect kinetic data [46]. The hand rim was mounted to one side of the load cell, and the other side of the load cell was mounted directly to the wheel such that the *x-y* plane of the transducer was oriented in the plane of the wheel. Load cell output voltages were recorded at a sampling frequency of 100 Hz and stored in a miniature data logger mounted on the wheel. Data were transferred to a personal computer that converted output voltages to force (N) and moments (Nm), and corrected crosstalk between the load cell channels at the conclusion of each trial. Baseline calibration data were collected prior to each data collection session. Data from the motion analysis system and the data logger were synchronized with an external trigger. A trial in which the subject had at

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least three complete propulsion cycles (push and recovery) within the view volume was considered analyzable.

Adjustable Wheelchair

A Quickie II ultra-lightweight sport wheelchair (Sunrise Medical, Longmont, CO) with 24 in. (61.0 cm) diameter rear wheels, pneumatic inner tubes, 20 in. (50.8 cm) diameter hand rims, 8 in. (20.3 cm) polyurethane front wheels, 17 in. (43.2 cm) seat width, 16 in. (40.6) seat depth, and 0° camber angle was used for all subjects. The side frames of the chair containing the axles were modified with a custom-designed aluminum mechanism fixed to the frame of chair, which permitted 8 cm changes in the axle's horizontal position and 10 cm changes in its vertical position with the turn of a knob (**Figure 2**). Nine different axle horizontal and vertical positions were studied. Moving the axle forward results in a seat unit that is posterior relative to the rear wheels, and moving the axle backward results in a seat unit that is anterior to the wheels. These positions were studied for all subjects and were chosen because they encompass the range of axle settings in a Quickie II wheelchair, which is commonly prescribed in our rehabilitation clinic (**Figure 3**). The backrest and footrest heights of the Quickie II chair were adjusted so that they were similar to the subject's personal chair.

Data Collection

Data were collected in a motion analysis laboratory as the subjects propelled the instrumented wheelchair over a 20 m section of smooth level tile floor. Prior to data collection, the upper-limb marker set described previously was applied to the left upper limb. The instrumented wheel was installed onto the left axle, and an equally weighted wheel was used on the right side to maintain chair balance. A matrix of seat positions consisting of four different vertical positions and one to three different horizontal positions defined the nine different experimental seat positions (**Figure 3**). The lowest axle position that we tested corresponded to a horizontal seat, as measured by an inclinometer. The highest axle position placed the seat unit in an anterior tilt angle of 10° relative to the horizontal. The order in which the seat positions were tested was randomized for each subject. Subjects were instructed to propel the wheelchair using only the handrims, without coasting, at a self-selected comfortable speed. Data were collected for approximately 5 s in the calibrated view volume, which permitted the capture of a minimum of three strokes for each seat condition and then downloaded from the data logger into a computer. Following each trial, a static data collection was obtained with the hand gripping the top dead center of the handrim. We did this to quantify the seat position by defining the elbow angle. The seat was adjusted to the next

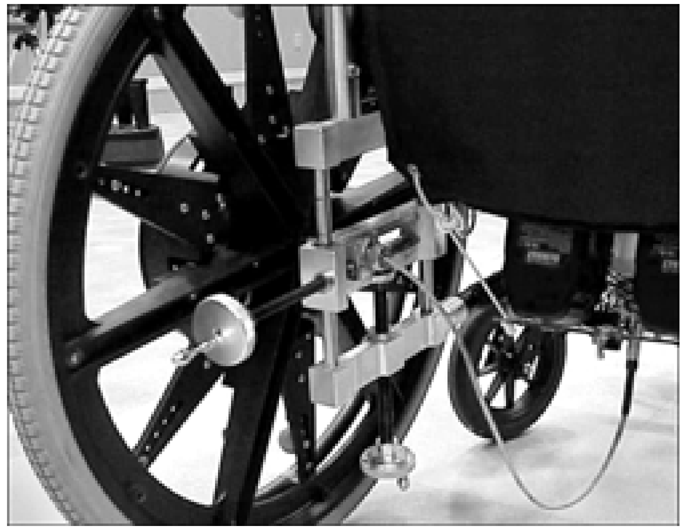

Figure 2.
Quick-adjust wheelchair axle. Position of axle changed in vertical and horizontal directions with turn of knob.

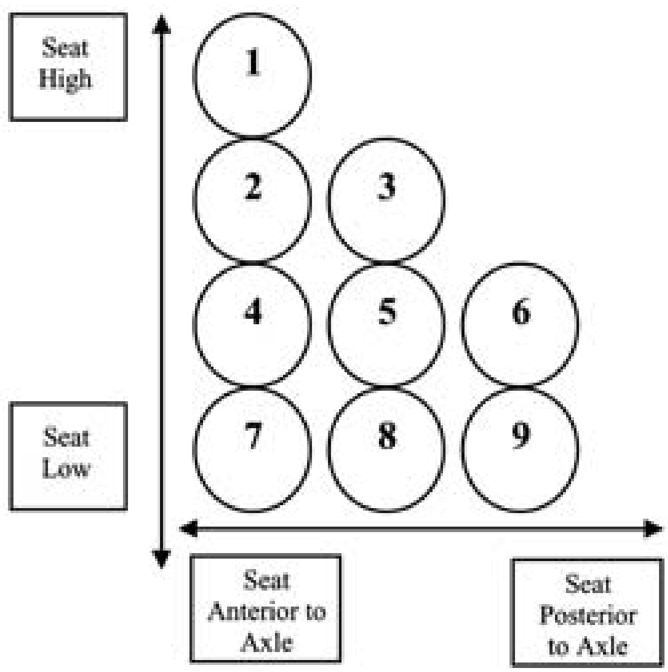

Figure 3.
Seat position matrix consisting of nine different positions.

randomly selected position and the procedure was repeated, with the subjects instructed to attempt to propel themselves at the same speed as in earlier trials. Data downloading, static position measurement, and seat changes required approximately 10 min.

Data Analysis and Reduction

As just noted, a minimum of three propulsion strokes were collected for each seat position condition. We verified consistency between strokes by examining the kinematic and kinetic data, and we identified and used a single representative stroke for analysis. The 3D marker coordinate positions were smoothed with the use of a generalized cross-validation spline smoothing routine (GCVSPL) with a cutoff frequency of 6 Hz [47]. Load cell data were filtered with the use of the GCVSPL routine at a cutoff frequency of 18 Hz, as determined by residual analysis [48]. All additional calculations were performed with custom routines written in MATLAB™ (The MathWorks Inc., Natick, MA), which have been validated previously [46]. Each analyzable stroke was normalized to percentage propulsion cycle for subsequent data analysis. The beginning of the stroke cycle was defined as the instant at which any of the three hand-rim force components became positive (after having been zero during the recovery phase).

Statistical Analysis

Statistical analyses were performed with SAS (SAS Institute, Cary, NC). An analysis of variance for repeated measurements with seat position as the main factor was applied to detect significant differences for the biomechanical variables. Post hoc analyses tests consisted of the Student-Newman-Keuls multiple range test for all main-effect means and the least-squares estimates of marginal means. Significance level was set at $p < 0.05$ for all statistical procedures.

RESULTS

Subject Characteristics

Thirteen wheelchair users were recruited, ten males and three females. Demographic characteristics are presented in **Table 1**.

Seat Position

Static elbow angles taken with the hand gripping top dead center of the handrim (0°—full elbow extension) in the different vertical and horizontal axle positions are shown in **Figure 4**. Elbow angle did not significantly change with movement of the seat's horizontal position (i.e., positions 2–3, 4–6, 7–9). On the other hand, vertical displacements of the seat (positions 1, 2, 4, and 7) resulted in statistically significant differences between the elbow angles ($p < 0.0001$).

Handrim Biomechanics: Effect of Seat Position

Group means and standard deviations for the temporal-distance parameters and hand-rim variables are shown in **Tables 2** and **3**. The effects of seat position on the temporal-distance parameters are listed in **Figure 5**. Seat position changes had no effect on chair speed, stroke frequency, stroke time, and stroke distance. The coefficient of variation for chair velocity was 5.1 percent within subject and 11.4 percent between subjects. With regard to push time, significant differences were observed between seat positions 1–3 (0.29–0.31 s) and positions 8–9 (0.37–0.38 s). Recovery time also showed

Table 1.
Subject characteristics.

Characteristics	Users $N = 13$ 10 Male, 3 Female	Nonusers $N = 20$ 10 Male, 10 Female	p-value*
Age (yr)	38.3 ± 10.3	27.0 ± 5.8	0.002
Weight (kg)	80.7 ± 18.9	74.2 ± 16.3	0.32
Height (cm)	176.5 ± 7.7	171.9 ± 9.5	0.14
Time Since Injury (yr)	11.5 ± 5.8	—	—
Level of Injury	T11 to L2	—	—

Note: Data are reported as mean ± standard deviation.
*Unpaired t-test

a significant difference between positions 2–3 (0.69–0.73 s) and positions 4–9 (0.61–0.65 s). As shown in **Figure 6**, the push angle (excursion of the hand on the pushrim) was significantly higher in seat positions 7–9 (80°–86°) compared to positions 1–6 (69°–78°).

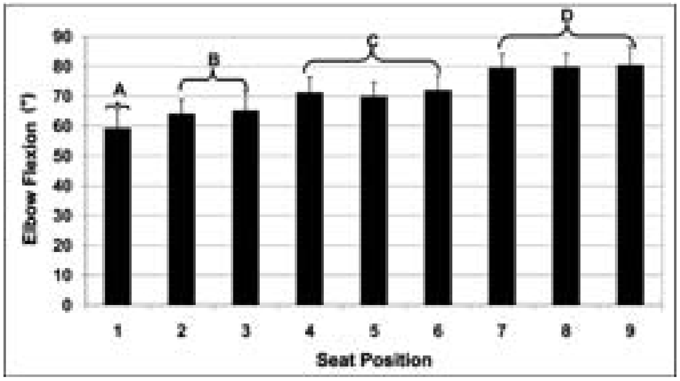

Figure 4.
Effect of seat position on elbow flexion angle, measured with hand gripping top dead center of hand rim; 0º equals full elbow extension. No difference between angles within groups A–D; significant difference in angles between groups A–D.

Peak radial forces were significantly higher in positions 8 (45.1 N) and 9 (42.9 N) than in the other seven seat positions, which had values that ranged from 27.2 to 36.9 N. Tangential force values did not significantly change with seat position. Axial force values were significantly higher in seat positions 8 and 9 (18.3–22.9 N) when compared to the other positions (**Figure 7**). Mean FEF and the peak propulsion moment did not vary with change in seat position (**Figures 8** and **9**).

DISCUSSION

This study determined whether changes in axle position that are usually made in the clinical setting to a Quickie II manual wheelchair would result in a significant change in timing variables or hand-rim force application. Our results demonstrate that the only timing variables that changed were push time, recovery time, and push angle. The findings that push time and push angle were highest in the low seat height positions confirm the results of

Table 2.
Temporal-distance parameters.

Variable	Users Mean (Standard Deviation)	Nonusers Mean (Standard Deviation)	*p*-value
Mean Chair Speed (m/s)	1.48 (0.16)	1.55 (0.16)	0.23
Stroke Time (s)	1.08 (0.18)	0.71 (0.11)	<0.001
Stroke Frequency (cycles/s)	1.23 (0.22)	1.49 (0.21)	0.003
Stroke Distance (m)	1.42 (0.24)	1.16 (0.18)	0.001
Push Time (s)	0.34 (0.04)	0.24 (0.02)	<0.001
Recovery Time (s)	0.66 (0.11)	0.49 (0.09)	<0.001

Table 3.
Hand-rim biomechanical variables.

Variable	Users Mean (Standard Deviation)	Nonusers Mean (Standard Deviation)	*p*-value
Grab Angle (°)	121.71 (10.1)	100.57 (8.5)	<0.0001
Release Angle (°)	44.67 (5.3)	39.46 (5.7)	0.02
Push Angle (°)	77.03 (10.21)	62.68 (10.24)	<0.001
Peak Radial Force (N)	34.4 (14.1)	43.1 (15.8)	0.04
Peak Tangential Force (N)	31.9 (14.9)	53.4 (17.3)	<0.0001
Peak Axial Force (N)	14.9 (6.8)	14.0 (8.6)	0.71
Mean fraction of effective force (Ft/Fn)	0.55 (0.06)	0.64 (0.07)	0.001
Peak Propulsion Moment (Nm)	8.30 (3.4)	12.3 (3.6)	0.004

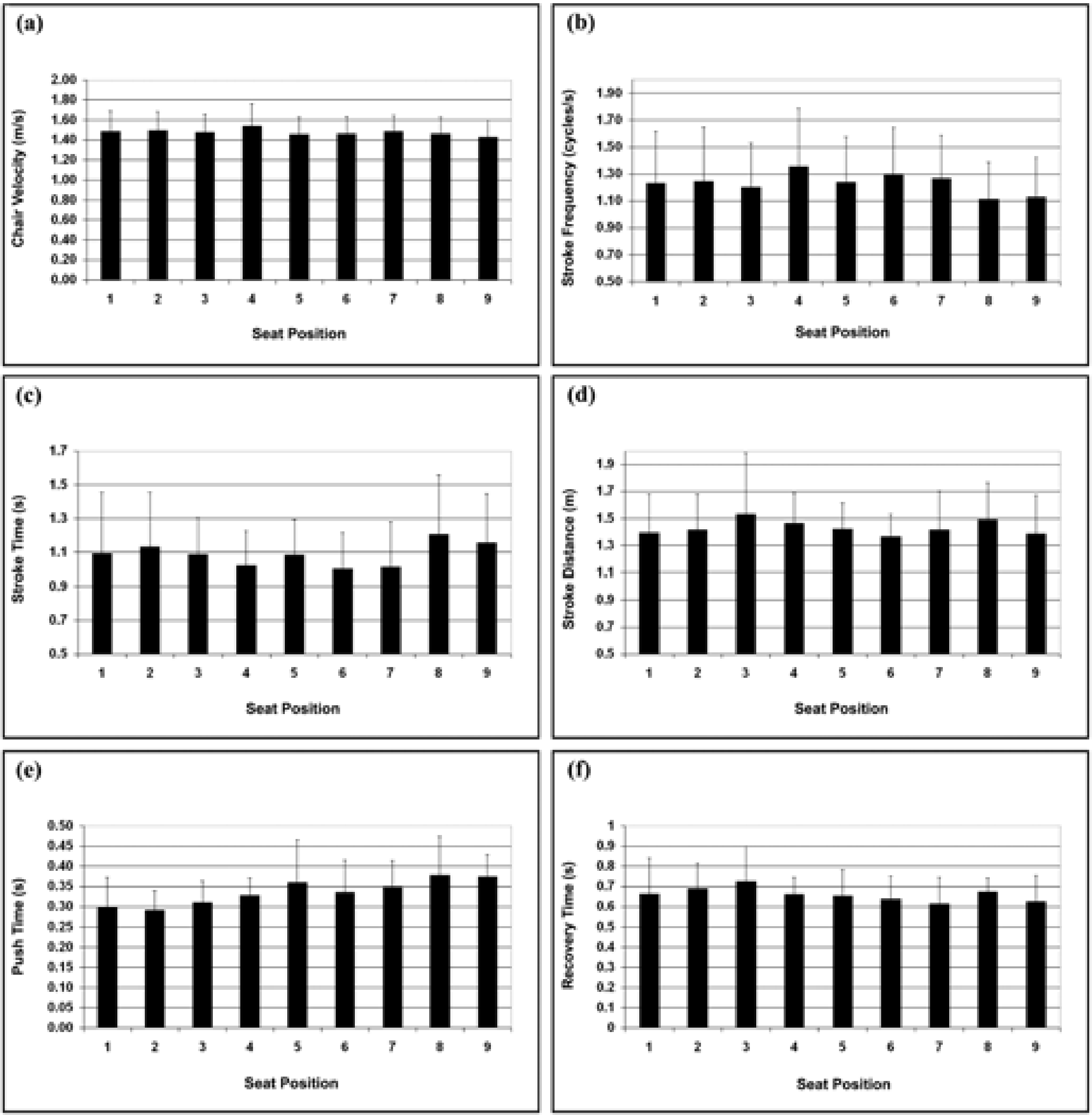

Figure 5.
Effect of seat position on temporal-distance parameters: **(a)** No difference in speed with change in seat position. **(b)** No difference in stroke frequency with change in seat position. **(c)** No difference in stroke cycle time with change in seat position. **(d)** No difference in stroke distance with change in seat position. **(e)** Significant difference between positions 1 and 5, 1–8, 1–9, 2–5, 2–7, 2–8, 2–9, 3–8, 3–9. **(f)** Significant difference between positions 2 and 7, 2–9, 3–4, 3–5, 3–6, 3–7, 3–9.

previous research by Maase et al., van der Woude et al., and Boninger et al. [24,30,41]. Decreasing the vertical distance between the axle and shoulder makes more of the handrim available for the push cycle, thus increasing the

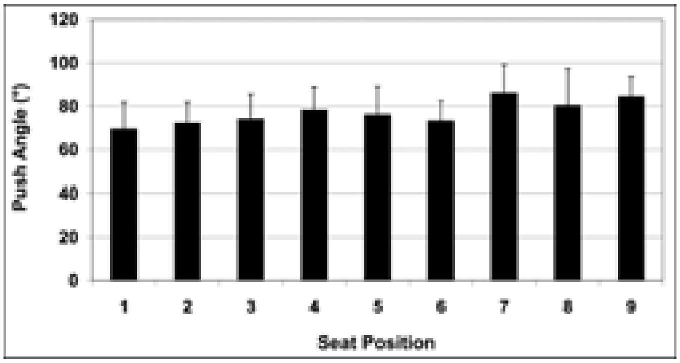

Figure 6.
Effect of seat position on excursion of hand on hand rim. Significant difference between positions 1and 4, 1–7, 1–8, 1–9, 2–7, 2–8, 2–9, 3–7, 3–8, 3–9, 4–1, 4–7, 4–9, 5–7, 5–9, 6–7, 6–8, 6–9, 7–8.

push angle, and contact time between hand and rim. This increased push angle was also shown in the Maase study to result in a reduction of stroke frequency for a constant speed due to force being applied over a longer period of time. In contrast, we did not find a significantly lower stroke frequency for the lower seat height positions as found in the Masse study. There was a trend for decreased stroke frequency in seat positions 8 and 9, but no statistical significance was demonstrated (p = 0. 21 and 0.25).

With regard to peak hand-rim force variables, we found no difference in tangential force output between the seat positions, but statistically higher peak axial and radial forces in the lowest seat positions (8 and 9) compared to the others. We know of no other research that has specifically examined hand-rim force components with change in seat position, so it was difficult to formulate any hypothesis on how they would change. Somewhat surprising was that seat position did not have any effect on tangential force output. In accordance with this finding, the FEF also did not change with seat position. The fact that we did not observe any significant differences may be somewhat due to the limited range of seat height adjustment. Again, our intention was to replicate the range of changes that are commonly made in the clinical setting and not make changes that would jeopardize chair and occupant stability and therefore not be used in practice. Within this clinically relevant range, the propulsive component of the hand-rim force did not change with seat position.

An alternate explanation for this finding is the work of Rozendaal and Veegar [49], which contends that force direction on the hand rim is optimized, given the mechanical constraints of the musculoskeletal and wheelchair

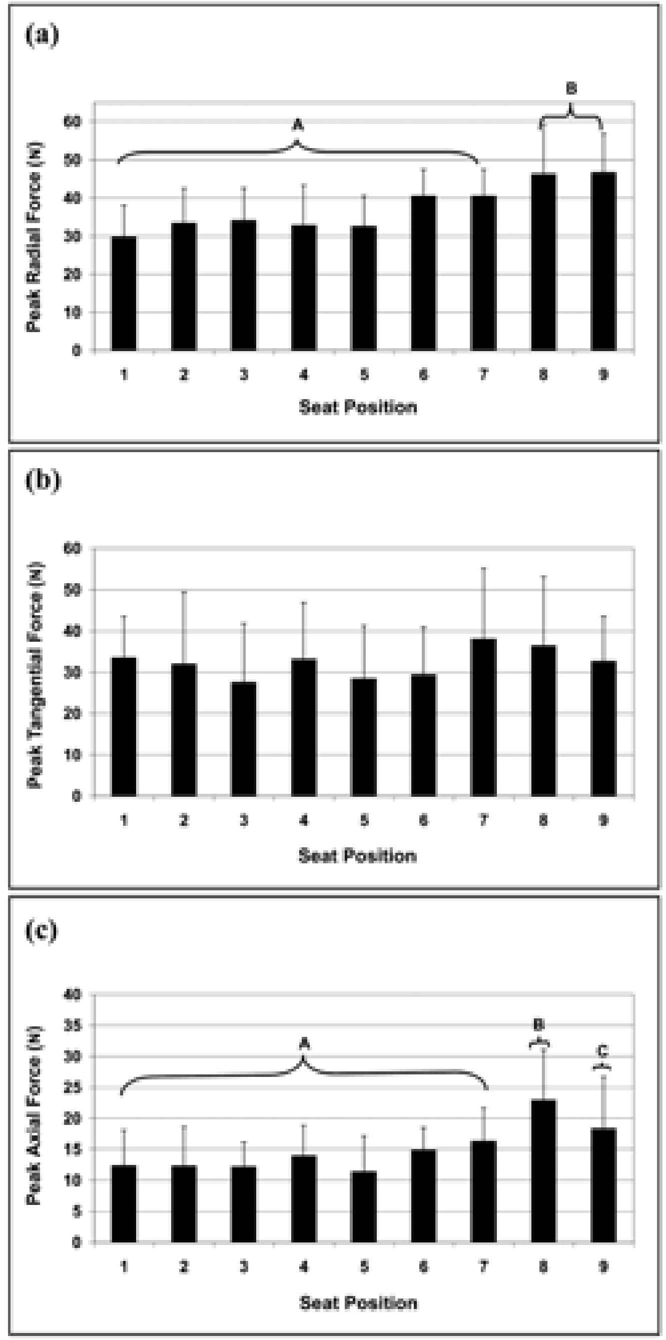

Figure 7.
Peak hand-rim forces as function of seat position: **(a)** Significant difference between peak radial force in positions in group A versus positions in group B; no difference within these two groups. **(b)** No difference in peak tangential force with change in seat position. **(c)** Significant difference between peak axial force in positions in groups A–C.

propulsion system. The findings from Rozendaal and Veegar's research suggest that we may not be able to sig-

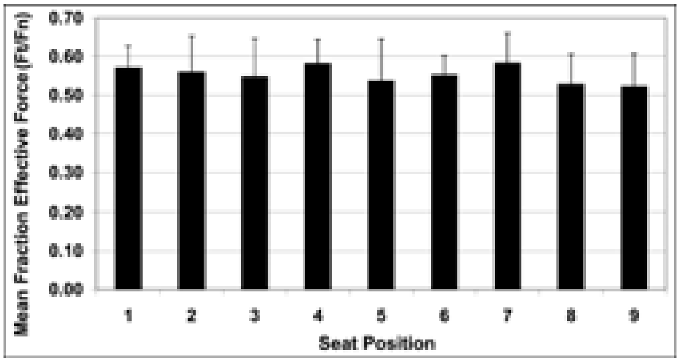

Figure 8.
Effect of seat position on mean fraction of effective force (FEF) (tangential force/total force). No difference in FEF in change in seat position.

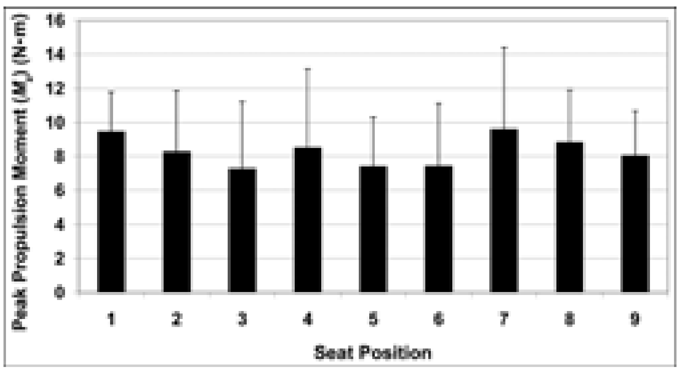

Figure 9.
Effect of seat position on peak propulsion moment measured at axle.

nificantly increase tangential force production and therefore the FEF without decreasing mechanical efficiency. For example, de Groot found that subjects were able to increase their FEF via visual feedback training but at the cost of a significantly lower mechanical efficiency (power output in relation to energy cost) [50]. This may be also true with change in seat position. The lower positions may positively affect the timing variables, but it will still take more energy to produce a higher tangential force output regardless of position, so subjects intuitively take the minimal energy path and do not significantly change their force on the rim. This does not mean that we should disregard proper chair fit. Clearly, some of the timing variables were positively affected by a lower seat height.

It was interesting to find that average peak radial and axial force components were significantly higher in the low seat positions. This implies that more force is being directed toward the axle and medially toward the chair occupant rather than toward propelling the wheel forward in the lowest seat height positions that we tested. This would seem to have a negative effect on propulsion efficiency. These increases may be due to the less than optimal position of the shoulder joint (increased shoulder abduction) in the lowest seat position. This suggests that although users may benefit from a lower seat height to improve their temporal variables, at some point the tradeoff is an increase in nonpropulsive forces if the seat is positioned too low for the subject's anthropometry. However, these peak radial, axial, and tangential forces all occur at different points during the push phase. When we compare the tangential force to total force at each point in the cycle, we can see from the FEF values that propulsion efficiency as defined by FEF does not change.

Finally, we found that the propulsion moment measured at the wheel hub did not change with seat position. Because the moment at the wheel is dependent on speed of propulsion and resistance and is directly related to tangential force output, this finding was not expected since these variables also did not change.

Our study contains a number of limitations. For example, wheelchair velocity was not strictly controlled and the subjects were only instructed to try to maintain a similar velocity between trials. This experimental setup permits variations in speed due to conditions as would be seen in the real world. However, it confounds the interpretation of some results, since fewer variables are controlled. In addition, our axle adjustment approach meant that the subjects all used the same wheelchair. This chair was similar to their own, but it may have affected their propulsion technique. Also, our findings are only specific to propulsion at a self-selected speed over a smooth level floor. Finally, we studied users with spinal cord injury. These users are good subjects in that their level of propulsive activities tends to be high; however, propulsion variables may be different for other conditions that result in reliance on a wheelchair for mobility.

CONCLUSION

Seat position changes had an effect on the timing variables of push time, recovery time, and push angle. Low seat positions resulted in an improvement in these variables. Tangential force production did not change with seat position. On the other hand, peak radial and axial pushrim forces were significantly higher in the lowest seat positions. The effectiveness of force application as defined by the FEF did not change with seat position. Further research is needed to determine whether the FEF is a valid indicator of propulsion efficiency, because the direction of force application is, in fact, already optimal given the constraints of the system.

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