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**A Theoretical Analysis of the Influence of Wheelchair
Seat Position on Upper Extremity Demand**

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by

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Dedication

This thesis is dedicated to my loving wife, Nicole, and our growing family.

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Abstract

A Theoretical Analysis of the Influence of Wheelchair Seat Position on Upper Extremity Demand

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The high demands of manual wheelchair propulsion put users at risk of additional pain and injury that can lead to further reductions in independence and quality of life. Seat position is an adjustable parameter that has been shown to influence propulsion biomechanics. As a result, a number of studies have attempted to optimize this position. However, due to complexities in quantifying upper extremity demand, seat position guidelines are often based on studies aimed at reducing indirect quantities (e.g., cadence, handrim forces, joint ranges of motion and muscle excitation levels) rather than more direct measures of demand (e.g., muscle stress and metabolic cost). Forward dynamics simulations provide an alternative approach to systematically investigate the influence of seat position on more direct measures of upper extremity demand. The objective of this study was to generate and analyze a set of forward dynamics simulations of wheelchair propulsion across the range of attainable seat positions to identify the optimal seat position that minimizes upper extremity demand (i.e., muscle stress, metabolic cost and muscle antagonism). The optimization results showed both metabolic cost and muscle

stresses were near minimal values at superior/inferior positions corresponding to top dead center elbow angles between 110 and 120 degrees while at an anterior/posterior position with a hub-shoulder angle between -10 and -2.5 degrees. These minimal values coincided with a reduction in the level of antagonistic muscle activity, primarily at the glenohumeral joint. Seat positions that deviated from these minimal values increased the level of co-contraction required to maintain a stable, smooth propulsive stroke, and consequentially increased upper extremity demand. These results can provide guidelines for positioning the seat to help reduce upper extremity overuse injuries and pain, and thus improve the overall quality of life for wheelchair users.

Table of Contents

List of Tables	ix
List of Figures	x
INTRODUCTION	1
METHODS	4
Overview.....	4
Musculoskeletal Model.....	4
Manual Wheelchair Model	7
Simulation of Wheelchair Propulsion.....	10
Simulation Analysis.....	11
RESULTS	13
The Influence of Seat Position on Metabolic Cost	14
The Influence of Seat Position on Muscle Stress	15
The Influence of Seat Position on Muscle Antagonism	15
DISCUSSION	19
Limitations & Future Work	21
CONCLUSIONS.....	24
Appendix.....	25
References.....	31
Vita.....	38

List of Tables

Table 1:	Average differences between initial position simulation kinematics and group-averaged experimental kinematics of 12 experienced wheelchair users	14
Table 2:	Individual DOF excess torque values	21
Table A1:	Upper extremity musculoskeletal model parameters	25

List of Figures

Figure 1:	3D musculoskeletal model used in the wheelchair propulsion simulations	6
Figure 2:	2D model of wheelchair propulsion used to determine contact and release angles	8
Figure 3:	Initial position simulation hand path	13
Figure 4:	Normalized metabolic cost versus seat position	16
Figure 5:	Overall muscle stress versus seat position	17
Figure 6:	Average excess torque versus seat position	18
Figure A1:	Plane of elevation excess torque versus seat position	26
Figure A2:	Elevation angle excess torque versus seat position	27
Figure A3:	Internal-external rotation excess torque versus seat position	28
Figure A4:	Elbow flexion-extension excess torque versus seat position	29
Figure A5:	Forearm pronation-supination excess torque versus seat position	30

INTRODUCTION

There are approximately 3.3 million wheelchair users in the United States (CDC, 2009), with the vast majority (>90%) of users relying on manual wheelchair propulsion as their primary method of mobility (Kaye et al., 2000). Upper extremity pain and injuries that frequently occur in wheelchair users (e.g., Boninger et al., 2005; Mercer et al., 2006) can be very debilitating and lead to a decrease in independence and quality of life (e.g., Pentland and Twomey, 1994; Gutierrez et al., 2007; Requejo et al., 2008a; Riggins et al., 2011). The high incidence of pain and injury is correlated with the high physical demand placed on the upper extremity during wheelchair propulsion (e.g., Boninger et al., 2005; Requejo et al., 2008a). In addition to generating the mechanical power required to propel the wheelchair, the upper extremity muscles must also help maintain joint stability (e.g., Requejo et al., 2008a; Requejo et al., 2008b). These stability requirements, along with the kinematic constraints of the push phase, require a large amount of intermuscular coordination and co-activation (e.g., van der Helm and Veeger, 1996; Rankin et al., 2010; Rankin et al., 2011; Rankin et al., 2012). Notably, the glenohumeral joint has relatively few stabilizing structures (PVACSCM, 2005), requiring the muscles responsible for stabilizing the joint to be highly active and have an elevated risk of injury (e.g., Veeger et al., 2002; Mulroy et al., 2004).

Seat position is an easily adjustable parameter that directly influences propulsion biomechanics (e.g., Boninger et al., 2000; Richter, 2001; Kotajarvi et al., 2004; Gorce and Louis, 2012) and upper extremity demand (e.g., Boninger et al., 2005; Gutierrez et al., 2005; Mulroy et al., 2005; Requejo et al., 2008a). Thus, identifying the optimal seat position that minimizes upper extremity demand holds great promise for reducing the risk of pain and injury. A number of studies have examined the influence of seat position on

propulsion mechanics and found relationships with specific biomechanical measures such as cadence (e.g., Masse et al., 1992; Boninger et al., 2000; Richter, 2001; Kotajarvi et al., 2004; Gorce and Louis, 2012), handrim forces (e.g., Boninger et al., 2000; Kotajarvi et al., 2004; van der Woude et al., 2009), joint ranges of motion (e.g., Hughes et al., 1992; Wei et al., 2003; Gorce and Louis, 2012) and electromyography (EMG) activity (e.g., Masse et al., 1992; Gutierrez et al., 2005; Louis and Gorce, 2010). High levels of these measures have been identified as risk factors for upper extremity pain and injury (e.g., Boninger et al., 2005; Gutierrez et al., 2005; Gorce and Louis, 2012).

Recent clinical guidelines based on these relationships suggest that seat position should be adjusted as far posterior as possible without compromising user stability (PVACSCM, 2005). They also recommend superior/inferior positions that correspond to an elbow angle between 100° and 120° when the hand is at the top dead-center (TDC) position on the handrim (full extension is 180°). However, recent studies have found with such low, posterior seat positions, the joint ranges of motion and muscle activity levels may be increased (Louis and Gorce, 2010; Gorce and Louis, 2012) and adversely affect upper extremity demand.

One challenge of assessing the influence of seat position on upper extremity demand is the inability to directly measure demand-related quantities such as muscle stress or co-contraction that may elevate joint loading. Because these measures are difficult to obtain experimentally (Lin et al., 2004; Erdemir et al., 2007), indirect measures are frequently used. For example, inverse dynamics techniques are often used to determine joint torques, but identifying individual muscle force and stress is challenging due to muscle redundancy and co-contraction (Zajac et al., 2002, 2003; Erdemir et al., 2007). In addition, systematically investigating the influence of seat position on upper extremity demand using experimental techniques is difficult and time-

consuming (e.g., to assess the influence of seat position on metabolic cost). As a result, most studies have investigated a limited number of seat positions (e.g., two or three).

Forward dynamics simulations provide an alternative approach to systematically examine the influence of wheelchair seat position on direct measures of upper extremity demand. Forward dynamics techniques have been successfully used to analyze various human movement tasks including pedaling (e.g., Raasch et al., 1997; Umberger et al., 2006; Rankin and Neptune, 2008), walking (e.g., Anderson and Pandy, 2001; Liu et al., 2006; McGowan et al., 2009) and running (e.g., Sasaki and Neptune, 2006; Miller et al., 2012). More recently, these techniques have been applied to analyze wheelchair propulsion to identify individual muscle contributions to push and recovery mechanics (e.g., Rankin et al., 2010; Rankin et al., 2011; Rankin et al., 2012). Simulations can also be used to systematically vary system parameters and analyze their influence on specific biomechanical measures. This technique has recently been used in the design of bicycle configurations (e.g., Rankin and Neptune, 2010) and lower limb prostheses (e.g., Fey et al., 2012).

The goal of this study was to use forward dynamics simulations of wheelchair propulsion to investigate how seat position influences upper extremity demand (i.e., muscle stress, metabolic cost and muscle antagonism). Understanding these relationships can help guide clinicians in determining the optimal wheelchair configurations that reduce upper extremity demand, and ultimately overuse injuries and pain in wheelchair users.

METHODS

Overview

A 3D musculoskeletal model of the upper extremity was combined with a model of a standard wheelchair to create forward dynamics simulations of manual wheelchair propulsion. Seat position was systematically varied and its influence on muscle stress, metabolic cost and muscle antagonism was quantified. Below, the musculoskeletal and wheelchair models, simulation generation and analysis are described in detail.

Musculoskeletal Model

The musculoskeletal model used in this study was based on a previously described model of the upper extremity (Holzbaur et al., 2005; Rankin et al., 2010; Rankin et al., 2011; Rankin et al., 2012) and will be summarized here (Figure 1). The model was developed using SIMM (Musculographics, Inc., Santa Rosa, CA, USA) and consisted of a trunk and right side upper arm, forearm and hand segments. There were three degrees-of-freedom (DOFs) at the shoulder, defined as plane of elevation, elevation angle and internal-external rotation, in accordance with the recommendations of the International Society of Biomechanics (Wu et al., 2005). The motion at the shoulder also included a scapulohumeral rhythm based on regression equations from cadaver data (Holzbaur et al., 2005). The model had two additional DOFs, elbow flexion-extension and forearm pronation-supination. The wrist was fixed in the standard anatomical position.

Twenty-six Hill-type musculotendon actuators (Table A1 in the Appendix) represented the major upper extremity muscles crossing the shoulder and elbow joints. Each actuator received its own excitation pattern, with the exception of three muscle

groups: the two actuators that represented the sternocostal portion of the pectoralis major, the three latissimus dorsi actuators and the two actuators representing the lateral triceps and anconeus. Within each of these groups, the actuators received the same excitation pattern, resulting in a total of twenty-two independent excitation patterns. The actuators were governed by intrinsic muscle force-length-velocity relationships (Zajac, 1989). The muscle excitation-activation dynamics were modeled using a first-order differential equation (Raasch et al., 1997) with muscle-specific activation/deactivation time constants (Winters and Stark, 1988; Happee and van der Helm, 1995). Passive torques were applied at the shoulder and elbow to represent ligaments and other passive joint structures that limit extreme joint positions. The dynamic equations-of-motion were generated using SD/FAST (PTC, Needham, MA).

Figure 1: 3D musculoskeletal model used in the wheelchair propulsion simulations. The model had 5 degrees-of-freedom: plane of elevation, elevation angle, internal-external rotation, elbow flexion-extension, and forearm pronation-supination. Twenty-six Hill-type musculotendon actuators represented the major upper extremity muscles crossing the shoulder and elbow joints. These actuators were: DELT1 (anterior deltoid), DELT2 (middle deltoid), DELT3 (posterior deltoid), PECM1 (pectoralis major, clavicular head), PECM2 (pectoralis major, sternocostal head portion 1 - sternum), PECM3 (pectoralis major, sternocostal head portion 2 - ribs), CORB (coracobrachialis), TMAJ (teres major), LAT1 (latissimus dorsi, thoracic portion), LAT2 (latissimus dorsi, lumbar portion), LAT3 (latissimus dorsi, iliac portion), SUBSC (subscapularis), INFSP (infraspinatus), TMIN (teres minor), SUPSP (supraspinatus), BRD (brachioradialis), BRA (brachialis), BICshort (biceps brachii, short head), BICLong (biceps brachii, long head), ANC (anconeus), TRIlateral (triceps brachii, lateral head), TRImed (triceps brachii, medial head), TRIlong (triceps brachii, long head), SUP (supinator), PQ (pronator quadratus) and PT (pronator teres).

Manual Wheelchair Model

A model of a standard manual wheelchair with circular handrims (radius (R_{HR}) of 0.267m, rim-to-rim distance of 0.635m) was generated and combined with the musculoskeletal model by defining the position of the hip relative to the rear axle of the wheelchair. This position was kept fixed throughout each simulation of wheelchair propulsion and was systematically varied in both the superior/inferior and anterior/posterior directions to represent a wide range of seat positions. The motion of the third metacarpophalangeal joint was prescribed to follow a path on the circular handrim during the push phase, which was further defined by the contact (θ_C) and release (θ_R) angles. These angles were dependent on the seat position, and thus were calculated using a simple 2D model (Figure 2, based on Richter, 2001) as follows:

$$L_{HS} = \sqrt{x_{HS}^2 + y_{HS}^2} \quad (1)$$

$$\theta_{HS} = \tan^{-1} \left(\frac{x_{HS}}{y_{HS}} \right) \quad (2)$$

$$\theta_C = 0.9 * \left(\theta_{HS} - \cos^{-1} \left[\frac{L_{HS}^2 + (R_{HR} + L_{FA})^2 - L_{UA}^2}{2L_{HS}(R_{HR} + L_{FA})} \right] \right) \quad (3)$$

$$\theta_R = 0.9 * \left(\theta_{HS} + \cos^{-1} \left[\frac{R_{HR}^2 + L_{HS}^2 - (L_{UA} + L_{FA})^2}{2R_{HR}L_{HS}} \right] \right) \quad (4)$$

where

L_{UA} = Length of the upper arm segment (0.267 m)

L_{FA} = Length of the forearm/hand segment (0.333 m)

x_{HS} = Anterior distance between the wheelchair axle/hub and the shoulder

y_{HS} = Superior distance between the wheelchair axle/hub and the shoulder

L_{HS} = Distance between the wheelchair axle/hub and the shoulder

θ_{HS} = Angle of the vector from the wheelchair axle/hub to the shoulder

θ_C = Contact angle

θ_R = Release angle

Figure 2: 2D model of wheelchair propulsion used to determine contact and release angles (θ_C and θ_R , respectively). The model consisted of an upper arm (length= L_{UA}), a forearm/hand segment (length= L_{FA}), and a wheel/handrim segment (radius= R_{HR}) with revolute joints at the shoulder, elbow, hand, and hub/axle. The shoulder was set at a fixed position relative to the hub/axle (total distance= L_{HS} , anterior/posterior component= x_{HS} , superior/inferior component= y_{HS}).

The push angle (θ_P) was then defined as the difference between the release angle and contact angle as:

$$\theta_P = \theta_R - \theta_C \quad (5)$$

Additional constraints ensured that the simulations emulated realistic submaximal propulsion biomechanics. For all seat positions, the average power output to the handrim was set at 10W (Lin et al., 2009). For the initial seat position ($x_{HS} = 0.0$, $y_{HS} = 0.75$ m), the push frequency (f_P) was set at 1 Hz (PVACSCM, 2005), which is equivalent to a cycle time (t_C) of 1 second. For all other seat positions, the push frequency was determined as:

$$f_P = f_{Pi}(\theta_{Pi}/\theta_P) \quad (6)$$

where f_{Pi} is the push frequency for the initial simulation and θ_{Pi} is the push angle for the initial simulation (Richter, 2001).

The push time (t_P) was set to 40% of the cycle time (Lin et al., 2009). Three propulsion cycles were simulated, and differences between the 3rd cycle initial and final joint angles, velocities and accelerations were minimized to replicate the periodic nature of propulsion. From the initial seat position ($x_{HS} = 0.0$, $y_{HS} = 0.75$ m), the position was systematically varied independently in the superior/inferior direction (0.05m increments) and the anterior/posterior direction (0.10m increments) throughout the range of feasible seat positions. This resulted in 53 seat positions with x_{HS} values that ranged from -0.5m to 0.4m and y_{HS} values that ranged from 0.35m to 0.85m.

Simulation of Wheelchair Propulsion

To simulate manual wheelchair propulsion at each seat position, we used dynamic optimization to identify the muscle excitation patterns that minimized the change in hand force and joint torques produced by the muscles (described below). The excitation pattern ($u(t)$) of each muscle group was defined as a sum of four Henning patterns (e.g., Rankin et al., 2011):

$$u(t) = \sum_{i=1}^4 u_i(t) \quad (7)$$

where $u_i(t)$ is the excitation value at time t for the i th Henning pattern. Each Henning pattern was defined as:

$$u_i(t) = \begin{cases} \frac{m_i}{2} \left[1 - \cos\left(\frac{2\pi(t-a_i)}{b_i-a_i}\right) \right] & , \ a_i \leq t \leq b_i \\ 0 & , \ otherwise \end{cases} \quad (8)$$

where m_i , a_i and b_i are the magnitude, onset and offset respectively. This resulted in 12 parameters for each muscle group (264 total parameters).

A simulated annealing optimization algorithm (Goffe et al., 1994) was used to find the optimal muscle excitation parameters that simulated wheelchair propulsion and minimized the change in hand force and torques (Ohta et al., 2004), with the cost function defined as:

$$J = \int_0^{t_c} (\dot{F}^T \dot{F} + w \dot{\tau}^T \dot{\tau}) dt \quad (9)$$

where $\dot{F}$ is the vector containing the time derivatives of tangential, radial and lateral components of the handrim force (in units of N), $\dot{\tau}$ is the vector of the time derivatives of the joint torques produced by individual muscle forces (Nm), and w is a weighting factor (m^{-2}). The weighting factor was set at 50 m^{-2} .

Simulation Analysis

To ensure that the simulation had reached steady-state, all simulation analyses were performed on the 3rd propulsion cycle. The seat positions were originally defined as anterior and superior offsets from the initial position ($x_{HS} = 0.0$, $y_{HS} = 0.75\text{m}$) with negative numbers denoting posterior and inferior positions, but in order to transform these positions into a coordinate system that scales with subject anthropometry, the elbow flexion angle and the angle of the hub-shoulder vector were also determined when the hand was at TDC. For the initial seat position, simulation kinematics were compared to the group-averaged experimental kinematics of 12 experienced wheelchair users (from Rankin et al., 2011). For all seat positions, metabolic cost and individual muscle stresses were calculated.

Metabolic cost was calculated using a previously described model (Umberger et al., 2003; Umberger, 2010) by determining the rate of metabolic energy expenditure at each time step and integrating over the entire cycle. This value was then normalized by the total work done on the handrim throughout the cycle.

At each time step, individual muscle stress (muscle force divided by the physiological cross-sectional area) was calculated. Individual muscle stresses were then averaged across all of the muscles over the entire cycle to provide a measure of the overall upper extremity muscle stress (σ_{ue}):

$$\sigma_{ue} = \frac{1}{t_c} \int_0^{t_c} \left(\frac{1}{n} \sum_{i=1}^n \sigma_i(t) \right) dt \quad (10)$$

where n is the number of total muscles in the musculoskeletal model.

Finally, muscle antagonism was investigated as a potential cause of increased upper extremity demand by determining the average total magnitude of joint torques produced by the muscle forces (τ_{total}) and average net torque produced by the combination of these muscle torques (τ_{net}) and quantifying the excess torque (τ_{excess}) as follows:

$$\tau_{total} = \frac{1}{n_{dof} * t_c} \int_0^{t_c} \left(\sum_{j=1}^{n_{dof}} \sum_{i=1}^{n_{mus}} |\tau_{i,j}(t)| \right) dt \quad (11)$$

$$\tau_{net} = \frac{1}{n_{dof} * t_c} \int_0^{t_c} \left(\left| \sum_{j=1}^{n_{dof}} \sum_{i=1}^{n_{mus}} \tau_{i,j}(t) \right| \right) dt \quad (12)$$

$$\tau_{excess} = \tau_{total} - \tau_{net} \quad (13)$$

where n_{mus} is the number of total muscles in the musculoskeletal model and n_{dof} is the number of DOFs being examined.

Three-dimensional surfaces were then created by plotting each measure of demand with respect to seat position, and these plots were used to identify the influence of seat position on upper extremity demand.

RESULTS

The forward dynamics simulations successfully produced the manual wheelchair propulsion task. They all used an under-rim recovery pattern, which is characterized by the hand dropping below the handrim as it returns to the position of initial handrim contact (e.g., Kwarciak et al., 2009; Figure 3) This type of recovery pattern is recommended by recent guidelines (e.g., PVACSCM, 2005; Kwarciak et al., 2012). The average difference between the experimental and initial position simulation kinematics was 1.5 standard deviations (Table 1).

Figure 3: Initial position simulation hand path. The wheelchair handrim is represented by the solid black line, and the simulation hand path is represented by the blue circles.

Table 1: Average differences between initial position simulation kinematics and group-averaged experimental kinematics of 12 experienced wheelchair users (from Rankin et al., 2011). Values are given in both degrees and numbers of standard deviations.

	Simulation Difference (Degrees)	Simulation Difference (# of Experimental SDs)
Plane of Elevation	8.7	1.5
Elevation Angle	7.5	0.5
Internal-External Rotation	16.4	1.1
Elbow Flexion-Extension	23.0	1.8
Forearm Pronation-Supination	31.5	2.4
All Degrees-of-Freedom	17.4	1.5

Across the range of seat positions, there were clear regions where metabolic cost, muscle stress and muscle antagonism were minimized.

The Influence of Seat Position on Metabolic Cost

The normalized metabolic cost (Figure 4) was minimized (8.5 J) at a seat position with anterior and superior offsets from the initial position of -0.15m and -0.03m, respectively (hub-shoulder angle: -11.6° , TDC elbow angle: 106.6°). Seat positions with anterior offsets between -0.20m and 0.13m (hub-shoulder angle: between -15° and 10°) and superior offsets between -0.05m and 0.03m (TDC elbow angle: between 100° and 120°) had an associated normalized metabolic cost less than 10J. The maximum level among the seat positions investigated (22.3J) occurred at a seat position with anterior and superior offsets of -0.50m and -0.35m, respectively (hub-shoulder angle: -51.1° , TDC elbow angle: 110.0°).

The Influence of Seat Position on Muscle Stress

The overall muscle stress (Figure 5) was minimized (72.2 kPa, 9.0% of the maximum possible stress level) at a seat position with anterior and superior offsets of -0.16m and 0.01m (hub-shoulder angle: -11.5° , TDC elbow angle: 116.9°). Seat positions with anterior offsets between -0.20m and -0.01m (hub-shoulder angle: between -15° and 0°) and superior offsets between -0.03m and 0.03m (TDC elbow angle: between 105° and 120°) had an associated overall muscle stress level less than 85 kPa (10.6% of the maximum possible stress level). The maximum level among the seat positions investigated (150.2 kPa, 18.8% of the maximum possible stress level) occurred at a seat position with anterior and superior offsets of -0.10m and 0.10m, respectively (hub-shoulder angle: -6.4° , TDC elbow angle: 149.6°).

The Influence of Seat Position on Muscle Antagonism

The average excess torque across all DOFs (Figure 6) was minimized (4.6 Nm) at a seat position with anterior and superior offsets of -0.09m and 0.04m (hub-shoulder angle: -6.4° , TDC elbow angle: 122.2°). Seat positions with anterior offsets between -0.14m and -0.03m (hub-shoulder angle: between -10° and -2.5°) and superior offsets between -0.01m and 0.05m (TDC elbow angle: between 110° and 130°) had an average excess torque of less than 5 Nm. The maximum average excess torque among the seat positions investigated (14.0 Nm) occurred at a seat position with anterior and superior offsets of 0.30m and -0.10m, respectively (hub-shoulder angle: 25.1° , TDC elbow angle: 117.5°).

Figure 4: Normalized metabolic cost versus seat position. Metabolic cost was normalized by the total work done on the handrim throughout the cycle. Seat position is defined by the hub-shoulder angle and the elbow angle when the hand is at TDC.

Figure 5: Overall muscle stress versus seat position. Individual muscle stresses were computed by dividing the muscle force by the physiological cross-sectional area, and then these values were averaged across all 26 muscles and across the entire propulsion cycle to calculate the overall muscle stress. Seat position is defined by the hub-shoulder angle and the elbow angle when the hand is at TDC.

Figure 6: Average excess torque versus seat position. Average excess torque is the time-averaged difference between the total torque and the net torque, averaged across all degrees-of-freedom. Seat position is defined by the hub-shoulder angle and the elbow angle when the hand is at TDC.

DISCUSSION

The goal of this study was to identify the influence of wheelchair seat position on upper extremity demand using a 3D musculoskeletal model and forward dynamics simulations of manual wheelchair propulsion. This approach allowed a detailed analysis of direct measures of upper extremity demand (i.e., muscle stress, metabolic cost and muscle antagonism) across the range of feasible seat positions. Muscle stress, metabolic cost and muscle antagonism were all near minimum values at seat positions with an anterior offset between -0.14m and -0.03m (hub-shoulder angle: between -10° and -2.5°) and a superior offset between -0.01m and 0.03m (TDC elbow angle: between 110° and 120°).

The optimal seat superior/inferior positions identified are similar to the recommendations of previous investigators (e.g., PVACSCM, 2005; Requejo et al., 2008a; van der Woude et al., 2009). An early study suggested that biomechanical efficiency is maximized at seat superior/inferior positions between 100° and 120° TDC elbow angle (van der Woude et al., 1989), while a more recent study expanded that range to include 130° (van der Woude et al., 2009).

One study analyzing muscle activity recommended low seat superior/inferior positions (Masse et al., 1992) while another warned against moving too low (Louis and Gorce, 2010). While it has also been cautioned that joint ranges of motion increase with lower seat positions (e.g., Wei et al., 2003; Gorce and Louis, 2012), seat superior/inferior position recommendations based on spatiotemporal variables such as cadence and push angle advocate TDC elbow angles between 100° and 120° (e.g., PVACSCM, 2005; Requejo et al., 2008a).

The results of the current study suggest that upper extremity demand is minimized at superior/inferior seat positions with TDC elbow angles between 110° and 120° .

Increased muscle stress, metabolic cost and muscle antagonism resulted from seat positions outside this range. This occurred despite the improvements in push angles and cadences that accompanied more inferior positions and the favorable reductions in joint ranges of motion that accompanied more superior positions.

Most seat position studies have only examined seat positions where the shoulder is either anterior or slightly posterior to the axle, which is likely due to wheelchair stability concerns. As a result, whether their recommendations have been based on improving muscle activity (e.g., Gutierrez et al., 2005), spatiotemporal variables (e.g., Boninger et al., 2000), kinetics (e.g., Mulroy et al., 2005), additional confounding effects such as rolling resistance (e.g., Brubaker, 1986), or a combination of these factors (e.g., PVACSCM, 2005; Requejo et al., 2008a), studies have often concluded that the seat should be set in the most posterior position. However, our results suggest there is a limit to how far posterior to position the seat, as we found that some measures of upper extremity demand increase once you move posterior beyond the -10° hub-shoulder angle.

The range of seat positions at which muscle antagonism, as quantified by the average excess torque, was minimized was similar to the seat position ranges that minimized both metabolic cost and muscle stress. Because higher levels of antagonistic muscle torques appeared to cause higher upper extremity demand, the individual DOF contributions to the average excess torque values were examined to determine which muscle torques are most influenced by seat position. Minimum values for all DOFs were found at TDC elbow angles between 114.7° and 123.2° , while the anterior/posterior positions varied from -35° to -3.7° hub-shoulder angle (Table 2). The torque associated with the elevation angle was the largest contributor to the DOF-average excess torque, while the torques associated with the other two glenohumeral joint DOFs were the next largest. In addition, the torque associated with plane of elevation had the largest percent

difference between its minimum and maximum values. Consequently, the trends of the antagonistic muscles of the glenohumeral joint predominantly influenced the average excess torque trends. This highlights the importance of the muscles crossing the glenohumeral joint to manual wheelchair propulsion and the influence of seat position on the potential for injury to these muscles.

Table 2: Individual DOF excess torque values. Minimum values for each DOF and their associated hub-shoulder angles and TDC elbow angles are provided. Maximum values are provided for comparison.

		Plane of Elevation	Elevation Angle	Internal-External Rotation	Elbow Flexion-Extension	Forearm Pronation-Supination	Average (all DOFs)
Minimum Excess Torque	Hub-Shoulder Angle (°)	-5.9	-6.3	-3.7	-34.3	-35.0	-6.4
	TDC Elbow Angle (°)	119.9	120.8	123.2	114.7	118.9	122.2
	Value (Nm)	1.9	8.0	4.2	3.4	1.3	4.6
Maximum Excess Torque Value (Nm)		20.8	28.2	15.6	13.3	4.8	14.0

Limitations & Future Work

A potential limitation of this study is that it did not consider the range of propulsion techniques utilized by wheelchair users. Spatial (i.e. contact angle, release angle) and temporal (i.e. push time, cadence) variables are influenced by propulsion technique and differences in upper extremity impairment (e.g., Newsam et al., 1996; Finley et al., 2004; Mulroy et al., 2004), training (e.g., Rodgers et al., 2001; Rice et al., 2010; Richter et al., 2011) and terrain (e.g., Newsam et al., 1996; Richter et al., 2007;

Chow et al., 2009; Cowan et al., 2009). The wheelchair propulsion technique analyzed in this study was derived from representative spatiotemporal values (e.g., Richter, 2001; PVACSCM, 2005; Lin et al., 2009), but other techniques exist. Thus, future work should consider analyzing other techniques. In addition, inter-subject variability in body anthropometrics was not considered, with the musculoskeletal model representing a 50th percentile able-bodied male. However, the physical capacity of wheelchair users varies with age, gender and nature of impairment and is generally less than that of non-wheelchair users (e.g., Janssen et al., 2002; Haisma et al., 2006). Future work should include the development of subject-specific models (e.g., Zajac et al., 2002; Fregly et al., 2012) to help determine the optimal seat position for individual users.

Another potential limitation is that the model consisted of only the right half of the upper extremity and assumed symmetry. Although a recent study has suggested that wheelchair propulsion is not a completely symmetric movement (Hurd et al., 2008), most wheelchair users utilize a symmetric propulsion technique (Goosey-Tolfrey and Kirk, 2003), and therefore the influence of asymmetries on the results would likely be small. In addition, the model was further simplified by fixing the wrist joint in the standard anatomical position, even though movement at the wrist joint can create a moment at the handrim. The absence of this moment may require an increase in elbow extensor activity (Veeger, 1999). However, wrist moments are much lower than those at the elbow and shoulder joints (e.g., Robertson et al., 1996; Guo et al., 2003; Sabick et al., 2004; Guo et al., 2006), and this increase would likely be seen across all seat positions. Thus, the fixed wrist was deemed to have a minimal influence on the results and conclusions of this study.

Another potential limitation is that the model did not include all of the trunk and wrist muscles, which could influence the metabolic cost estimates. However, the

metabolic cost values corresponded to biomechanical efficiencies between 4.5% and 11.7%, which were similar to previously reported values (e.g., Goosey et al., 1998; Hintzy and Tordi, 2004; van der Woude et al., 2009). In addition, this study considered relative differences across seat positions; therefore, the lack of these muscles would likely not influence the study conclusions.

Finally, there is uncertainty in which criteria the neuromuscular control system uses to resolve muscle redundancy and complete submaximal performance tasks such as wheelchair propulsion (Todorov, 2004; Erdemir et al., 2007). Therefore, the cost function that minimized the change in hand force and torques used to generate the simulations may not be the one used by the nervous system during wheelchair propulsion. However, similar dynamic cost functions (minimizing change in torque and/or hand force) have frequently been used to successfully reproduce upper extremity movements (e.g., Uno et al., 1989; Nakano et al., 1999; Ohta et al., 2004; Svinin et al., 2005). In addition, the cost function used ensured a smooth, stable motion in both the joint and handrim coordinate frames while limiting co-contraction similarly to functions that are based on reduction of muscle activations, forces or stresses. Thus, we believe that the cost function used generated the essential characteristics of manual wheelchair propulsion to achieve our study goals.

CONCLUSIONS

This study examined the influence of wheelchair seat position on upper extremity demand and found that metabolic cost, muscle stress and muscle antagonism were all near minimum values at positions with an anterior offset between -0.14m and -0.03m (hub-shoulder angle: between -10° and -2.5°) and a superior offset between -0.01m and 0.03m (TDC elbow angle: between 110° and 120°). Previous guidelines have recommended a seat positioned as posterior as possible. However, we found there is a limit to how far posterior to position the seat, as some measures of upper extremity demand increased once you move posterior beyond the -10° hub-shoulder angle. We also found that the influence of seat position on muscle antagonism is primarily dictated by the muscles crossing the glenohumeral joint. These results can provide guidelines for positioning the seat to help reduce upper extremity overuse injuries and pain, and thus improve the overall quality of life for wheelchair users.

Appendix

Table A1: Upper extremity musculoskeletal model parameters. Fiber type composition values were based on Johnson et al. (1973). All other values were taken from Rankin et al. (2010) and Rankin et al. (2011).

Muscle	Abbreviation	Origin	Insertion	Peak Isometric Force (N)	Optimal Fiber Length (m)	Tendon Slack Length (m)	Pennation Angle (deg)	Fiber Type Composition (% Fast Twitch)
Anterior deltoid	DELT1	Clavicle	Humerus	1142.6	0.12	0.123	22	42.9
Middle deltoid	DELT2	Scapula	Humerus	1142.6	0.1078	0.1095	15	42.9
Posterior deltoid	DELT3	Scapula	Humerus	259.88	0.1367	0.038	18	42.9
Pectoralis major, clavicular head	PECM1	Clavicle	Humerus	364.41	0.1242	0.0228	17	57.7
Pectoralis major, sternocostal head portion 1 (sternum)	PECM2	Thorax	Humerus	515.41	0.1385	0.089	25	56.9
Pectoralis major, sternocostal head portion 2 (ribs)	PECM3	Thorax	Humerus	390.55	0.1385	0.152	25	56.9
Coracobrachialis	CORB	Scapula	Humerus	242.46	0.0932	0.097	0	52.7
Teres major	TMAJ	Scapula	Humerus	425.39	0.1624	0.02	16	49.5
Latissimus dorsi, thoracic portion	LAT1	Thorax	Humerus	389.1	0.254	0.12	25	49.5
Latissimus dorsi, lumbar portion	LAT2	Thorax	Humerus	389.1	0.2324	0.1765	19	49.5
Latissimus dorsi, iliac portion	LAT3	Thorax	Humerus	281.66	0.2789	0.1403	21	49.5
Subscapularis	SUBSC	Scapula	Humerus	1377.81	0.0873	0.033	20	54.7
Infraspinatus	INFSP	Scapula	Humerus	1210.84	0.0805	0.05	18.5	54.7
Teres minor	TMIN	Scapula	Humerus	354.25	0.0741	0.0813	24	54.7
Supraspinatus	SUPSP	Scapula	Humerus	487.82	0.0682	0.0595	7	40.7
Brachioradialis	BRD	Humerus	Radius	261.33	0.1726	0.133	0	60.2
Brachialis	BRA	Humerus	Ulna	987.26	0.0858	0.065	0	49.5
Biceps brachii, short head	BICshort	Scapula	Radius	435.56	0.1321	0.2123	0	57.7
Biceps brachii, long head	BIClong	Scapula	Radius	624.3	0.1557	0.3	0	57.7
Anconeus	ANC	Humerus	Ulna	350	0.027	0.018	0	67.4
Triceps brachii, lateral head	TRIlal	Humerus	Ulna	624.3	0.1138	0.098	9	67.4
Triceps brachii, medial head	TRImed	Humerus	Ulna	624.3	0.1138	0.0908	9	67.4
Triceps brachii, long head	TRIlong	Scapula	Ulna	798.52	0.134	0.217	12	67.4
Supinator	SUP	Ulna	Radius	476	0.033	0.028	0	52.7
Pronator quadratus	PQ	Ulna	Radius	75.48	0.0282	0.005	10	52.7
Pronator teres	PT	Humerus	Radius	566.22	0.0492	0.098	10	52.7

Figure A1: Plane of elevation excess torque versus seat position. Plane of elevation excess torque is the time-averaged difference between the total torque and the net torque, evaluated for the individual DOF defined by the shoulder plane of elevation. Seat position is defined by the hub-shoulder angle and the elbow angle when the hand is at TDC.

Figure A2: Elevation angle excess torque versus seat position. Elevation angle excess torque is the time-averaged difference between the total torque and the net torque, evaluated for the individual DOF defined by the shoulder elevation angle. Seat position is defined by the hub-shoulder angle and the elbow angle when the hand is at TDC.

Figure A3: Internal-external rotation excess torque versus seat position. Internal-external rotation excess torque is the time-averaged difference between the total torque and the net torque, evaluated for the individual DOF defined by the shoulder internal-external rotation. Seat position is defined by the hub-shoulder angle and the elbow angle when the hand is at TDC.

Figure A4: Elbow flexion-extension excess torque versus seat position. Elbow flexion-extension excess torque is the time-averaged difference between the total torque and the net torque, evaluated for the individual DOF defined by the flexion-extension at the elbow. Seat position is defined by the hub-shoulder angle and the elbow angle when the hand is at TDC.

Figure A5: Forearm pronation-supination excess torque versus seat position. Forearm pronation-supination excess torque is the time-averaged difference between the total torque and the net torque, evaluated for the individual DOF defined by the pronation-supination of the forearm. Seat position is defined by the hub-shoulder angle and the elbow angle when the hand is at TDC.

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