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The effect of seat position on manual wheelchair propulsion biomechanics: a quasi-static model-based approach

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Abstract

The position of the seat relative to the rear wheels is generally adjusted to modify the rearward stability of the wheelchair. Recent studies have shown that seat position also has an effect on propulsion biomechanics and suggest that seat position can be optimized. A quasi-static wheelchair propulsion model was developed to investigate the mechanism by which seat position affects propulsion biomechanics. Inputs to the model include the length of the user's arm segments, the position of the user's shoulder, the size of handrim used and the force profile on the handrim. Outputs from the model include joint kinematics, joint torques, push angle, and push frequency. Handrim force profile was determined by averaging the force profile of five wheelchair users. Force profiles were measured using the SMARTWheel. The effect of seat position on push angle was found to be directly affected by the length of the position vector from the hub of the wheel to the shoulder and indirectly affected by the angular orientation of the vector. Decreasing hub to shoulder length was found to increase push angle, decrease push frequency, decrease shoulder torque and increase elbow extension torque. It is suggested that future research investigating the role of seat position on propulsion biomechanics include both the kinematics and kinetics of the upper extremity. © 2002 IPEM. Published by Elsevier Science Ltd. All rights reserved.

Keywords: Wheelchair propulsion biomechanics; Seat position; Quasi-static propulsion model; SMARTWheel

1. Introduction

The manual wheelchair is a profoundly enabling technology for people with mobility impairments. The traditional depot-style wheelchair, found in hospitals and airports was for many years the most common wheelchair used. It is characterized today as having very few adjustments and not designed for daily long-term use [1]. Its successor, the daily rehab wheelchair, is designed with many adjustable features to maximize comfort and safety.

The primary adjustable feature on a manual wheelchair is the location of the seat with respect to the rear wheels. The seat can generally be adjusted horizontally forward and backward as well as vertically up and down. Adjusting the seat position allows the stability, maneuverability and comfort of the wheelchair to be customized for the particular functional capability and activity level of each individual user. Horizontal seat adjustment is used to modify rearward stability and sensitivity to cross-slope. Vertical seat adjustment is used to optimize fit for each user's arm length. While certain adjustments to the seat position can improve wheelchair stability and maneuverability, they may adversely affect propulsion biomechanics [2–5].

Brubaker et al. found that with the seat adjusted forward users experienced a decrease in propulsion efficiency while, at all seat heights tested, users with longer arms experienced an increase in propulsion efficiency over those users with shorter arms [2]. The investigators draw an analogy between the seat configuration of the wheelchair for a particular individual to the size of a shoe for a particular foot. In a similar study, van der Woude et al. also found an association between seat position and propulsion efficiency [3]. In addition to decreasing propulsion efficiency, it was found that increasing the seat height resulted in an increase in push frequency. Hughes et al. investigated seat positioning using adjustments relative to the user's size [4]. The seat was adjusted vertically such that with the users hand at

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Nomenclature

θ_P	push angle
θ_C	contact angle
θ_R	release angle
$\mathbf{r}_{HH}$	position vector from the hub of the wheel to the hand on the handrim
L_{UA}	length of the upper arm segment
L_{FA}	length of the forearm–hand segment
R_{HR}	radius of the handrim
$\mathbf{r}_{HS}$	position vector from the hub of the wheel to the shoulder
L_{HS}	magnitude of the position vector from the wheel hub to the shoulder
θ_{HS}	angle of the position vector from the hub of the wheel to the shoulder
θ_{Si}	initial angle of the shoulder
θ_S	angle of the shoulder
θ_E	angle of the elbow, relative to the upper arm segment
F	magnitude of the resultant force applied to the handrim
$F(\mathrm{r})$	radial component of force applied to the handrim
$F(\mathrm{t})$	tangential component of force applied to the handrim
θ_F	angle of the resultant force vector
T_S	torque applied to the shoulder
T_E	torque applied to the elbow
W	energy added to the wheel during the push phase
θ_W	wheel angle
f_P	push frequency

top-dead-center on the handrim, the forearm flexion angle was 90° and then 100°. The seat was adjusted horizontally in increments defined by a percentage of the users total arm length. Seat position was found to affect the push angle as well as the range of motion of the shoulder and elbow joints during propulsion. Push frequency was not found to increase with increasing seat height. However, the investigators believed that this result was due to the average change in seat height being only 3.3 cm.

Boninger et al., in a study of 40 full-time wheelchair users in their own wheelchairs, found a correlation between their seat position and push angle, push frequency as well as rate of rise of the force on the handrim during propulsion [5]. Those users with their seats adjusted further backwards were found to have an increased push angle, a decreased push frequency and a decreased rate of rise of the force on the handrim. Those users with their seats adjusted lower were found to have an increased push angle. The investigators recommend adjusting the seat as far backwards as is safe and comfortable for the individual user in an effort to minimize the rate and repetition of loading of the arms during propulsion.

It is clear from the results of these studies that seat position does affect propulsion biomechanics but is not clear which variables are truly affected and by how much. It is hypothesized that a mathematical model of the wheelchair and user can be useful in discovering the mechanism by which changes in propulsion biomechanics occur and offer predictions as to the magnitude of these changes. Results from such a model would serve to guide future research and to help establish guidelines for achieving an optimal seat position based on user characteristics.

2. Methods

A simplified quasi-static wheelchair propulsion model was developed. The model is two dimensional, and is similar to propulsion on a dynamometer in that the wheelchair does not move forward but rather remains stationary as the rear wheel rolls. The model consists of four rigid bodies: an upper arm, forearm–hand combination, handrim and wheel (Fig. 1). The length of the forearm–hand segment is the length from the elbow to the center of the palm of the hand with the wrist neutral. The shoulder joint and the hub of the wheel are fixed in an inertial reference frame. The hand is constrained to the handrim by a hinge joint and the handrim is rigidly attached to the wheel. Inputs to the model include the position of the shoulder with respect to the wheel hub, the handrim diameter, upper arm length, forearm–hand length and the force applied to the handrim. Since the inclusion of gravity was not expected to significantly affect the results, the model was further simplified by neglecting the mass properties of bodies.

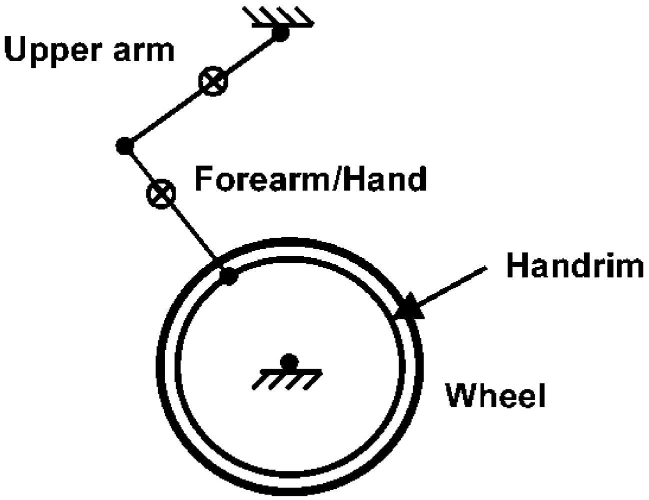

Fig. 1. Quasi-static wheelchair propulsion model. Consists of four rigid bodies: upper arm, forearm–hand, handrim and wheel. Shoulder and wheel hub are hinge joints and are fixed in inertial space. The hand is constrained to the handrim by a hinge joint. Length of the forearm–hand segment is from the elbow to the middle of the palm of the hand with the wrist neutral.

Push angle (θ_P) is defined as the difference between the release angle (θ_R) and the contact angle (θ_C). The contact angle is the angle of the position vector from the wheel hub to the hand ($\mathbf{r}_{HH}$) at the beginning of the push. With $\mathbf{r}_{HH}$ oriented vertically upwards, the angle is considered 0°. Clockwise rotations represent a positive angular displacement and counterclockwise rotations represent a negative angular displacement. The contact angle is defined such that the forearm–hand segment is perpendicular to the handrim. Release angle is defined such that the elbow is completely extended, which is represented as 0° relative to the upper arm segment. While the definitions of contact and release angles are not exact representations of what is observed in the physical system, they capture the fundamental behavior. Using the above definitions and utilizing the Law if Cosines, the contact and release angles are given by:

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

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

where L_{UA} is the length of the upper arm, L_{FA} is the length of the forearm–hand segment, R_{HR} is the radius of the handrim, L_{HS} is the magnitude of the position vector from the wheel hub to the shoulder ($\mathbf{r}_{HS}$) and θ_{HS} is the angle of $\mathbf{r}_{HS}$. θ_{HS} is defined similarly to θ_C and θ_R in that with $\mathbf{r}_{HS}$ oriented straight upwards, θ_{HS} is considered to be 0°. θ_{HS} is positive if $\mathbf{r}_{HS}$ is rotated clockwise and negative if it is rotated counterclockwise. The shoulder angle is measured relative to its neutral position hanging down along the side of the body. Extension of the shoulder results in a negative shoulder angle while flexion results in a positive angle. The initial shoulder angle (θ_{Si}) is given by:

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

The force applied to the handrim was determined by averaging the handrim force during propulsion for five wheelchair users. Subjects propelled their own wheelchairs on a dynamometer at 1.4 m/s. The dynamometer resistance was set to simulate propelling across a 2% (1:50) grade. Handrim forces and moments were measured using a SMARTWheel [6] mounted on the subject's wheelchair. Force and moment data was collected at 240 Hz for 20 s and then filtered using a fourth-order lowpass digital Butterworth filter with a 20 Hz cut-off frequency. The SMARTWheel collects forces and moments in an inertial coordinate system. A coordinate system transformation was conducted to transform the resulting forces to a wheel-fixed coordinate system. The resulting force components were radial, tangential, and lateral with respect to the handrim. Since the model is two dimensional, only the radial and tangential components were considered. For each subject, the ten consecutive pushes with an average velocity closest to that of the entire trial were used in the data analysis. Those ten pushes were then averaged to form a single characteristic push. Each characteristic push was then averaged together to form a generalized handrim force profile.

In the model, the generalized handrim force profile is scaled such that it is applied over the entire push angle. The resultant handrim force has both magnitude (F) and direction (θ_F). θ_F is described in a handrim fixed coordinate system and is 0° when the force is purely tangential and 90º when the force is purely radial. The shoulder (T_S) and elbow (T_E) joint torques required to generate the handrim force are given by:

$$T_E = (F\sin((\theta_S+\theta_E) - (90-\theta_F-\theta_W))(L_{FA}) \tag{4}$$

$$T_S = (F\cos((\theta_S+\theta_E) - (90-\theta_F-\theta_W)))(L_{UA}\cos(90-\theta_E)) \tag{5}$$

where θ_W is the wheel angle, which is defined similarly to the contact and release angles. Mechanical energy added to the wheel during a push is a function of the tangential force applied to the handrim, the radius of the handrim and the rotational displacement of the wheel. The resulting work (W) done to the wheel during a push cycle can be approximated using the average torque applied to the wheel and the total displacement of the wheel and is given by:

$$W = R_{HR}\bar{F}_t\theta_P \tag{6}$$

In order to maintain the same average propulsion speed, the rate at which energy is added to the wheel needs to remain constant. If the amount of work per push varies with seat position, then the push frequency (f_P) needs to be proportionally increased or decreased to account for such changes. For a given initial push frequency (f_{Pi}) and work per push (W_i), the push frequency is given by:

$$f_P=f_{Pi}(W_i/W) \qquad (7)$$

The model was evaluated using upper arm and forearm/hand lengths based on anatomical measurements for a 50th percentile male of 26.7 and 33.3 cm, respectively [7]. The handrim diameter was 53.4 cm, typical of that found on a 61 cm wheelchair wheel. L_{HS} was initially set such that with the user's arms hanging straight down, the center of the palm of the hand touched the hub (L_{UA}+L_{FA}). L_{HS} was then increased by 2.5-cm increments to L_{HS}+10 cm and decreased by 2.5-cm increments to $L_{HS}-10$ cm. θ_{HS} was evaluated at 0, 10 and −10°. The initial push frequency was set to 1.0 Hz.

3. Results

The effect of modifying the hub to shoulder distance on push angle and push frequency is shown in Table 1. Push angle increases with decreasing shoulder to hub distance. Push frequency decreases with decreasing shoulder to hub distance. The effect of modifying both the hub to shoulder distance and its orientation on the initial shoulder angle is shown in Table 2. The initial shoulder angle increases with decreasing shoulder to hub distance. The initial shoulder angle also increases as the orientation of the vector from the hub to the shoulder rotates clockwise about the hub.

The resultant handrim force profile with radial and tangential components is given in Fig. 2. It is characterized by an initial rapid rate of rise, followed by a gradual increase and then a decrease. The effect of varying hub to wheel distance on the joint torque required of the shoulder and elbow is shown in Fig. 3. Joint torque requirements were determined for hub to shoulder distances from −10 to 10 cm from the initial configuration in increments on 2.5 cm. The shoulder is primarily in flexion, represented as positive torque. The elbow is initially in flexion and then transfers to extension for the remaining majority of the push. As hub to shoulder distance is increased, the required shoulder flexion torque increases and the elbow extension torque decreases.

Table 1
Effect of changes in wheel hub to shoulder distance (dL_{HS}) on contact angle (θ_C), release angle (θ_R), push angle (θ_P), work done on the wheel (W) and push frequency (f_P). Initial distance from hub to shoulder is equal to the combined length of the upper and forearm–hand segments

dL_{HS}(cm)	θ_C (deg)	θ_R (deg)	θ_P (deg)	W (J)	f_P (Hz)
−10	−28	99	126	38.8	0.82
−7.5	−28	93	121	37.1	0.86
−5.0	−28	88	115	35.4	0.90
−2.5	−27	82	110	33.7	0.95
0	−27	77	104	32.0	1.00
2.5	−26	72	98	30.3	1.06
5.0	−25	68	92	28.4	1.12
7.5	−24	63	86	26.6	1.20
10	−23	58	80	24.7	1.30

Table 2
Effect of changes in wheel hub to shoulder distance (dL_{HS}) on the initial shoulder angle (θ_{Si}) for three different angles of the vector connecting the hub to the shoulder (θ_{HS}). θ_{HS} is 0° when the shoulder is directly above the hub. Initial distance from hub to shoulder is equal to the combined length of the upper and forearm–hand segments

dL_{HS} (cm)	θ_{Si} (deg) (θ_{HS}=−10°)	θ_{Si} (deg) (θ_{HS}=0°)	θ_{Si} (deg) (θ_{HS}=10°)
−10	−84	−94	−104
−7.5	−79	−89	−99
−5.0	−74	−84	−94
−2.5	−69	−79	−89
0	−64	−74	−84
2.5	−60	−70	−80
5.0	−55	−65	−75
7.5	−50	−60	−70
10	−46	−56	−66

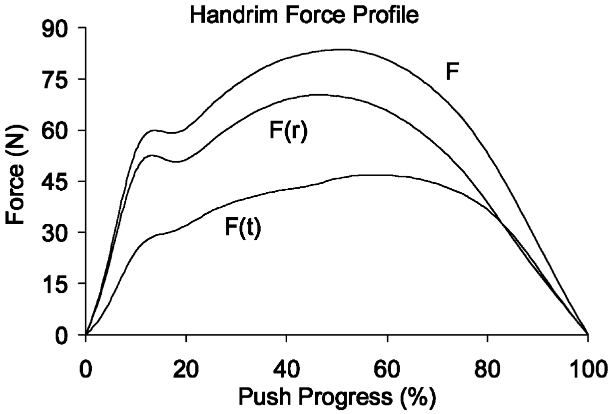

Fig. 2. Generalized handrim force profile determined by averaging the force profile of five wheelchair users. Handrim force (F) is made up of two components, tangential [F(t)] and radial [F(r)]. The tangential component acts to turn the wheel while the radial component serves to provide friction between the hand and the handrim.

4. Discussion

While the model is a greatly simplified representation of the physical system, it offers insight into the effects of seat position on propulsion biomechanics. From Eqs.

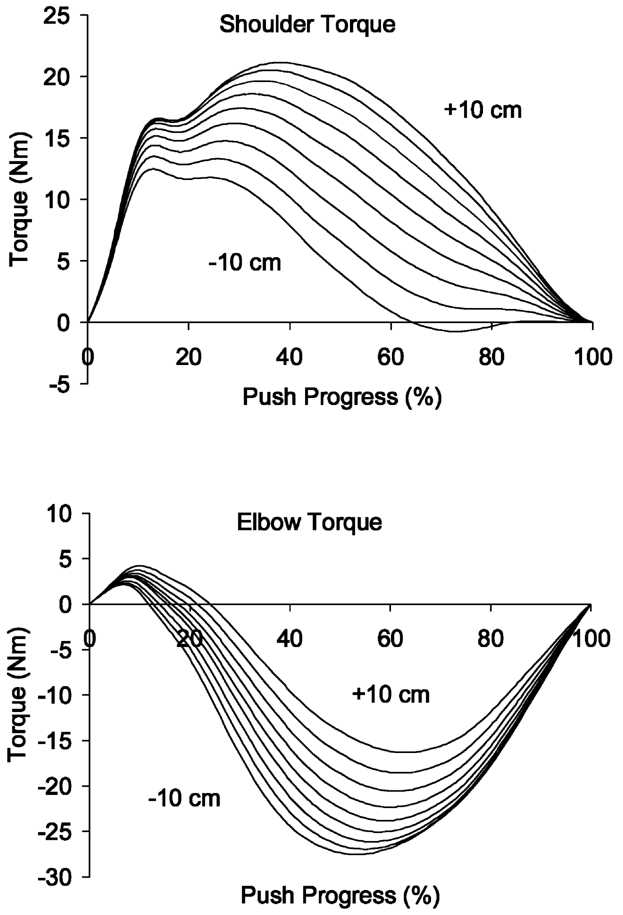

Fig. 3. Shoulder and elbow joint torque required to produce the measured handrim force profile. Joint torque requirements evaluated for hub to shoulder distances varying from −10 cm to +10 cm of the initial distance. Initial distance from hub to shoulder is equal to the combined length of the upper and forearm–hand segments.

(1) and (2) it can be seen that contact and release angles are a function of arm segment lengths, handrim size and the length of the vector from the wheel hub to the shoulder. This relation implies that the orientation of the vector from the hub to the shoulder is not influential in determining push angle. From the results in Table 1, push angle increases with decreasing length from the hub to the shoulder. Wheelchair seat position adjustments are made horizontally and vertically. In order to maintain a constant hub to shoulder length, an adjustment in the horizontal direction would need to be compensated by an adjustment in the vertical direction. Boninger et al., found that wheelchair users with their seats adjusted more forward had a decreased push angle [5]. This result suggests that compensations in the vertical direction for horizontal seat adjustments are not being made.

Although push angle is not affected by the orientation of the vector from the hub to the shoulder, initial shoulder angle is. From Eq. (3), it can be seen that changes in the angular orientation of the vector from the hub to the shoulder results in an equivalent change in the initial shoulder angle. Initial shoulder angle is important since limitations in joint range of motion could prevent the arm from taking advantage of the entire push angle available to it. Studies of propulsion kinematics suggest that the initial shoulder angle during propulsion may be somewhere between −50 and −75° [8,9]. In considering this limiting factor, positioning the seat such that the initial shoulder angle remains within its comfortable extension limit would ensure that the entire push angle can be accessed by the arm.

From Eq. (7), it can be seen that the energy added to the wheel during the push is proportional to the push angle. Since push frequency is proportional to the energy added to the wheel during the push, push frequency is also be proportional to push angle. Push frequency is an important consideration as it has been found to be associated with repetitive stress injuries of the wrist [10]. Push frequency is minimized by maximizing push angle. This relationship is similar to that found by Boninger et al. [5].

Prior to consideration of the predicted shoulder and elbow joint torque profiles, the position of the seat relative to the wheel should always be such that the hub to shoulder distance is minimized within the boundaries of the shoulder range of motion. However, as shown in Fig. 3, the hub to shoulder distance may have a profound effect on the joint torques required of the shoulder and elbow. Decreasing the hub to shoulder distance is predicted to decrease the shoulder torque required but to increase the elbow extension torque required. The short-term consequences of these predicted changes in joint torque are simply that the elbow would work harder while the shoulder worked less. The long-term consequences of these predicted changes in joint torque on the wheelchair user are not clear. Based on the results of this study, it is suggested that future research investigating the role of seat position on propulsion biomechanics include both the kinematics and kinetics of the upper extremity, in addition to those biomechanical metrics previously investigated.

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