A 2-D model of wheelchair propulsion

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Abstract

Purpose: To illustrate the potential benefits of kinetic and kinematic models in the exploration of biomechanical studies as illustrated using a simple 2-D static optimization model of wheelchair propulsion.

Method: A four-bar linkage analysis was used to determine sagittal plane motion through the range of wheelchair propulsion. Using anthropometric measures of wheelchair users, this analysis determined the angles of shoulder and elbow flexion/extension at a given point in the propulsion cycle. Maximal strength inputs for the model were collected from isokinetic measurements of shoulder and elbow moments. The torque inputs were given as functions of sagittal plane joint angles. Through selection of appropriate model performance criteria, optimization techniques determined shoulder and elbow torque contributions throughout the propulsion cycle. Variations in the model parameters of anterior-posterior (AP) seat position and handrim size were used to show potential of model to evaluate wheelchair configuration using the performance criteria of propulsive moment (Mo) and efficiency as defined by fractional effective force (FEF).

Results: The model was able to predict the magnitude and direction of force applied to the handrim from shoulder and elbow moments. These joint moments may be examined along with the generated wheelchair axle propulsion moment. While the model showed no significant changes in either Mo or FEF for AP seat changes, an increase in handrim size was shown to increase FEF.

Conclusions: This model was able to simulate wheelchair propulsion and allow for performance analyses. The open nature of the model allowed for tweaking of the kinematic inputs to examine the sensitivity of such factors as seat position and handrim size in wheelchair propulsion. Strength inputs to the model may also be altered to study the potential effects of strength training or muscle weakness.

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Introduction

Over one million people in the US rely on manual wheelchair propulsion for mobility. Spinal cord injury (SCI), aging, and mobility limiting conditions all contribute to increase the number of new wheelchair users each year. While wheelchair propulsion enhances mobility, it is not without cost. The demand on the upper extremities in wheelchair propulsion can be strenuous, particularly for the elderly or those suffering from muscle weakness. Studies on SCI patients reveal a disproportionately large number of complaints of pain and injuries of the upper extremities. The shoulder seems to be the most affected, with 31%–50% of SCI patients reporting shoulder pain. In contrast, 5%–16% of wheelchair-using individuals with paraplegia experience elbow pain.

Despite the number of injuries occurring among the large and increasing population of wheelchair users, the relationship between upper extremity loads in propulsion and the risk of musculoskeletal injury remains poorly understood. Further, the role of wheelchair fit and design factors on propulsion are largely unknown. Also, it may be difficult to detect subtle variations in propulsion technique for changes resulting from wheelchair fit (e.g. seat position and handrim size) or on variations in terrain. Analyses of such changes however, are well suited to analytical modelling. Such a model of the upper arm, forearm, and wheelchair may be used to examine the loading conditions at the shoulder, elbow and hand during wheelchair propulsion. More importantly, it could be used to examine the interaction between the users and the wheelchair. Through parametric analysis it would also be possible to evaluate wheelchair fit. The purpose of this paper is to illustrate the potential of such a concept by using a simplified two-dimensional sagittal plane model to evaluate the efficiency and power generation during a propulsion cycle when three different wheel sizes and three anterior-posterior axle positions are selected.
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Methods

In order to simulate wheelchair propulsion both the kinematics of an individual in a wheelchair, and the potential strength input available to propel the wheelchair must be known. This model consists of identifying the position of a subject in a wheelchair, characterizing the maximum potential muscle contribution of the shoulder and elbow, and combining the kinematic and strength data to predict propulsion moments. The model and all optimization routines were developed using MATLAB and the MATLAB Optimization Toolbox (The MathWorks, Inc., Natick, MA, USA).

DETERMINATION OF HAND POSITIONS AND JOINT ANGLES THROUGHOUT PROPULSION

The model determines the kinematics of wheelchair propulsion using four-bar linkage analysis where the wrist joint was assumed to be fixed. The four segments considered in the model consisted of the upper arm, the lower arm, the vector from the wheelchair axle to the point of hand contact on the pushrim, and the vector from the wheel axle to the shoulder. While the shoulder-axle link is not a rigid link per se, it can be considered as such for the purposes of our quasi-static analysis. For the model, the lengths of the upper and lower arm and the wheel radius were specified as \( r_{ES} \), \( r_{EH} \), and \( r_{OH} \), respectively (see figure 1). The shoulder position \( S \) is given as \( x_S \) and \( y_S \), measured with respect to the wheelchair axle. For each stage of the propulsion cycle the angle \( \theta_W \), denoting the position of the hand, \( H \), on the wheel rim, is known. Solution of the four-bar linkage analysis then gives the angles of the elbow and shoulder (\( \theta_E \) and \( \theta_S \), respectively). Then, for a given shoulder position and hand position, the angles of shoulder and elbow flexion could be derived.

MAXIMUM SHOULDER AND ELBOW JOINT STRENGTH CHARACTERISATION

Maximum isokinetic joint strength for the shoulder and elbow in both flexion and extension (MSF, MSE, MEF and MEE respectively) were measured. Maximal input strength data was collected on a normal subject using a Kin-Com 125 AP isokinetic dynamometer (Biodex Medical Systems, Shirley, NY) at 120°/sec. This rate was chosen by analysis of experimental data from seven experienced wheelchair users in which the average shoulder flexion/extension angular velocities during self-selected moderate wheeling were 125±27°/sec. These data were used in the model as the constraints of moment generation for the shoulder and elbow. Strength data was collected for several trials. The trials were then combined and a second-order polynomial regression was performed to give the strength data as a function of joint angle. Thus for any specified joint angle, a maximal joint moment could be obtained.

QUANTIFICATION OF JOINT KINETICS

The unknown handrim force, \( F_H \), can be obtained using an optimization procedure throughout the entire range of hand positions. The criteria selected to optimize by will greatly influence the results generated by the
model. Given the prevalence of shoulder injuries in wheelchair users described along with subsequent reporting of the shoulder's large contribution to wheelchair propulsion, it was elected to maximize the moment at the shoulder. This optimization was run for each hand position as the wheel rotates through a propulsion cycle in 5° increments. The formulation of the constrained minimization optimization problem is as follows:

Maximize \( M_S \)  
Subject to:  
\[ M_S = P_S \times F_h \]  
\[ M_E = P_E \times F_h \]  
\[ M_O = P_O \times F_h \]  
\[ -M_{\text{lex}} \leq M_S \leq M_{\text{flex}} \]  
\[ -M_{\text{lex}} \leq M_E \leq M_{\text{flex}} \]  

where the unknown independent variable, \( F_h \), is the force vector applied by the hand at the handrim. \( M_S \) and \( M_E \) are the moments generated at the shoulder and elbow, respectively, due to \( F_h \). \( M_S \) and \( M_E \) will be maximally bounded by the isokinetic torque measurements for the derived joint angle as measured above. \( M_O \) is the moment about the wheel axle generated by \( F_h \) at the handrim. As shown in figure 1, \( P_S \), \( P_E \), and \( P_h \) are the position vectors of the shoulder, elbow, and wheel axle relative to the point of force application on the handrim. Once \( F_h \) is determined (along with its components \( F_x \) and \( F_y \)), it can be broken down for further analysis into its radial and tangential components (\( F_r \) and \( F_t \)). \( F_r \) and \( F_t \) are defined as follows:

\[ F_r = F_x \cos(\theta_W) - F_y \sin(\theta_W) \]  
\[ F_t = F_x \sin(\theta_W) + F_y \cos(\theta_W) \]  

For this model, \( F_r \) is defined as positive towards the centre of the wheel, and \( F_t \) is positive in the direction of wheel travel. Additionally, the stroke efficiency will be given in terms of the fractional effective force (FEF), which is defined as:

\[ \text{FEF} = \frac{F_t}{F_h} \]  

The larger the tangential component of the applied force, the more effective the propulsion.

Results

Once a solution is found, much information may be gathered about a wheelchair propulsion cycle. Figure 2 shows how both the shoulder and elbow angle change when the axle is 2.5 cm posterior to the shoulder (axle \(-2.5\)), vertically aligned (Axle 0), or 2.5 cm anterior to the shoulder (axle +2.5). Similarly, figure 3 depicts the propulsion moments and FEF values for the various axle positions. Given the slight changes in shoulder and elbow angles throughout the propulsion cycle, it is no surprise that the propulsive moments and efficiencies are also similar.

Figures 4 and 5 show how the kinematics and kinetics of wheelchair propulsion are affected by handrim size. As the size of the handrim increases, the shoulder and elbow are not only active in propulsion for more of the propulsion cycle, but the joints themselves are put through a greater range of motion as well. However, as shown in figure 5a, the propulsion moment is still very similar. The variations in shoulder and elbow kinematics rather show up in marked increases in FEF as the handrim radius increases (figure 5b).

Conclusions

This simple model of wheelchair propulsion can provide much useful data in the simulation of a wheelchair propulsion cycle. All of the model's strength and anthropometric data are readily accessible and easily manipulated to allow desired perturbations to the simulation. As has been shown in this study, the impact of altered seat position or handrim size may be easily simulated prior to the establishment of potentially costly and time-consuming experimental testing. Additionally, it should be noted that in a past experimental study of the effect of handrim size on physiological and perceived exertion, Gayle found that peak heart rate, peak oxygen consumption, as well as perceived exertion were all decreased for larger handrims among paraplegic men. This would seem to corroborate our finding that FEF increases with handrim diameter. Similarly, through increasing or decreasing of potential muscle moment generation, the significance of the effects of muscle training or weakness may be examined.
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Figure 2 Model solutions for shoulder (A) and elbow flexion angle (B) by stroke angle. Results are shown for three anterior-posterior axle alignments: axle 2.5 cm behind the shoulder (axle -2.5); axle aligned vertically with the shoulder (axle 0); and axle 2.5 cm anterior to the shoulder (axle +2.5). Small adjustments made to the anterior-posterior position of the seat do not seem to affect joint kinematics.

Figure 3 Calculated propulsion moment (A) and FEF (B) by stroke angle for three axle alignments: axle 2.5 cm behind the shoulder (axle -2.5), axle aligned vertically with the shoulder (axle 0), and axle 2.5 cm anterior to the shoulder (axle +2.5). The small differences in shoulder and elbow kinematics seen in figure 2 do not result in significant changes to either propulsion moment or FEF.

Figure 4 Model solutions for shoulder (A) and elbow flexion angle (B) by stroke angle. Results are shown for three handrim radii (14, 20, and 25.5 cm). Note the larger range of propulsion angles and wider ranges of motion for both the shoulder and elbow as handrim size increases.
Discussion

Future directions for the model allow for increased sophistication to be included. The kinematics of the model currently assume that the wrist is not an active participant in propulsion, but rather is fixed. Additionally, the model makes use of a relatively fixed shoulder position with respect to the seat. Although there is some movement allowed for through the propulsion cycle, the prescribed motion is not currently based on experimental data.

The model currently determines hand force applied to the wheel based on maximization of shoulder contribution. In the future, optimization schemes that maximize the generated wheel moment while attempting to minimize muscle output may be implemented. Additionally, more robust optimization routines may allow the shoulder to find a position from which greater propulsion moments may be generated.

The model may even be expanded to accommodate three-dimensional motion. At this time, however, it is felt that a two-dimensional model examining wheelchair kinetics and kinematics in the sagittal plane, in which the majority of the activity occurs, is sufficient.

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References