Optimum Propulsion Technique in Different Wheelchair Handrim Diameter

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Abstract

Variability in the propulsion technique to manual wheelchair users due to differences in level of injury and wheelchair fit makes detection of subtle changes in technique due to manipulation of individual variables nearly impossible. However, such changes are well suited to analytical modeling techniques, since variables can be manipulated systematically and the effects of the manipulation can be easily quantified. Therefore, we develop a simple two-dimensional model of the upper arm, forearm, and wheelchair wheel to study changes in wheelchair-user interface. Variations in handrim diameter were examined for their influence in the propulsive moment generated by each subject. The larger the handrim, the greater the moment about the wheel axle $M_r$ could attain. The tangential force $F_t$ applied to the handrim was similar among three handrim diameters. However, the radial component of force applied to the handrim increased as handrim diameter decreased. Increasing rim diameter increased the fraction effective force. Both predicted $M_r$ and $F_t$ reached the peak value in the terminal propulsion phase, in agreement with experimental results of the quasi-static wheelchair propulsion, but not dynamic, cyclic wheelchair propulsion. During the middle to terminal propulsion phase, the optimized force vector is not tangential to the handrim and instead, it pass through the upper arm segment which is comparable to dynamic wheelchair propulsion. Optimum analysis could help us to identify the effect of handrim diameter and investigate the related characteristics. Also, through modeling, the mechanical constraints of force application and the inadequate of pushing technique are further understood.

Keywords: Wheelchair, Biomechanics, Optimization, Model

Introduction

Wheelchair design would affect the performance of the propulsion. The relationship between handrim wheelchair design and performance has been studied for the effect of features such as seat height\cite{1-3}, fore-aft position\cite{2, 3}, camber\cite{4}, wheel handrim tube diameter\cite{5}, handrim contours\cite{6}, handrim diameter\cite{7}, wheelchair weight and different types of wheelchairs. Variability in the propulsion technique to manual wheelchair users due to differences in level of injury and wheelchair fit makes detection of subtle changes in technique due to manipulation of individual variables nearly impossible. However, such changes are well suited to analytical modeling techniques, since variables can be manipulated systematically and the effects of the manipulation can be easily quantified.

Only the hand force tangential to the rim could help on the wheel progression in manual wheelchair propulsion. The pushrim resultant force ($F_r$) is divided to two components for clinical relevancy, the component of force applied tangential to the handrim $F_t$ and the component of force applied in the radial direction $F_r$. The fraction effective force ($F_{eff}$) is an indication of the amount of the propulsive force that propels the wheelchair forward, equal to the $F_t$ divided by $F_r$. Among those forces, the tangential is the largest and the radial forces is less\cite{8, 9}. The moment of the hub is the largest than those in other directions and is the only source contributing for wheelchair progression\cite{9}. However, in past study, the measured handrim force is not directed tangentially to the handrim. It is always more downward and inward to the hub of wheel\cite{10}. The $F_{eff}$ value is only about 55\% ~ 75\%\cite{8, 9}. Almost 50\% of the forces exerted at the pushrim are not directed toward forward motion and, therefore, a certain part of applied energy is wasted. Because the force pattern by experienced wheelchair user is probably close to the optimal, there must be other factors determine the force direction.

van der Helm et al.\cite{1996} used the developed musculoskeletal model to simulate the muscle force in the
shoulder during wheelchair propulsion[11]. They collect static applied handrim force in five hand positions and five different load levels at per hand positions. The upper extremity position and the measured handrim forces serve as an input to calculate utilizing inverse dynamic model. Output variables are muscle forces subject to an optimization criterion. The criterion is minimization of the sum of squared muscle stresses in this study. There are largest external moments for hand in top dead center, the highest position of handrim. And this phenomenon differs from the pattern of dynamic propulsion. Also, the force direction during quasi-static wheelchair propulsion is more tangential to the handrim than it appears in dynamic wheelchair propulsion.

Rozendaal and Veeger (2000) tried to simulate the handrim force direction based on the experimental data from wheelchair users [12]. The generated force tangential to the handrim could have greater mechanical effect on the wheel progression, but if the force is perpendicular to the line from hand to elbow or the line from hand to the shoulder will have large musculoskeletal cost on the muscles of these two joints. They use the ratio of mechanical effect and musculoskeletal cost in wheelchair propulsion cost as criteria and optimize it to find the simulated applied force direction. The direction of simulated force data during middle and terminal parts of propulsion is comparable to the actual force measured by experiment. However, the force direction during initial propulsion is upward, different from the downward pattern during real wheelchair propulsion. Also, the maximum cost-effect ratio can be obtained in initial propulsion is smaller than at the end of propulsion, means in the terminal propulsion phase is an appropriate position for generate larger force on the handrim. It is different from the results of dynamic wheelchair propulsion that the greatest applied force appears in middle propulsion.

Collection of biomechanical data is essential for understanding the handrim wheelchair propulsion. However, to further clarify how upper extremity segments and muscles interact to execute the motor task, it is necessary for the development of biomechanical models[13]. Through modeling, different aspects of the man-machine-environment system could be studied. The mechanical constraints of force application could be understood. The reasons for ineffective force could be investigated. The inadequate pushing technique could be explained[14, 15]. Therefore, the purpose of this study was to develop a two-dimensional model of the upper arm, forearm, and wheelchair wheel to study changes in wheelchair-user interface. Variations in handrim diameter were examined for their influence in the propulsive moment generated by each subject.

Research Design and Methods

Experiment data collection

The Hi Res ExpertVision™ system (Motion Analysis Corp., Santa Rosa, CA, USA) was used to record the trajectories of the fifteen reflective markers placed on selected anatomic landmarks unilaterally on each subject at 60 Hz. An instrumented wheel system was used to measure directly three-dimensional dynamic forces and moments on the handrim during wheelchair propulsion in a laboratory setting. Three sizes of handrim diameter (54, 43, and 32 cm) were assigned to the subject with randomized order. They were represented as large, middle and small, respectively. The subject had to propel at least five repetitions of wheelchair for each size of handrim. Each variable was averaged for these five trials to represent the subject performance for this handrim size. The markers’ positions were used to define the coordinate system of linkage and to determine the joint centers of upper extremity and further in the model. The point of force applied on the handrim was assumed to be the second metacarpophalangeal (MCP) joint, as has been assumed by several other investigators [16, 17]. The forces and moments exerted on the handrim of the wheelchair during propulsion were recorded by the load cell on the instrumented wheel. The radial and tangential force components were calculated using the location of the wheel center and second MCP joint.

Analytic Modeling of Propulsion Mechanics

The rationale of the model was that, given a subject-specific profile of the strengths of each of the upper extremity joints as function of joint angle, there was an optimal direction of force application to the handrim to maximize the propulsion moment about the wheel axle at each instant throughout the propulsion cycle. This optimal direction can be determined for each instant by formulating a linear optimization problem which aims to maximize the moment about the wheel axle, \( M_s \), subject to the constraints of the subject’s shoulder, elbow, and wrist joint moment-generating capabilities for the joint angles specified. The formulation was as follows (Figure 1):

Maximize \( M_s \)
Subject to:
\[
M_s = P_x \times F_s
\]
\[
M_s = P_x \times F_s
\]
\[
M_s = P_x \times F_s
\]
\[
-M_{se} \leq M_e \leq M_{se}
\]
\[
-M_{se} \leq M_e \leq M_{se}
\]

where the unknown independent variable, \( F_{ho} \), was the force vector applied by the hand at the handrim, which consists of two components (\( F_x \) and \( F_z \)), in the tangential and radial directions respectively; \( M_s \) and the \( M_e \) were the flexion/extension moments at the shoulder and elbow joints, respectively. \( M_s \) was the moment about the wheel axle generated by the force \( F_h \) at the handrim; \( P_s \), \( P_e \), and \( P_r \) were the position vectors of the shoulder, elbow, and wheel axle relative to the point of force application on the handrim. \( M_{se} \) and \( M_{se} \) were the maximum shoulder joint strengths in extension and flexion, respectively; \( M_{se} \) and \( M_{se} \) were the maximum elbow joint strengths in extension and flexion, respectively. The maximum joint strength for each position can be obtained from the measured (isokinetic) strength.
profile of the individual subject. The distance between the wrist joint center and the 2nd MCP was relatively short. Hence, the forearm was represented by the elbow joint center to 2nd MCP and the wrist joint was neglect at this model. Simplified model was feasible, since hand has to grasp the rim and follow the path of rim that the movement pattern of wrist was quite constant during propulsion phase[11, 13].

Results

Stick diagram representation of upper extremity for wheelchair propulsion in different handrim was showed in Figure 2 with 0.05 sec interval where solid and dot lines represent 32 cm and 54 cm, respectively. Always the upper extremity segments moved downward and forward during propulsion phase and upward and backward during recovery phase. The upper extremity segments has greater movement excursion and increased linear velocity while propelling the large handrim than the small handrim. The positions of upper extremity segments and joints in maneuvering large handrim are higher than those in controlling the small handrim. In this model, the shoulder goes from a position of extension to further extension slightly at initial, then after that toward flexion. For the elbow, in initial propulsion phase, it goes from a flexion position into greater flexion. At approximately top dead center the elbow begins extend until the end of stroke.

Figure 1. Optimization analytic model of wheelchair propulsion mechanic

Figure 2. Stick diagram representation of upper extremity for wheelchair propulsion with 0.05 sec interval during propulsion phase (solid line: 32 cm; dot line: 54 cm)
Figure 3. Progression moment about the wheel axle ($M_r$) for different handrim size, large (a), medium (b) and small (c)

Figure 4. Force applied to the handrim ($F_h$) for different handrim size, large (a), medium (b) and small (c)
Figure 5. Fraction effective force applied to the handrim ($F_{\text{eff}}$) for different handrim size, large (a), medium (b) and small (c).

Figure 6. Position vector of the point of force application on the handrim relative to wheel axle ($P_r$) for different handrim size, large (a), medium (b) and small (c).
Figure 7. Measured progression moment about the wheel axle (Measured $M_r$) for different handrim size, large (a), medium (b) and small (c).

Figure 8. Measured force applied to the handrim (Measured $F_h$) for different handrim size, large (a), medium (b) and small (c).
By maximize the $M_r$, we found the larger the handrim, the greater $M_r$ could attain (Figure 3). The tangential force $F_t$ applied to the handrim was similar among three handrim diameters (Figure 4). However, the radial component of force applied to the handrim increased as handrim diameter decreased. The fraction effective force ($F_{eff}$) increased with increasing rim diameter (Figure 5). Both predicted $M_r$ and $F_t$ reached the peak value in the terminal propulsion phase. For medium and small handrim, during initial propulsion phase, the point of force application on the handrim was much posterior to the wheel axis compared with large handrim (Figure 6). This phenomenon was more significant as the handrim diameter getting smaller. Under this condition, the direction of $F_t$ would be changed and the loading of elbow extensor ($M_{el}$) and shoulder flexor ($M_{sf}$) would be increased up to the physical constraints. The experiment data show greater variability for both the progression moment (Figure 7) and the tangential component of applied force (Figure 8).

**Discussion**

During propulsion almost 50% of the forces exerted at the pushrim are not directed toward forward motion and, therefore, a part of applied energy is wasted. However, some investigators do not agree with the concept that the non-tangential is wasted or just misdirected [18]. To apply a push force in mechanically most optimal direction, which is always tangential to the rim. This leads to a contradictory situation in which the elbow joint is extending while at the same time a flexor moment ought to be generated for mechanically optimal results. The produced negative power, hence, is ineffective regarding co-ordination and physiology. Our results support this concept; the optimized force direction is not tangential to the handrim and intersects upper arm segment at a point between the elbow and shoulder joints (Figure 9). Optimized force direction is acquired by both shoulder flexors and elbow extensors, which reach their physical constraints. If the optimized force direction were changed to tangential to the handrim, the shoulder flexor would reach its constraint with smaller wheel progression moment.

The developed model in this study is quasi-static and simulates static maximum muscle isometric contractions at various handrim positions. The results show the force vector is roughly tangential to the handrim. The force vector is upward when the hand position is before reaching the top dead center and downward when the hand passes the top dead center. It agrees with the experimental result studied by the van der Helm (1996)[11]. However, the force direction before the top dead center is largely differed from the experiment results of the dynamic wheelchair propulsion. The force direction of dynamic wheelchair propulsion is always downward during whole propulsion phase including
the period of hand position is behind the top dead center [8, 9, 19]. It may due to the reason, if in initial propulsion the push force is upward, the elbow flexor must be activated. But halfway through the propulsion phase the applied force must be changed to downward for progress the wheel that the elbow extensor takes turn to be activated. The muscular activity switch from elbow flexor to elbow extensor may cause movement more complex and inefficiency [10, 18, 20]. However, during static non-cyclic propulsion, the problem of switch the elbow flexor to extensor in movement mechanism disappears.

The predicted maximum applied force and progression moment occur at terminal propulsion, which is different from in vivo dynamic experimental results with peaks occurring in the middle propulsion phase. Indeed, there exist differences between quasi-static movement and dynamic cyclic movement. At initial handrim contact, the upper extremity just touches the handrim and the applied force started from zero. After then, to accelerate the wheel for progression, the applied handrim force increases and reaches its maximum in mid-propulsion phase. Later, the upper extremity segments have to be decelerated by muscle eccentric contraction for reposition in recovery phase. However, during quasi-static propulsion, the above-mentioned specific movement requirements disappear. It may be more feasible to determine the potential musculoskeletal performance during various hand positions in dynamic wheelchair propulsion.

EMG studies have ever been investigated for the muscular activation type during wheelchair propulsion in literatures [18, 21]. They found the shoulder flexors (anterior deltoid and pectoralis major) are highly activated during the most of propulsion phase. The elbow flexors (long head of biceps brachii) and elbow extensors (triceps brachii) showed a bimodal pattern, the elbow flexor activated in early propulsion and elbow extensor activated in middle and terminal propulsion phase. They conclude the shoulder flexors are the primary mover and elbow flexors and extensors are necessary for an effective force direction [18]. Some investigators analyzed the torque and power output curve during the wheelchair propulsion and found a slope change or even a negative declination in the torque curve, about half of the propulsion phase [10, 18]. This phenomenon coincides with the switch in muscular activity from elbow flexor to elbow extensor [18]. These results reveal the disadvantage propulsion design of a standard handrim wheelchair. During dynamic wheelchair propulsion, the progression moment reaches its maximum value in middle propulsion phase as required as the movement pattern. However, our model reveals the hand position in middle propulsion is not a good one for upper extremity to generate a large force on handrim, because this force, near perpendicular to the line from hand to shoulder, will result in a large shoulder moment. Similarly, propulsion force also acting near perpendicular to the line from hand to elbow will produce a large external elbow moment [11, 12]. In terminal propulsion phase, the applied force acting along with the line from hand to shoulder as well as along with the line from hand to elbow will enforce the upper extremity to generate larger force for generate larger progression moment. This model could help us get more insight of the generation of maximum progression moment and be as an index for wheelchair configuration assessment. From the different maximum generated progression moment in different hand positions, it is helpful for us to reconsider if the handrim wheel design should be changed to have the user propel the handrim with potentially generating greater progression moment. Like in wheelchair racing, users always flex trunk anterior to propel the handrim with the hand positions are much anterior to top dead center. And this is a hand positions capable for generating larger progression moment because their upper extremity joints have lever arm advantage to tolerate greater external loading before the muscles could not tolerate.

The effect of handrim size has been only investigated by Van Der Woude et al. (1988) [7]. They use five hand rim diameters (0.3, 0.35, 0.38, 0.47, 0.56 m) to examine the effects on physiological and movement parameters. They find the smaller hand rim has lower metabolic cost and higher systematic mechanical efficiency and concluded this may due to the decreased segmental excursions of the upper extremity and lower linear hand velocity. Our model reveals for the larger handrim the progression moment could be increased. However, in literature, no studies are found to investigate the effect of handrim size on progression moment during propulsion. From our results, we may conclude that using a small handrim is more energy conservation, but requires large applied force on the handrim for the same amount of progression moment.

A simple 2-D model has already been developed. This model will be refined in steps, each with progressively increasing sophistication. The assumptions applied in the initial development of the model will be step-wise relaxed or modified to improve the comparison of the analytic results with the experimental measurements. Wheelchair propulsion involves a fully three-dimensional motion of the upper-extremity and trunk. However, in the initial model development, we feel it is justified to concentrate on the plane of the dominant movement, namely, the sagittal plane. The initial model will be formulated by considering the motions of shoulder flexion and extension and elbow flexion and extension. The results of kinematics data of upper extremity will be further determined by four-bar linkage model to simulate the upper extremity movement by only acquiring the basic anthropometric data, for example the length of upper arm and forearm as well as the shoulder position related to the wheel axle. Also muscle strength model can be taken into account, using the isokinetic data of upper extremity, and determines the angle-force relationship by regression analysis. And the results may be used in the kinematics data of upper extremity determined by four-bar linkage model to find the muscle strength that will be used at particular joint angle as the constraints of the model in future. Further simulation of wheelchair dynamic propulsion seems must consider the minimization of energy losses criterion. Optimum analysis could help us identify the effect of
handrim diameter and investigate the characteristics along with.

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References