Forward Dynamic Simulation of Wheelchair Propulsion

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Introduction

Human wheelchair propulsion reflects the continual interaction between the nervous system (Control), the musculoskeletal system (Dynamics), and the environment (Wheelchair). To advance our understanding of push-phase dynamics and to test hypotheses regarding multijoint control, an experimentally validated dynamic model needs to be developed.

Previous wheelchair propulsion models have estimated upper extremity net joint moments and muscle forces using only inverse dynamics. [1-4]. Changes in mechanical loading have been observed to vary with modifications in the direction and magnitude of the measured reaction force between the hand and wheelchair pushrim. Systematic evaluation of selective modifications to wheelchair propulsion techniques and seat design on reaction forces at the hand-rim interface requires development of a forward dynamics model. This study uses experimental results to validate an upper extremity forward dynamics model of wheel chair propulsion.

Methods

Experimental

One wheelchair user volunteered to participate in this study in accordance with the Institutional Review Board at the Ranchos Los Amigos National Rehabilitation Center, Downey, CA. Reflective markers were used to monitor the 3D motion of the hand, forearm, upper arm, and trunk segments. Three markers were also placed on the right wheel to track wheel rotation (VICON, 50 Hz). The force applied to the wheelchair during propulsion was measured using force transducers (2500 Hz) mounted along the spokes of the wheel.

Modeling

Body segment kinematics were calculated from marker data projected in the sagittal plane. A four segment two-dimensional dynamic model of the human body was created using a dynamic simulation software package ADAMS (Mechanical Dynamics, Ann Arbor, Michigan, USA). Segments for the trunk, head, upper arm, and forearm were represented as rigid bodies connected by revolute joints with scaled body segment parameters [5] (Figure 1). The wheel was modeled as a rigid body element with a moment of inertia and resistive torque specific to the wheelchair dynamics for the individual user.

A visco-elastic surface model was implemented to describe the vertical and horizontal reaction forces at the wrist-pushrim interface [6], consistent with the observed kinematics of the user.

$$F_y = C\left|y_f\right|^{2.5} - D\left|y_f\right|\dot{y}_f$$

where y_f is the vertical displacement of the wrist into the pushrim, C = 4x10^6 N/m$^{2.5}$, and D = 3x10^5 N s/m^2. The same constants were used for the horizontal direction.

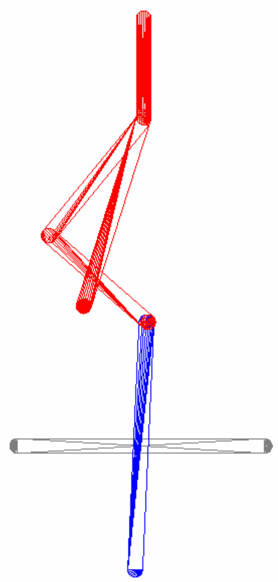

Figure 1: Dynamic model with human body (Red) interacting with wheel (Blue)

Validation of the model was done in the following steps: 1) Wheel motion was defined by applying measured force on wheel 2) Experimental-based 2D shoulder and elbow joint motions were defined by wheel motion 3) Forward simulations were driven with combination of feedforward and feedback joint torques 4) Feedforward torques were optimized to minimize error in simulated reaction forces

The feedback torques were proportional+derivative (PD) joint controllers of the form

$$\tau = k(\theta - \theta^*) + b(\dot{\theta} - \dot{\theta}^*)$$

where θ^* and $\dot{\theta}^*$ are the desired joint angle and joint angular velocities, θ and $\dot{\theta}$ denote the current joint state, and k and b are proportional and derivative gain constants. The PD joint controllers ensure that experimental-based joint trajectories are maintained during optimization. Feedforward torques are parameterized as seven point splines and were allowed to vary to minimize the error between simulated and measured (denoted '*') horizontal and vertical reaction forces.

$$\varepsilon = \sqrt{(F_x - F_x^*)^2 + (F_y - F_y^*)^2}$$

The combination of PD joint controllers and the optimization procedure yields a set of joint torques that produces a set of motions and reaction forces close to experimental measurements.

Parametric Study: Once the feedforward torques were determined as the best fit with experimental measures, the initial torso angle was parameterized. Torso angle was adjusted from -20 (rotated forward) to +20 (rotated backward) degrees from its initial orientation (defined from right horizontal). Feedforward torques remained constant. Simulated reaction forces were recorded and compared.

Results

The simulated reaction forces matched the measured force-time curves in both horizontal and vertical directions with root mean squared errors of 3.28 N and 2.48 N, respectively (Figure 2). The match between simulated and measured elbow and shoulder joint angles was reasonable (up to 8 degrees difference in elbow angle).

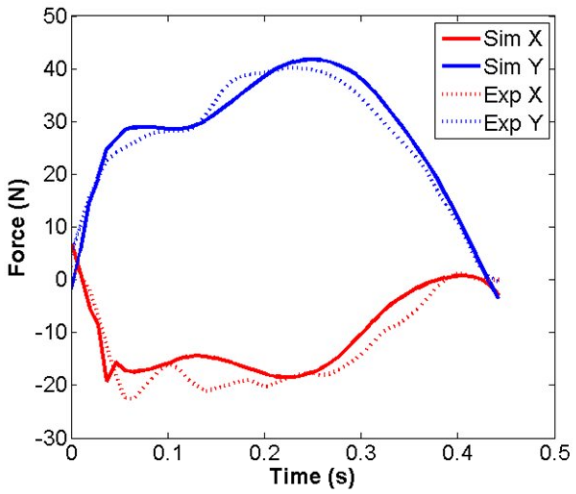

Figure 2: Simulated vs. Experimental Reaction Forces

Simulated peak vertical force was reduced as initial torso angle decreased (more horizontal) and increased with a more vertical torso angle (Figure 3). A similar effect was found in simulated horizontal force.

Discussion

The dynamic model was able to reproduce the measured reaction forces between the forearm and wheel pushrim. The accuracy of the simulated joint trajectories was reasonable and is expected to improve if out of plane motion of the arm and/or wheel is accounted for in the model.

The parametric study illustrates how variation of initial torso angles may contribute to modifications in peak vertical and horizontal reaction forces. The results of this parametric study are currently being validated against experimental data.

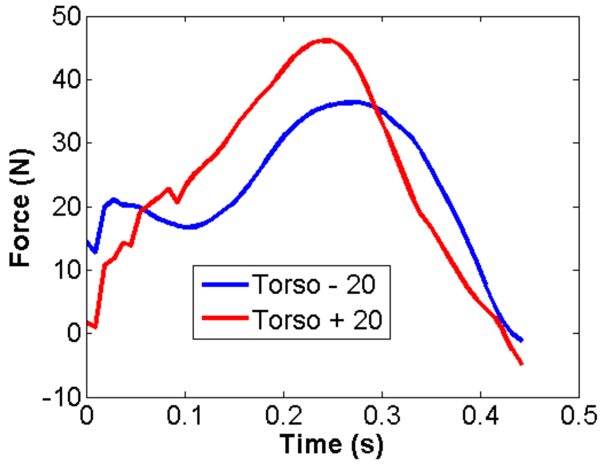

Figure 3: Simulated vertical reaction forces at different torso angles

Conclusion

An experimentally validated forward dynamic model of the push phase of wheelchair propulsion was developed. The parametric study illustrates how an experimentally validated forward dynamic model can be used to test hypotheses regarding control strategies of wheel chair users. Future work will use the model to test hypotheses under different conditions such as seat position, trunk orientations, and control strategies implemented.

References

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