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Three-Dimensional Dynamic Analysis of Wheelchair Propulsion

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A three-dimensional (3-D) inverse dynamic model of wheelchair propulsion was developed using the Newton-Euler method based on body coordinate systems. With this model, the arm was assumed to be three rigid segments (hand, forearm, and upper arm) connected by the wrist, elbow, and shoulder joints. A symbolic method was adopted to generate the equations of motion. The model was used to compute the joint forces and moments based on the inputs obtained from a 3-D motion analysis system, which included an instrumented wheelchair, video cameras, and a data acquisition system. The linear displacements of markers placed on the joints were measured and differentiated to obtain their velocities and accelerations. Three-dimensional contact forces and moments from hand to handrim were measured and used to calculate joint forces and moments of the segments.

Key Words: model, joint forces, joint moments

The number of studies focusing on wheelchair propulsion kinetics is limited (Robertson & Cooper, 1993; Strauss, Maloney, Ngo, & Phillips, 1991; Ting, Sheth, & Brubaker, 1991; Veeger, Rozendal, & Van der Woude, 1991). Long-term use of manual wheelchairs may predispose wheelchair users to overuse injuries of the shoulders, elbows, and wrist due to repetitive stroke patterns and total reliance on the upper body musculature for propulsion. Dynamic analysis of wheelchair propulsion may reveal the force and moment distribution in each joint and may also help improve wheelchair design.

Researchers who have modeled wheelchair propulsion have considered either the two-dimensional dynamic or three-dimensional (3-D) static case. Some of those models analyzed only the forearm and upper arm. Veeger, Van der Woude, and Rozendal (1991) developed a 3-D dynamic model and used an experimental setup that enhanced understanding of wheelchair propulsion dynamics in nondisabled subjects. In the present study, we developed a new dynamic model for computing the 3-D joint forces and moments of the upper limb specifically for lower-limb-disabled wheelchair users during wheelchair propulsion.

A Dynamic Model of Wheelchair Propulsion

For this model, the hand, forearm, and upper arm were considered as three rigid links connected by the wrist, elbow, and shoulder joints. The hand was considered to have two angular displacements (radial/ulnar deviation and flexion/extension), the forearm to have flexion/extension motion, and the upper arm to have abduction/adduction and flexion/extension motions. Supination/pronation of the forearm and internal/external rotation of the upper arm were neglected because of difficulty in obtaining accurate kinematic data for these movements during wheelchair propulsion. A body coordinate system was defined for each limb (hand, forearm, and upper arm). The hand was considered the point of force application to the handrim. The handrim point of force application was assumed to be coincident with the fifth metacarpal head. Angular motion of the limbs was computed from the 3-D displacements of corresponding markers placed at the joints that were videotaped by two cameras. These relative rotations were used to obtain the transformation matrices. By using the 3-D displacements of each segment along with handrim reaction forces and moments, this model enabled us to calculate joint forces and moments.

The schematic diagram shown in Figure 1 illustrates the coordinate systems for the wheelchair propulsion model. The model consisted of three rigid bodies (hand, forearm, and upper arm) connected by three joints (wrist, elbow, and shoulder). Coordinate defini-

Figure 1 — Global and local coordinate systems for the wheelchair propulsion model. Body segments (trunk, upper arm, forearm, hand) and segmental motions as a function of moments about local axes (flexion/extension, abduction/adduction, ulnar/radial deviation) are indicated. Note: $x'_w - y'_w - z'_w$ coordinates are parallel to $x_w - y_w - z_w$ coordinates.

tions are reported using the guidelines of the International Society of Biomechanics (Wu & Cavanagh, 1995). The coordinate systems of handrim, hand, forearm, upper arm, trunk, and global were defined as follows:

- $x_w-y_w-z_w$: handrim coordinates (tangential, radial, and medial-lateral, respectively)
- $x_h-y_h-z_h$: hand coordinates (radial-ulnar, proximal-distal, and palmar-dorsal, respectively)
- $x_l-y_l-z_l$: forearm coordinates (anterior-posterior, proximal-distal, and medial-lateral, respectively)
- $x_u-y_u-z_u$: upper arm coordinates (anterior-posterior, proximal-distal, and medial-lateral, respectively)
- $x_t-y_t-z_t$: trunk coordinates (anterior-posterior, proximal-distal, and medial-lateral, respectively)
- $X-Y-Z$: global coordinates (right-handed orthogonal triad fixed in the ground with +Y axis upward and parallel with the field of gravity, +X axis in a plane perpendicular to the Y axis and forward, and +Z axis in a plane perpendicular to the Y axis and to the side)

Coordinate Transformations

The transformation from one coordinate system to the other was usually obtained by means of three successive rotations. In this paper, Euler's angles (Meirovitch, 1970) with an x - y - z rotation sequence were adopted. The transformation matrix with three successive rotations can be written as

$$\underline{R}_{i+1,i}(\theta_3, \theta_2, \theta_1) = \begin{bmatrix} C\theta_2 C\theta_3 & C\theta_1 S\theta_3 + S\theta_1 S\theta_2 C\theta_3 & S\theta_1 S\theta_3 - C\theta_1 S\theta_2 C\theta_3 \\ -C\theta_2 S\theta_3 & C\theta_1 C\theta_3 - S\theta_1 S\theta_2 S\theta_3 & S\theta_1 C\theta_3 + C\theta_1 S\theta_2 S\theta_3 \\ S\theta_2 & S\theta_1 C\theta_2 & C\theta_1 C\theta_2 \end{bmatrix} \quad (1)$$

where $S\theta_j = \sin(\theta_j)$ and $C\theta_j = \cos(\theta_j)$. θ_1 is rotation about the x -axis, θ_2 is rotation about the y -axis, and θ_3 is rotation about the z -axis.

This transformation matrix can be simplified depending on the degrees of freedom that describe the motion of each link. For example, the hand was modeled with two degrees of freedom (rotations about x and z axes), and thus θ_2 was taken as zero. The forearm had one degree of freedom (rotation about z axis), and therefore θ and θ_2 were set to be zero. Finally, the upper arm had two degrees of freedom (rotations about x and z axes), so θ_2 was taken as zero.

Equations of Motion

To reduce the burden and avoid potential human errors, the symbolic language Maple was used to generate the equations of motion. The equations of motion for each segment were derived based on the free body diagram in Figure 2. The force and moment equations of the i^{th} body are

$$\underline{F}_{i-1,i}^i + \underline{F}_{i+1,i}^i + \underline{F}_{g,i}^i = m_i \underline{a}_i^i \quad (2)$$

$$\underline{M}_{i-1,i}^i + \underline{M}_{i+1,i}^i + \underline{\tilde{I}}_{i,1} \underline{F}_{i+1,i}^i + \underline{\tilde{I}}_{i,2} \underline{F}_{i-1,i}^i = \underline{\dot{H}}_i^i \quad (3)$$

$$\dot{\underline{H}}_i^i = \frac{d}{dt}(\underline{I}_i \underline{\omega}_i) = \underline{I}_i \underline{\alpha}_i + \tilde{\underline{\omega}}_i \underline{I}_i \underline{\omega}_i \quad (4)$$

where $\underline{F}_{i-1,i}^i$ = force from (i-1)th body to ith body expressed in ith body coordinates; $\underline{F}_{i+1,i}^i$ = force from (i+1)th body to ith body expressed in ith body coordinates; $\underline{F}_{gi}^i$ = gravitational force of body i expressed in ith body coordinates; $\underline{a}_i^i$ = linear acceleration of body i expressed in ith body coordinates; m_i = mass of body i; $\underline{M}_{i-1,i}^i$ = moment from (i-1)th body to ith body expressed in ith body coordinates; $\underline{M}_{i+1,i}^i$ = moment from (i+1)th body to ith body expressed in ith body coordinates; $\dot{\underline{H}}_i^i$ = rate of change of angular momentum of body i expressed in ith body coordinates; $\underline{I}_i$ = principle mass moment of inertia of body i; $\underline{r}_{i,1}$ = distal length of body i; $\underline{r}_{i,2}$ = proximal length of body i; $\sim$ = symbol for skew symmetric matrix; $\underline{\omega}_i$ = angular velocity of body i; and $\underline{\alpha}_i$ = angular acceleration of body i.

Data Collection

Instrumented Wheelchair

The wheelchair measurement system included two Peak 3-D CCD cameras, a VCR/monitor assembly, an image processing unit, a 3-D force/torque transducer, a speed potentiometer, an amplifier, an analog-to-digital unit, and a PC. A schematic of the wheelchair propulsion measurement system is shown in Figure 3. A Quickie 2HP lightweight wheelchair (Everest & Jennings, Camarillo, CA) instrumented with an AMTI (Advanced Mechanical Technology, Inc., Newton, MA) MC3A-6-1000 multicomponent force/torque transducer in its wheel hub was used to record the handrim forces and moments (Strauss et al., 1991). The AMTI force/torque transducer uses bonded strain gauges to measure forces and moments in three dimensions (six channels). This transducer allows a maximum torque (Mz) of 56.5 N · m and maximum plane-of-wheel forces (Fx and Fy) of 2,224 N. Output of the transducer is in microvolts, with full-scale output less than 18 μ V on all channels. An 8-bit absolute position optical encoder monitors the angular position of the wheel, transducer, and handrim assembly.

Figure 2 — Free body diagram of body i used for derivation of equations of motion for the model.

Figure 3 — Schematic of the wheelchair propulsion measurement system.

We calibrated the wheelchair using data processing to remove the zero load offset data from the data acquired during subject testing, to convert the analog data output to appropriate units, and to remove the effect of mechanical cross-coupling that occurs from channel to channel. A program was used to remove the zero offset data from the analog data taken during subject testing by first matching the encoder value at each time increment with the corresponding encoder value in the zero load offset data file, then subtracting the force and moment values in the zero load offset data file from the respective force and moment values. Another program converted the analog output to appropriate units and removed the effect of mechanical cross-coupling. For this program, the user is prompted to input a previously determined 6×6 transformation matrix that relates known input to system measured output. The units in this matrix are in volts per force or moment applied. Mechanical cross-coupling is accomplished by multiplying each row of the analog data by the inverted transformation matrix. Within-system calibration testing provided a nominal range of accuracy for the moment channels (M_x , M_y , M_z) of $\pm 0.5 \text{ N} \cdot \text{m}$ and for the force channels (F_x , F_y , F_z) of $\pm 2.0 \text{ N}$ (Scott, 1990).

Three-Dimensional Motion Analysis

A Peak5 3-D motion analysis system (Peak Performance Technologies, Englewood, CO) was used to collect the kinematic data. We calibrated the system by videotaping a precisely calibrated cube with 24 known coordinate points. This frame was then digitized, and direct linear transformation computations (Wood & Marshall, 1986) were performed. The calibration frame used the Cartesian (X - Y - Z) coordinate system. We corrected for camera lens distortion (Panasonic WV d5000 and AG450) by videotaping a known size square and calculating the correction with the aspect ratio portion of the software package. System accuracy of the Peak Performance Motion Measurement System evaluated by the company under static conditions exhibited a 0.50 mm error or ratio to size field.

Members of our lab have performed dynamic evaluations and determined the angular calculations to be within 1° at speeds up to $300^\circ/\text{s}$. Videotaping for this project was performed at 60 Hz. Spherical retroreflective markers were used. For this demonstration, the subject rolled the instrumented wheelchair at a velocity of 3 km/hr over a 10 m distance that included the area for videotaping. A cycle from the midrange was selected for analysis.

The kinematic data collection played a very important role in calculating joint forces and moments. The marker displacements were recorded using cameras, and joint angles, velocities, and accelerations were calculated using the Peak Performance software. Rotational matrices were formed for intermediate transformation of data from one local coordinate system to the other using relative angles.

The displacement data were differentiated to form velocity and acceleration vectors. The linear acceleration of each marker was used to interpolate the linear acceleration of the center of mass of each limb, which was transformed into respective body coordinates. The contact forces and moments, measured in three directions (F_x , F_y , F_z , M_x , M_y , and M_z) from the hub transducer, were transformed from handrim to hand coordinates and were used as the input to the hand equations. Although high-frequency noise due to differentiation was present, the contact forces and moments were considered as zero during the release of the hand. The actual moment applied by the hand to the wheelchair rim was obtained by the following formula:

$$\underline{M}_T = \underline{M}_m - \underline{r}_w \underline{F} \quad (5)$$

where $\underline{F}$ is the measured contact force from hand to handrim, $\underline{r}_w$ is the radius of the hand rim of the wheelchair, $\underline{M}_m$ is the measured moment from hand to hand rim of the wheelchair, and $\underline{M}_T$ is the actual moment from hand to hand rim of the wheelchair.

The displacements collected by the video cameras were differentiated to form velocity and acceleration vectors. The raw data were smoothed using a Butterworth low-pass filter with a cutoff frequency of 6 Hz. Strokes were defined as beginning at the point of initial handrim loading. Contact phase was defined as the entire time of handrim loading, with recovery phase being defined as the time when the hand was not in contact with the handrim. These motion vectors, forces, and torques as well as the anthropometric data were the input variables to a program that computes the forces and torques of the wrist through an inverse dynamics process. The recursive program then determines the forces and torques of the elbow and shoulder, respectively.

Marker Configuration

Previous work (Rodgers et al., 1994) determined optimal marker placement for wheelchair motion analysis. Seven markers were used in the wheelchair model to identify the motion of each limb. The locations of these markers, as shown in Figure 4, were the fifth metacarpal head, styloid process, lateral epicondyle of the humerus, acromion, greater trochanter, wheel hub, and handrim. The cameras were positioned such that all the markers could be viewed in both cameras at any time during the experiment.

Inertia Properties of the Human Body

Inertial characteristics of the limb segments were estimated based on cadaver work of Hanavan (1964). To illustrate the model, one nondisabled male subject was chosen for data collection. The length of each limb and the subject's weight were measured and used

Figure 4 — Marker locations used in motion analysis of wheelchair propulsion to obtain kinematic data for model input.

Table 1 Anthropometric Data

Segment	Segment weight/ total body weight	Center of mass/ segment length		Moment of inertia $\text{kg}\cdot\text{m}^2 \times 10^{-4}$		
		Proximal	Distal	I_{xx}	I_{yy}	I_{zz}
Hand	0.006	0.506	0.494	3.8	3.8	3.8
Forearm	0.016	0.430	0.570	71	8	71
Upper arm	0.027	0.436	0.494	179	23	179

to calculate the segment geometry and weights based on the information in Table 1 (Baughman, 1983; Miller & Nelson, 1976; Winter, 1990).

Results

The measured contact forces and the calculated moments from hand to handrim in global coordinates are shown in Figures 5 and 6. These forces and moments were then transformed into hand coordinates and used as inputs to the hand equations. The forces and moments at the wrist, elbow, and shoulder joints were computed recursively in their body

Figure 5 — Measured handrim contact forces during wheelchair propulsion in global coordinates: F_x (—), F_y (···), and F_z (---), which are tangential, radial (normal), and medial-lateral forces, respectively.

Figure 6 — Calculated moments from the hand to the handrim during wheelchair propulsion in global coordinates: M_x (—), M_y (···), and M_z (---), which are the moments about the tangential, radial (normal), and medial-lateral axes, respectively.

coordinates. Figures 7 and 8 show the forces and moments at the wrist joint. The force in the y-direction (longitudinal or superior-inferior) was the largest of the directional forces at the wrist. The radial/ulnar deviation moment was larger than the moment in flexion/extension. The forces and moment at the elbow joint are shown in Figures 9 and 10. The forces were similar to those at the wrist joint, with the y-direction (longitudinal or superior-inferior) force again being the largest of the three directional forces. Only the flexion/extension moment was available at the elbow joint. The forces and moments at the shoulder joint are shown in Figures 11 and 12. The force distribution at the shoulder joint was different than that of the wrist and elbow joints. The force in the x-direction (anterior-posterior) was dominant as compared to the y and z direction forces, whereas the flexion/extension moment (M_z) was dominant as compared to the abduction/adduction moment (M_x) at the shoulder joint.

The magnitude of joint forces generally started from zero, increased up to the maximum as the person pushed the handrim, and decreased to zero during release. The medial-lateral force at all three joints was small compared to the anterior-posterior and longitudinal forces. The longitudinal forces were negative because the y-axis was directed upward. The flexion/extension moments were dominant at the shoulder and elbow, while the ulnar/radial deviation moment dominated at the wrist. Moments at the shoulder joint were greater than moments at the elbow and wrist joints. Since the hand rotated from the x to the y axis during propulsion, the moments about the z-axis were positive.

Results indicate that the model could be used in experiments involving interventions that would be expected to affect joint stresses. These interventions might include handrim material alterations, hand or wrist bracing, changes in body positioning, and/or exercise training intervention.

Figure 7 — Computed forces at the wrist joint during wheelchair propulsion: f_x (—), f_y (···), and f_z (---), which are anterior-posterior, proximal-distal, and medial-lateral, respectively. Forces in the y-direction (compressive forces) were largest at the wrist.

Figure 8 — Computed moments at the wrist joint during wheelchair propulsion: m_x (—) and m_y (---), which are flexion-extension and radial-ulnar deviation, respectively. The radial-ulnar deviation moment (m_y) was largest at the wrist.

Figure 9 — Computed forces at the elbow joint during wheelchair propulsion: f_x (—), f_y (---), and f_z (···), which are anterior-posterior, proximal-distal, and medial-lateral, respectively. Compressive force (f_z) was largest at the elbow.

Figure 10- The computed flexion/extension moment at the elbow joint during wheel chair propulsion shown as m_x (---).

Figure 11 — Computed forces at the shoulder joint during wheelchair propulsion: f_x (—), f_y (···), and f_z (---), which are anterior-posterior, proximal-distal, and medial-lateral, respectively. Force in the anterior direction (f_x) was dominant.

Figure 12 — Computed moments at the shoulder joint during wheelchair propulsion: m_x (—) and m_z (---), which are abduction-adduction and flexion-extension moments, respectively. Shoulder moments about the z-axis (flexion/extension) were largest.

Summary

A method was presented for determining joint forces and moments in 3-D coordinates to describe the dynamics of wheelchair propulsion. Results indicate that the model could be used in experiments involving interventions that would be expected to affect joint stresses. The results presented in this paper were in close agreement with Veeger's results using nondisabled subjects (Veeger, Rozendal, & Van der Woude, 1991). Though the supination/pronation of forearm and internal/external rotation of the upper arm were neglected because of hardware limitations, the forearm and upper arm rotated about the longitudinal axis during wheelchair propulsion. For this reason, y moments at all three joints were not reported. Due to the vibration of markers during propulsion and the inherent effects of differentiation, the linear acceleration of the markers had high-frequency nonrandom noise that is probably responsible for the nonuniformity of the force and moment curves. This, however, can be improved using higher speed cameras and smaller markers. The assumption that we used in developing the model, that the handrim point of force application was coincident with the fifth metacarpal head, will also influence the moments calculated about the wrist depending on the variability of hand grasp used (Cooper, Robertson, VanSickle, Boninger, & Shimada, 1996).

Within these constraints, the model provides insight into stroke mechanics which may assist practitioners in assessing stress reduction interventions for wheelchair users and which may enhance wheelchair design. The model could be used in experiments involving handrim material alterations, hand or wrist bracing, changes in body positioning, and/or exercise training intervention.

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