RELATING SHOULDER JOINT FORCES AND RANGE OF MOTION DURING WHEELCHAIR PROPULSION

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INTRODUCTION

Nearly one half of manual wheelchair users (MWUs) have shoulder pain and/or injury even after short periods of time in a wheelchair (Nichols, 1979). Repetitive loading on the shoulder joint and the large range of motion (ROM) that the shoulder passes through during wheelchair propulsion has been implicated as a causative factor of pain and injury among MWUs. At the start of a propulsion stroke, the shoulder is generally in extension, abducted and internally rotated (Cooper, 1999). As the stroke progresses, both internal rotation and abduction decreases and the shoulder is flexed. Shoulder angles are at their extremes at the beginning and end of the stroke. If the forces acting at the shoulder are also at their peak magnitudes, this may cause the arm to be more prone to injury. The purpose of this paper was to investigate the timing between peak shoulder angles and forces during wheelchair propulsion. The measures analyzed in this paper can be helpful in the prevention and treatment of shoulder pain and injury in MWUs.

PROCEDURES

Ten (5 men and 5 women) experienced MWUs with a spinal cord injury at the T-4 level or below provided written informed consent to participate in this study. The average age and years post injury was 34.6 ± 7.0 years and 11.4 ± 2.6 years, respectively.

Subjects' own personal wheelchairs were fitted bilaterally with SMART^{Wheels}, force and torque sensing pushrims (VanSickle, (1995). Skin mounted markers of the OPTOTRAC motion analysis system (Northern Digital, Inc.) were attached to the following bony landmarks: acromion process, lateral epicondyle, olecranon, 3rd and 5th metacarpalphalangeal joints, radial and ulnar styloids. A rigid body made of carbon fiber composite was secured to the sternum to measure trunk movement. Wheelchairs were secured to a dynamometer with a resistance comparable to that of a tile floor. Participants were instructed to propel at a steady-state speed of 0.9 m/s (2mph) for about a minute during which 20 seconds of kinetic and kinematic data were collected.

Shoulder joint angles and forces were determined using a local coordinate system approach (Cooper *et al*., 1999). The output variables of the biomechanical model include 3-D net joint forces acting at the glenohumeral joint (anterior/posterior, superior/inferior, and medial/lateral components) and shoulder angles expressed in anatomical terms (sagittal and horizontal flexion/extension, abduction/adduction, and internal/external rotation). The time at which maximum and minimum angles and forces occurred were determined using MATLAB (Mathworks, Inc.). The difference between each peak shoulder angle and peak force component was calculated for each stroke. Time differences were averaged across the first five strokes.

RESULTS AND DISCUSSION

The average times in between when the peak force component and peak angle occurred was shortest for the group of MWUs for several force component/angle combinations: superior force and maximum sagittal flexion angle, lateral force and minimum abduction angle, anterior force and maximum horizontal extension angle, and superior force and minimum abduction angle (see Table 1). For most subjects, each peak force/angle combination occurred at the end of the propulsion stroke with the exception of the horizontal extension angle and anterior force component which occurred at the beginning of the stroke. A representative plot showing the peaks of the superior force and sagittal flexion angle is shown in Figure 1.

Wheelchair propulsion requires the shoulder to move over a large ROM. At the extreme ends of its ROM, the shoulder was also found to be subjected to peak loading conditions. At the beginning of the push phase, an anteriorly directed force is acting at the same time the shoulder is maximally extended (in the horizontal plane). At the end of the push phase the force acting upward through the joint was at its peak when the shoulder was fully flexed (in the sagittal plane) and while minimally abducted. Both of these peak force and angle combinations may be causing the head of the humerus to be forced up further into the joint which can over time result in impingement under the acromioclavicular arch and subsequent inflammation. Distributing the propulsion forces more uniformly along the pushrim may help reduce the incidence of shoulder joint pain and injury in MWUs.

Table 1 –Time differences between peak shoulder force components and shoulder angles (Mean ± SD)

Peak Force Components	**Peak Angle**	**Mean Time Difference (msec)**
Superior	Max sagittal Flexion	45.7 ± 22.2
	Min abduction	18.6 ± 11.5
Lateral	Min abduction	51.8 ± 18.5
Anterior	Max horizontal extension	91.2 ± 50.7

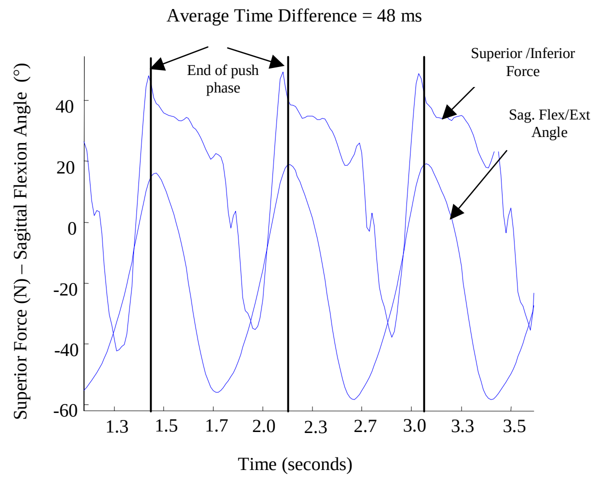

Figure 1: Superior/inferior shoulder force component and sagittal flexion/extension angle at 2 mph for a single subject

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