ORIGINAL ARTICLE

Shoulder Biomechanics During the Push Phase of Wheelchair Propulsion: A Multisite Study of Persons With Paraplegia

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ABSTRACT. Collinger JL, Boninger ML, Koontz AM, Price R, Sisto SA, Tolerico ML, Cooper RA. Shoulder biomechanics during the push phase of wheelchair propulsion: a multisite study of persons with paraplegia. Arch Phys Med Rehabil 2008;89:667-76.

Objectives: To present a descriptive analysis and comparison of shoulder kinetics and kinematics during wheelchair propulsion at multiple speeds (self-selected and steady-state target speeds) for a large group of manual wheelchair users with paraplegia while also investigating the effect of pain and subject demographics on propulsion.

Design: Case series.

Setting: Three biomechanics laboratories at research institutions.

Participants: Volunteer sample of 61 persons with paraplegia who use a manual wheelchair for mobility.

Intervention: Subjects propelled their own wheelchairs on a dynamometer at 3 speeds (self-selected, 0.9m/s, 1.8m/s) while kinetic and kinematic data were recorded.

Main Outcome Measures: Differences in demographics between sites, correlations between subject characteristics, comparison of demographics and biomechanics between persons with and without pain, linear regression using subject characteristics to predict shoulder biomechanics, comparison of biomechanics between speed conditions.

Results: Significant increases in shoulder joint loading with increased propulsion velocity were observed. Resultant force increased from 54.4±13.5N during the 0.9m/s trial to 75.7±20.7N at 1.8m/s ($P<.001$). Body weight was the primary demographic variable that affected shoulder forces, whereas pain did not affect biomechanics. Peak shoulder joint loading occurs when the arm is extended and internally rotated, which may leave the shoulder at risk for injury.

Conclusions: Body-weight maintenance, as well as other interventions designed to reduce the force required to propel a wheelchair, should be implemented to reduce the prevalence of shoulder pain and injury among manual wheelchair users.

Key Words: Biomechanics; Rehabilitation; Shoulder; Spinal cord injuries; Wheelchairs.

PEOPLE WITH SPINAL CORD injury (SCI) often rely on their ability to propel a manual wheelchair for independent mobility. Wheelchair propulsion requires a person to impart a force to the wheelchair pushrim to move forward. As a result, the joints of the upper limb are loaded repeatedly as the manual wheelchair user performs activities of daily living.[1-5] The shoulder joint in particular is designed for mobility, not load bearing. This may be the reason that many manual wheelchair users report shoulder pain. Estimates of shoulder pain among manual wheelchair users with paraplegia range from 30%[6] to 73%.[7]

Many investigators believe that repetitive loading during wheelchair propulsion, termed *overuse syndrome*, is a potential cause for pain.[8-10] Our most recent investigation[11] supported this idea, because joint kinetics resulting from wheelchair propulsion were linked to shoulder pathology. Mercer et al[11] found that people who experienced larger forces and moments were more likely to have coracoacromial pathology or to exhibit signs of pathology on physical examination. It has been well documented that manual wheelchair users experience shoulder pain; however, it is not known how pain affects shoulder biomechanics during wheelchair propulsion.

A few studies[1-5,12-14] have described 3-dimensional (3D) shoulder biomechanics during propulsion. Most of these studies have been conducted at a single site with a relatively small number of subjects, usually fewer than 20 participants.[1,3,4,12,14] Some studies focused solely on shoulder kinetics[3,4] and others on shoulder kinematics.[12,14] The largest study we are aware of reported kinetics and kinematics of wheelchair propulsion for 47 manual wheelchairs users with varying medical conditions.[2] Comparisons between studies are difficult because of differences in testing conditions. An instrumented wheelchair ergometer was used in some studies[2-5,14]; others[1,13] tested subjects in their own wheelchairs but on a dynamometer setup. Also, different coordinate systems are used when reporting joint kinetics and kinematics.[2,4,12-14] Inconsistency also exists in the propulsion speeds, with some studies focusing on steady-state speeds[1,2,5,12,13] and others examining only self-selected velocities.[3,4]

Our goal is to present a descriptive analysis and comparison of shoulder kinetics and kinematics during wheelchair propulsion at multiple speeds (self-selected and steady-state target

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speeds) for a large group of manual wheelchair users with paraplegia and to investigate the effect of pain and subject demographics on propulsion. It is important to study self-selected velocity because the way a person propels the wheelchair on an everyday basis may be linked to pathology. However, because biomechanics vary with propulsion speed,[3,4,13] target speeds are valuable for directly comparing biomechanic variables between subjects. We hope to move toward a more standardized description of shoulder kinematics by using the International Society of Biomechanics (ISB)–recommended Euler angle sequence.[15] Glenohumeral forces and moments will be referenced to local coordinate systems with 3 rotational degrees of freedom. Despite differences in testing conditions, previous studies[3,4,13] with different setups have observed increased joint loading at faster different speeds of propulsion, and we expected the current study to confirm those results in the largest subject population to date. We also believe that persons with shoulder pain will experience less joint loading because of a modified propulsion style. Using a multisite approach to recruit a large group of subjects also allows us to investigate the influence of subject demographic characteristics such as age, years since injury, injury level, height, weight, and sex on propulsion biomechanics. We hypothesized that increased subject weight would result in increased joint loading. Finally, we describe the relationship between the timing of peak shoulder kinetics and arm posture, because loading the shoulder in vulnerable positions may contribute to shoulder pathology.

METHODS

Three sites participated in data collection: the Human Engineering Research Laboratories (HERL) in Pittsburgh, PA; Kessler Medical Rehabilitation Research and Education Corp (KMRREC) in Orange, NJ; and the University of Washington (UW) in Seattle, WA. This study was approved by each site's institutional review board.

Participants

A total of 61 subjects (21 from HERL, 20 from KMRREC, 20 from UW) volunteered and provided informed consent before participation in this study. All subjects used a manual wheelchair as their primary means of mobility, were over 18 years old, and had an SCI below T1 that had occurred more than 1 year before participation in the study. Each subject also had a wheelchair with quick-release wheels to ensure compatibility with the kinetic measurement device. People were excluded from this study if they had a history of fractures or dislocations in the arms including the shoulder, elbow, and wrist; upper-limb dysthetic pain as a result of a syrinx or complex regional pain syndrome type II; or if they had upper-limb pain that prohibited them from propelling a manual wheelchair. Nondominant-side data were used for all analyses. Five of the 61 subjects were left-handed; all others were right-hand dominant. Demographic information including height, weight, age, years since injury, injury level, and sex was collected from all subjects. Subjects were also asked 2 questions about shoulder pain: (1) Have you had any shoulder pain in the last month? (2) Does your shoulder hurt you while you are propelling your wheelchair?

Instrumentation and Data Collection

Wheelchair dynamometer. Each subject's wheelchair was secured to a dynamometer that had 2 independent rollers, 1 for each wheel. The resistance of the rollers is comparable to propelling over a tile surface.[16] All 3 dynamometers used in this study were fabricated and assembled at HERL, and they are checked and maintained every 6 months at each site. Subjects were instructed to acclimate themselves to the dynamometer setup before testing. Real-time speed and direction feedback were displayed on a monitor in front of subjects during the trials. Subjects participated in 3 speed trials: a self-selected comfortable pace, 0.9m/s (3.2km/h [2mph]), and 1.8m/s (6.4km/h [4mph]). Subjects performed the self-selected trial first so that they would not be influenced by the speed display, followed by the 0.9m/s and 1.8m/s trials. After a subject reached a steady-state speed, data were collected for 20 seconds. Subjects were allowed to rest as needed, approximately 1 minute, between trials.

Kinetic data. The SmartWheel,[a] a 3D force- and torque-sensing device, was used at each site to measure propulsion kinetics at the pushrim.[17] Each SmartWheel was fitted bilaterally at HERL and KMRREC, and unilaterally at UW, to each subject's own wheelchair. At UW, the SmartWheel was fixed to each subject's nondominant side while an inertia-matched wheel was fitted to the other side. Because UW only had access to 1 SmartWheel, the nondominant side was chosen because it may be less affected by pathology not related to wheelchair propulsion. Attaching the SmartWheel to a subject's own wheelchair does not change the wheel placement, alignment, or camber. Kinetic data were collected at 240Hz and digitally filtered with an eighth-order, zero-phase, low-pass Butterworth filter with a 20-Hz cutoff frequency. Kinetic data were down-sampled to 60Hz for comparison with kinematic data. Previously, investigators of this multisite study identified differences in pushrim kinetics between sites, presumably because of small differences in rolling resistance of the dynamometers. A method based on deceleration on the dynamometer was developed to correct for the kinetic differences to combine the data in future analyses.[18] This method calculates a coefficient of friction for each dynamometer system based on rolling resistance and normal force (a percentage of the subject's body weight distributed by the rear wheel). Differences in coefficients of friction between sites and individual body weights were used to adjust data from the collaborating sites (KMRREC, UW) so that data were comparable with the lead site (HERL). Adjusted pushrim forces were used as input to the inverse dynamics model.

Kinematic data. Different motion-capture systems were used at each of the 3 sites, but all were capable of outputting 3D marker position data relative to a global origin (located between the 2 rollers of the wheelchair dynamometer). The HERL site used 2 Optotrak 3020 systems,[b] the KMRREC site used a Vicon 612 Workstation,[c] and the UW site used a Qualisys MCU240 system.[d] The resolution of the Optotrak system is .01mm at a camera distance of 2.25m, and the maximum residual marker error of both the Vicon and Qualisys motion capture systems is less than 1.5mm. The same marker set was used at all 3 sites and included markers at the third metacarpophalangeal joint, radial styloid, ulnar styloid, lateral epicondyle, acromion, sternal notch, C7 vertebrae, T3 vertebrae, and greater trochanter. Each site was responsible for determining the optimal sampling frequency and interpolation methods for their motion-capture system. For final analysis, all kinematic data were down-sampled to 60Hz. Kinematic data were digitally filtered with a fourth-order, zero-phase, low-pass Butterworth filter with a 7-Hz cutoff frequency.

Data Analysis

Inverse dynamics. Cooper et al[1] previously described the anthropometric model used for this study. Segment lengths and upper-extremity circumferences of all subjects were measured