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Shoulder joint kinetics and pathology in manual wheelchair users

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Abstract

Background. Manual wheelchair users rely heavily on their upper limbs for independent mobility which likely leads to a high prevalence of shoulder pain and injury. The goal of this study was to examine the relationship between shoulder forces and moments experienced during wheelchair propulsion and shoulder pathology.

Methods. Kinetic and kinematic data was recorded from 33 subjects with paraplegia as they propelled their wheelchairs at two speeds (0.9 and 1.8 m/s). Shoulder joint forces and moments were calculated using inverse dynamic methods and shoulder pathology was evaluated using a physical exam and magnetic resonance imaging scan.

Findings. Subjects who experienced higher posterior force (Odds Ratio (OR) = 1.29, $P = 0.03$), lateral force (OR = 1.35, $P = 0.047$), or extension moment (OR = 1.35, $P = 0.09$) during propulsion were more likely to exhibit coracoacromial ligament edema. Individuals who displayed larger lateral forces (OR = 4.35, $P = 0.045$) or abduction moments (OR = 1.58, $P = 0.06$) were more likely to have coracoacromial ligament thickening. Higher superior forces (OR = 1.05, $P = 0.09$) and internal rotation moments (OR = 1.61 $P = 0.02$) at the shoulder were associated with increased signs of shoulder pathology during the physical exam.

Interpretation. Specific joint forces and moments were related to measures of shoulder pathology. This may indicate a need to reduce the overall force required to propel a wheelchair in order to preserve upper limb integrity. Potential interventions include changes to wheelchair setup, propulsion training, or alternative means of mobility.

Keywords: Shoulder; Joint kinetics; Spinal cord injury; Biomechanics; Pathology; Inverse dynamics

1. Introduction

It is well documented that manual wheelchair users (MWUs) with paraplegia have a high prevalence of shoulder pain and injury (Ballinger et al., 2000; Bayley et al., 1987; Boninger et al., 2001; Escobedo et al., 1997; Lal, 1998; Nichols et al., 1979; Pentland and Twomey, 1991; Sie et al., 1992; Subbarao et al., 1995). Estimates of shoulder pain among manual wheelchair users with paraplegia range from 30% (Ballinger et al., 2000) to 73% (Pentland and Twomey, 1991). A recent review article noted that shoulder pain is often a result of musculoskeletal pathology (Dyson-Hudson and Kirshblum, 2004). Another study reported that the acromioclavicular joint of the shoulder is the most susceptible to degenerative changes (Lal, 1998). MWUs rely on their upper extremity for independent

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mobility and other critical functions, and thus shoulder pain can be debilitating. One study found that pain was the only factor correlated with lower quality-of-life scores (Lundqvist et al., 1991).

"Overuse syndrome" has been described as one potential cause for pain in MWUs (Bayley et al., 1987; Nichols et al., 1979; Subbarao et al., 1995). Manual wheelchair propulsion places repeated loads on the upper limbs, with a stroke cycle time of less than a second. The shoulder joint experiences loading throughout the propulsion cycle (Cooper et al., 1999; Finley et al., 2004; Kulig et al., 1998; Mulroy et al., 2005; Rodgers et al., 1994). Ergonomics literature has previously identified a link between repetitive loading tasks and musculoskeletal disorders (Andersen et al., 2003; Frost et al., 2002; Leclerc et al., 2004; NIOSH, 1997). In a report of musculoskeletal disorders in the workplace, the National Institute for Occupational Safety and Health defined repetitive activities for the shoulder as activities that involve cyclical flexion, extension, abduction, or rotation of the shoulder joint (NIOSH, 1997). Wheelchair propulsion, while not an occupational task, fits this definition. The effects of repetition can be magnified when combined with awkward postures or loading of the upper extremity such as occurs in wheelchair propulsion (Andersen et al., 2002; Frost et al., 2002; NIOSH, 1997).

It is important to understand what biomechanical factors may predispose individuals to musculoskeletal shoulder pathology so that interventions can be developed. Task performance modification based on ergonomic analysis has proven effective in reducing risk factors for pain and upper extremity pathology in various work settings (Carson, 1994; Chatterjee, 1992; Orgel et al., 1992). Additionally, research has shown that many interventions can be applied to alter propulsion biomechanics (Boninger et al., 2005).

Previous studies have reported shoulder joint forces and moments during propulsion (Cooper et al., 1999; Finley et al., 2004; Kulig et al., 1998; Mulroy et al., 2005; Rodgers et al., 1994), but none have investigated a relationship to shoulder pathology. Since shoulder pain and injury are so common among MWUs, we hope to elucidate biomechanical risk factors for shoulder injury so that potential interventions can be developed. The goal of this study was to calculate shoulder forces and moments during two speeds of manual wheelchair propulsion and determine if biomechanics were related to shoulder pathology. We hypothesized that higher shoulder forces and moments during wheelchair propulsion would be correlated to a higher incidence of shoulder pathology as measured by physical examination and magnetic resonance imaging (MRI).

2. Methods

2.1. Subjects

Subjects were recruited from two primary sources: wheelchair vendors and discharge records from a large inpatient spinal cord injury (SCI) rehabilitation unit. A letter was sent to all potential subjects stating the purpose of the study and asking them to contact the laboratory if they wished to participate in the study. This recruiting method allowed us to identify all individuals with SCI, not just those currently being followed through regular clinic visits. The study was approved by our institutional review board and 33 individuals, 23 males and 10 females, provided informed consent prior to participation in this study. All subjects used a manual wheelchair as their primary means of mobility and had a spinal cord injury below the level of T1 that occurred after the age of 18. Subjects were excluded from this study if they had a history of fractures or dislocations in the arms including the shoulder, elbow and wrist or upper limb dysthetic pain as a result of a syrinx or complex regional pain syndrome Type II (formerly known as reflex-sympathetic dystrophy [RSD]). Subjects were also excluded if they had upper limb pain that prohibited them from propelling a manual wheelchair. We also asked all subjects if they had experienced pain while propelling their wheelchair in the last month, but did not exclude subjects based on their response. Each subject propelled his or her own wheelchair during testing. All analyses for this study were performed using data collected from the non-dominant side of the subject. Four of the 33 subjects were left-handed, while all others were right-hand dominant.

2.2. Instrumentation and data collection

2.2.1. Kinetic data

Each subject's own wheelchair was fitted bilaterally with SMART$^{\text{Wheels}}$ (Three Rivers Holdings, LLC, Mesa, AZ) (Cooper et al., 1997) and secured to a dynamometer system using a four-point tie-down system. The SMART$^{\text{Wheel}}$ measures three-dimensional forces and torques applied to the pushrim. Attaching the SMART$^{\text{Wheel}}$ to the subject's own wheelchair does not change the wheel placement, alignment, or camber. All subjects were instructed to acclimate themselves to the dynamometer setup prior to testing. Kinetic data were collected at 240 Hz and digitally filtered with a 8th order zero-phase lowpass Butterworth filter with a 20 Hz cutoff frequency. Kinetic data was downsampled to 60 Hz for comparison with the kinematic data.

2.2.2. Wheelchair dynamometer

The dynamometer is comprised of two independent rollers, one for each wheel. The resistance of the rollers is comparable to propelling over a tile surface (DiGiovine et al., 2001). Real time speed and direction feedback were displayed on a monitor in front of the subject during the trials. During testing, subjects were instructed to propel at two speeds: 0.9 m/s and 1.8 m/s (2 and 4 mph). After each subject reached a steady-state speed, data were collected for 20 s. Subjects were given a rest period of at least one minute between trials.

2.2.3. Kinematic data

Two three-camera Optotrak (Northern Digital, Inc., Waterloo, Ontario) motion analysis systems were used to track the position of infrared markers placed on bony landmarks of the upper extremity including the third metacarpophalangeal joint, radial styloid, ulnar styloid, lateral epicondyle and acromion (Cooper et al., 1999). Markers were also placed on the wheel hubs during data collection. Axle position was calculated at the center of the circle created by the spinning wheel hub marker. Kinematic data were collected at 60 Hz and digitally filtered with a 4th order zero-phase lowpass Butterworth filter with a 7 Hz cutoff frequency.

2.3. Measures of shoulder pathology

We looked at two measures of shoulder pathology: physical exam and MRI.

2.3.1. Physical exam

The same physician conducted a physical exam focused on signs of shoulder injury on all the subjects. Subjects were tested for pain or discomfort during: resisted abduction and internal rotation, resisted internal rotation, resisted external rotation, resisted abduction, palpation over the sub-deltoid bursa, and palpation of the biceps tendon. Six clinical exams were performed and a score was given for each (0 = symptom/sign absent, 1 = equivocal finding, 2 = symptom/sign present). A composite physical exam score was defined as the sum of all six measures and ranged from 0 indicating no pathology to a maximum possible score of 12. This physical exam protocol was used in a previously published study from our laboratory (Boninger et al., 2001).

2.3.2. Magnetic resonance imaging (MRI)

Subjects underwent an MRI scan of their non-dominant shoulder to investigate shoulder pathology. The non-dominant side was tested because it may be less prone to degenerative changes due to work or leisure related activities. The MRI protocol was specifically designed to detect rotator cuff pathologies (Boninger et al., 2001). MRI scans were reviewed by a radiologist for distal clavicular edema, acromioclavicular degenerative joint disorder (ACDJD), AC edema, acromial edema, subacromial osseous spur formation, entheseal edema, coracoacromial (CA) ligament edema, and CA ligament thickening. For each of the eight pathologies, MRI findings were scored on a scale as follows: 0 = absent, 1 = mild, 2 = moderate, 3 = severe (Kjellin et al., 1991; Liou et al., 1993). This method of interpretation (i.e., breaking the reading of MRI down into finite elements) is designed to avoid global judgments of pathology. Such global judgments may neglect important information and are less reliable because of the differential weighting of information across judges. MRI scans were also reviewed for rotator cuff tears. Only one subject had a rotator cuff tear on the non-dominant side. Since the occurrence was so low, we decided to exclude rotator cuff tears from the analysis and focus on other measures of shoulder pathology.

2.4. Data analysis

2.4.1. Anthropometric model

Cooper et al. previously described the anthropometric model used for this study (Cooper et al., 1999). Segment lengths and upper extremity circumferences of all subjects were measured as input to Hanavan's mathematical model which calculates the inertial properties of each body segment (Hanavan, 1964). In this model, the hand is modeled as a sphere and the forearm and upper arm are both modeled as frustums of right circular cones. Segment center of mass locations were also estimated using the methods described by Hanavan.

2.4.2. Inverse dynamics

An inverse dynamics model was created to transform pushrim forces to the glenohumeral joint (Cooper et al., 1999). Non-dominant side data were used for all subjects and calculations for the inverse dynamics model were performed using Matlab (The Mathworks Inc., Natick, MA). Pushrim forces measured by the SMART$^{\text{Wheel}}$ were used as input to the model to calculate joint reaction forces and moments at the shoulder in the global coordinate system as described in previous work (Cooper et al., 1999). The forces and moments are then transformed to the local coordinate system of the proximal segment of the shoulder joint, the trunk (Fig. 1). The shoulder joint reaction forces and moments describe the external demand placed on the joint during propulsion.

All reported forces and moments are referenced to the trunk local coordinate system and are presented with left shoulder sign convention, since the majority of the subjects were left-hand non-dominant. (Fig. 1) This allowed for averaging of all data regardless of the subject's non-dominant side. Since the Cooper et al. study was published, the inverse dynamics program has been modified for use with the kinematic marker set described above, while approximating the local coordinate systems recommended by the International Society of Biomechanics (ISB) (Wu et al., 2005). We defined our trunk coordinate system as described by Eqs. (1)–(3).

The z-axis (+medial/−lateral) of the trunk points from the left acromion (ACR_L) to the right acromion (ACR_R).

$$\vec{z}_{\text{trunk}} = \frac{\vec{\text{ACR}}_\text{R} - \vec{\text{ACR}}_\text{L}}{\|\vec{\text{ACR}}_\text{R} - \vec{\text{ACR}}_\text{L}\|} \tag{1}$$

The x-axis (+anterior/−posterior) of the trunk is perpendicular to the z-axis and a vector pointing from the midpoint of the acromions (ACR_{mid}) to the midpoint of the two wheelchair axles (AXLE_{mid})

$$\vec{x}_{\text{trunk}} = \vec{z}_{\text{trunk}} \times \frac{\vec{\text{AXLE}}_{\text{mid}} - \vec{\text{ACR}}_{\text{mid}}}{\|\vec{\text{AXLE}}_{\text{mid}} - \vec{\text{ACR}}_{\text{mid}}\|} \tag{2}$$

Fig. 1. Trunk local coordinate system.

The y-axis (+superior/−inferior) is perpendicular to the x- and z-axes

$$\vec{y}_{\text{trunk}} = \vec{z}_{\text{trunk}} \times \vec{x}_{\text{trunk}} \tag{3}$$

For each shoulder force or moment, the peak values during the push phase of propulsion were calculated. Push phase was determined through visual inspection of pushrim force and moment data. All forces and moments are reported with reference to an anatomical direction rather than positive and negative values along or about a specific axis. The sign convention for a left shoulder is as follows:

- F_x: +anterior, −posterior
- F_y: +superior, −inferior
- F_z: +medial, −lateral
- M_x: +abduction, −adduction
- M_y: +external rotation, −internal rotation
- M_z: +flexion, −extension

During the push phase, posterior, superior, and lateral forces were observed at the shoulder. Abduction, internal rotation, and extension moments also occurred during the push phase of propulsion. The expected sign convention was confirmed through visual inspection of all data. Additionally, stroke frequency and mean velocity were calculated using data from the SMART$^{\text{Wheels}}$. For each subject, peak values were averaged over every stroke in the 20 s steady-state trial.

2.4.3. Statistics

Descriptive analysis including means and standard deviation (SD) for continuous variables was initially performed to describe the subject characteristics and unadjusted mean of biomechanical variables at two different speeds. All shoulder kinetic variables were measured at two speed conditions for each subject, multivariable mixed effects models were developed to test the difference of biomechanical variables, including shoulder forces and moments, between the two speeds while adjusting for subject age and mass. A logistic regression model was used to examine the association between shoulder pathology and the subject characteristics of age, mass, years of injury and gender, in which individual MRI scores and physical exam score were dichotomized as (0 = no symptom, 1 = yes symptom) due to the skewed distribution. Additionally, logistic regression models were used to test the association between shoulder biomechanics and shoulder pathology (MRI measures, physical exam score) and the subject characteristics of age and mass. Shoulder biomechanics were represented as the average of the peak force or moment for the two speed conditions. Odds ratios (OR) were calculated to describe the relationship between shoulder biomechanics and pathology. Four subjects were missing data from either the 2 mph or 4 mph trial because the instrumentation failed during one of the speed conditions. The linear regression method was used to input the missing shoulder kinetic data for these subjects so that all 33 subjects could be used in the analysis (Little and Rubin, 2002). Statistical analyses were conducted using SAS version 8 (SAS Institute, Inc., Cary, NC). Verifications of model assumptions and fit were carried out via examination of residual plots and the Hosmer–Lemeshow goodness-of-fit test as appropriate (Hosmer and Lemeshow, 1989). Significance was set at $P < 0.05$ and trends were defined as $0.05 \leqslant P < 0.10$.

3. Results

3.1. Subjects

Thirty-three individuals, 23 males and 10 females, participated in this study. The subjects had an average height and mass of 1.74(SD = 0.1) m and 75.6(SD = 15.6) kg respectively. All subjects were between 20 and 70 years old and the average age of the group was 37.8(SD = 11.2) years. The average time since injury to the test date was 12.4(SD = 6.1) years ranging from 1 to 25 years. Only 4 out of 33 subjects reported experiencing pain while propelling their wheelchair in the month prior to testing. Because of this small number of positive responses, we analyzed all subjects as a single group, rather than conducting a separate analysis for those subjects who had experienced pain.

3.2. MRI and physical exam

MRI scans of the non-dominant shoulder of all 33 subjects were reviewed for eight markers of shoulder pathology. MRI results for a subsample of this subject group ($n = 28$) were previously presented in a paper focused on shoulder imaging abnormalities in individuals with paraplegia (Boninger et al., 2001). The frequencies and types of abnormalities observed on the MRI scans are summarized in Table 1. MRI score distribution is also presented. All subjects, except one, had one or more of the presented abnormalities. All pathologies except osseous spur were present in more than half of the subjects, with coracoacromial ligament edema being the most common. Subjects exhibited mild and moderate abnormalities for all eight MRI measures. Severe abnormalities were noted for five of the eight measures. Ten out of 33 subjects (30%) reported discomfort during one or more clinical tests during the physical exam. The subject group had an average physical exam score of 1.03, with observed scores ranging from 0 to 10. The median and mode of the physical exam scores for the ten individuals with pathology was 2.

Using logistic regression, subject age was not significantly related to the physical exam score or any of the MRI measures. Subject mass was significantly associated with physical exam score ($P = 0.05$), acromioclavicular joint edema ($P = 0.04$), and coracoacromial ligament thickening ($P = 0.02$). Odds ratios and 95% confidence intervals describing these three relationships are presented in Table 2. Higher body mass increases the odds of having shoulder pathology as indicated by physical exam. Additionally, heavier individuals have greater odds of exhibiting acromioclavicular joint edema or coracoacromial ligament thickening.

Table 1
MRI abnormalities of the non-dominant shoulder

$n = 33$	Abnormal *n* (%)	Score = 1 *n*	Score = 2 *n*	Score = 3 *n*
Distal clavicular edema	18 (55)	10	8	0
AC joint DJD	17 (52)	7	8	2
AC joint edema	17 (52)	12	3	2
Acromial edema	19 (58)	16	3	0
Osseous spur	10 (30)	7	2	1
Entheseal edema	22 (67)	13	9	0
CA ligament edema	29 (89)	13	15	1
CA ligament thickening	21 (64)	9	9	3

AC = acromioclavicular, DJD = degenerative joint disease, CA = coracoacromial.
MRI score: 1 = mild, 2 = moderate, 3 = severe.

Table 2
Relationship between subject mass and shoulder pathology

	Subject mass (kg)	
	OR	95%CI
Physical exam score	1.03	(1.00, 1.06)
AC joint edema	1.02	(1.00, 1.05)
CA ligament thickening	1.03	(1.00, 1.05)

AC = acromioclavicular, CA = coracoacromial, OR = odd ratio, CI = confidence interval.

3.3. Shoulder biomechanics

Figs. 2 and 3 depict representative raw force and moment data respectively from a single subject for the 0.9 m/s condition. The values reported correspond to the force or moment applied to the trunk via the humerus at the glenohumeral joint as a result of applying a force to the pushrim during wheelchair propulsion.

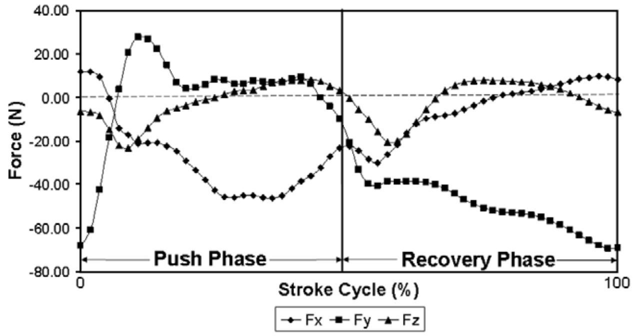

Fig. 2. Representative force data during one stroke of wheelchair propulsion.

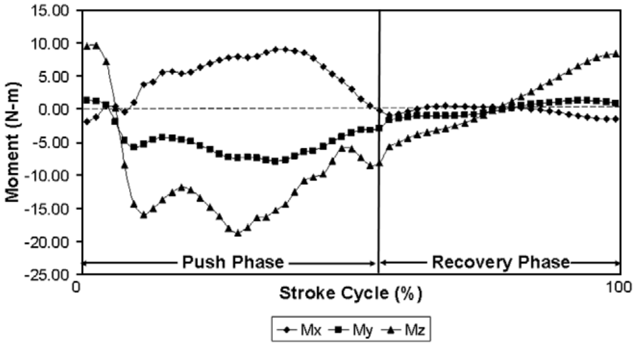

Fig. 3. Representative moment data during one stroke of wheelchair propulsion.

A multivariable analysis was performed to test the influence of speed, subject age, and subject mass on the biomechanical variables. Table 3 summarizes the unadjusted peak forces and moments of the non-dominant shoulder during the push phase of propulsion for the two speed conditions. The *P*-value ($P < 0.01$ for all conditions) describes the significance of speed in the multivariable model. Also reported is stroke frequency and mean velocity. As expected, speed significantly influenced all of the biomechanical variables. Subjects used larger forces and moments, as well as increased stroke frequency, to propel at the higher velocity.

Table 3
Raw mean of biomechanical variables at two speeds of manual wheelchair propulsion

	0.9 m/s ($n = 31$)	1.8 m/s ($n = 31$)
	Mean (SD)	Mean (SD)
Posterior force (N)	46.5 (17.5)	58.8 (15.3)*
Superior force (N)	13.7 (18.0)	42.3 (21.7)*
Lateral force (N)	21.9 (8.0)	35.0 (10.9)*
Abduction moment (N-m)	4.8 (2.7)	7.0 (3.3)*
Internal rotation moment (N-m)	7.2 (0.54)	9.2 (3.7)*
Extension moment (N-m)	17.6 (7.6)	22.8 (8.6)*
Stroke frequency (1/s)	0.96 (0.18)	1.30 (0.24)*
Mean velocity (m/s)	1.01 (0.15)	1.65 (0.30)*

* $P < 0.01$ for all variables.

3.4. Shoulder biomechanics and demographics

Based on the results of the multivariable analysis described in Section 3.3, subject age did not significantly influence shoulder force and moments, however it was associated with stroke frequency ($\beta = 0.009$, $P = 0.006$) and mean velocity ($\beta = -0.005$, $P = 0.07$). Older individuals propelled with a higher cadence, but still maintained a slower velocity. Subject mass was significantly associated with posterior force ($\beta = 0.22$, $P = 0.0007$), lateral force ($\beta = 0.009$, $P = 0.006$), internal rotation moment ($\beta = 0.03$, $P = 0.02$), and extension moment ($\beta = 0.11$, $P = 0.0009$). These relationships imply that increased subject mass results in a higher shoulder joint force or moment in any of these directions.

3.5. Shoulder biomechanics and shoulder pathology

Peak shoulder kinetics at the two speeds were averaged to create a single force or moment variable in each direction. These biomechanical variables were examined for their relationship to individual MRI measures and physical exam score (both dichotomized) while controlling for subject age and mass. Subjects who experienced a higher posterior force had a significantly higher prevalence of coracoacromial ligament edema (OR = 1.29, $P = 0.03$). Individuals who demonstrated higher lateral forces during propulsion were more likely to have CA ligament edema (OR = 1.35, $P = 0.047$) and CA ligament thickening (OR = 4.35, $P = 0.045$). Increased internal rotation moment increased the odds of showing signs of shoulder pathology during the physical exam (OR = 1.61 $P = 0.02$). Other biomechanical variables showed trending ($0.05 \leqslant P < 0.10$) associations with shoulder pathology measures. Peak superior force was associated with pathology findings during the physical exam (OR = 1.05, $P = 0.09$). Individuals experiencing a larger abduction moment were more likely to exhibit CA ligament thickening (OR = 1.58, $P = 0.06$). Finally, subjects who experienced higher extension moments were more likely to have CA ligament edema (OR = 1.35, $P = 0.09$).

4. Discussion

To our knowledge, this is largest study to investigate shoulder kinetics during wheelchair propulsion, and the first study to examine the relationship between kinetics and pathology. Our kinetic results are in general agreement with previous studies (Finley et al., 2004; Kulig et al., 1998; Mulroy et al., 2005). Kulig et al. reported significant increases in shoulder forces and moments with faster speed (Kulig et al., 1998). They also observed the same directional force and moment components during the push phase of propulsion. Direct comparison is difficult because of differences in laboratory equipment, speed conditions, and computational techniques. Our study is the only one to test individuals in their own wheelchair, aside from other studies completed in our laboratory.

Similar to results published for a smaller subject group (Boninger et al., 2001), a large percentage of participants in this study had MRI findings indicative of shoulder pathology. All but one subject presented at least one abnormal finding on MRI. Increased body mass (Boninger et al., 1999, 2003) and age (Escobedo et al., 1997; Lal, 1998) have previously been associated with upper limb pathology so we included these subject characteristics in our analysis. Heavier body mass was found to increase the odds of having AC joint edema or CA ligament thickening. Only one subject in this study had a rotator cuff tear, compared to 57% of individuals in a previous study. The most likely difference between this study and that of Escobedo et al. is differences in the subject populations. The average age of the subjects in the Escobedo study was 59 years old, and the average years since injury was 26. Subjects in our current study were younger (mean = 38 years old) and their injuries were more recent (mean = 12 years since injury). Approximately thirty percent of subjects in this study experienced pain or discomfort during the physical exam. These subjects tended to be older and weigh more.

Subject demographics were also associated with shoulder biomechanical variables. In particular, body mass was associated with higher forces (posterior and lateral) and moments (internal rotation and extension) during the push phase of propulsion. Posterior force and extension moment are a direct result of the tangential force applied to the pushrim required to propel the wheelchair forward. These are also the largest directional components of the forces and moments. Increased body mass increases the rolling resistance of the wheelchair, requiring the manual wheelchair user to apply more force to the pushrim to maintain the same speed. Age was not significantly associated with shoulder forces and moments, however older individuals used a higher stroke frequency and still went slower than younger individuals.

When controlling for age and body mass, we identified relationships between shoulder kinetics and specific shoulder pathology. Significant MRI measures were localized to the coracoacromial arch and included coracoacromial (CA) ligament edema, and CA ligament thickening. In particular, subjects who experienced higher posterior force, lateral force, or extension moment during propulsion were more likely to exhibit CA ligament edema. Individuals who displayed larger lateral forces or abduction moments were more likely to have CA ligament thickening. It should be noted that pathological findings on MRI are not necessarily symptomatic. However, CA ligament thickening in particular may contribute to narrowing of the supraspinatus outlet and has been associated with surgically confirmed rotator cuff tears (Farley et al., 1994). Additionally, our goal is to be able to identify risk factors for clinical pathology (i.e., MRI measures) before the individual becomes symptomatic or develops a more serious injury. Only four of the subjects in this study had experienced pain while propelling their wheelchair in the month prior to their participation. This indicates that the associations between shoulder kinetics and pathology found in this study may be independent of pain. Once radiological shoulder pathology becomes painful, an individual may modify their propulsion biomechanics to improve comfort during propulsion. This study was able to identify relationships between shoulder kinetics and clinical pathology in a population with a low incidence of pain. Future studies may need to recruit a greater number of subjects and collect more detailed pain information, such as the Wheelchair User's Shoulder Pain Index (Curtis et al., 1995), in order to investigate how pain may be related to shoulder kinetics and pathology.

Subjects who experienced higher superior forces and internal rotation moments at the shoulder during propulsion showed signs of shoulder pathology during the physical exam. The superior force and internal rotation moment both contribute in different ways to impingement syndrome, which can be identified through the physical exam. An imbalance of internal and external rotators of the shoulder is a risk factor for impingement (McMaster et al., 1991). Individuals who consistently propel with higher internal rotation moments at the shoulder may develop such an imbalance. A superior force at the shoulder drives the humerus up towards the acromion which may compress the rotator cuff and result in impingement (Kulig et al., 1998; Reyes et al., 1995).

It should be noted that the joint moments presented in this study are referenced to the trunk coordinate system, and not the humerus. This was done to maintain consistency in reporting shoulder joint forces and moments in reference to the same local coordinate system. Our model does not account for the coupling between humeral position and moment direction. Specifically, as abduction approaches 90°, a moment described as internal/external rotation in our study, may actually be more accurately described as horizontal adduction/abduction. Previous studies have reported the maximum abduction angle reached during the push phase of propulsion to be less than 40°, which lessens the effect of coupling on the results presented in this paper (Cooper et al., 1999; Finley et al., 2004).

All associations between shoulder joint pathology and shoulder forces and moments were in the same direction; people who experienced higher shoulder forces and moments were more likely to exhibit shoulder pathology. While we were able to identify specific force and moments that related to shoulder pathology, in the bigger picture this may reflect a need to reduce the overall joint loading experienced during propulsion. Higher shoulder joint forces and moments are a result of increased force application at the pushrim to propel the wheelchair as well as inertial effects due to arm motion. Therefore, one way to reduce shoulder joint forces and moments would be to reduce the overall amount of force required to propel the wheelchair.

One limitation of this study is that all subjects propelled on a dynamometer roller system which is a simulated propulsion environment. However, this setup allowed us to collect stable kinetic and kinematic data and to control subject velocity to two target speeds using visual feedback that is directly linked to the dynamometer system. This setup has been fully characterized and used in a number of studies (DiGiovine et al., 2001). Additional work should be done to collect biomechanical data during real-world propulsion over various surfaces. This study also had a relatively small sample size, however we were still able to find an association between shoulder pathology, subject characteristics, and shoulder joint kinetics. Again, this study is the largest investigation of shoulder biomechanics during wheelchair propulsion to date. We are currently participating in a multi-site longitudinal study of manual wheelchair users with paraplegia (Boninger et al., 2005). The data collected in this study will allow us to conduct additional investigation of the relationship between shoulder kinetics and pathology with a larger sample size.

Although not studied in this paper, interventions need to be developed to reduce these force and moment components, as well as the overall force required to propel a wheelchair. For instance, MWUs should be provided with the lightest weight possible adjustable wheelchairs. A very light wheelchair will reduce the rolling resistance of the system and decrease the required propulsive force. Axle height (Boninger et al., 2005) and fore-aft (Boninger et al., 2000) position can be adjusted to maximize access to the pushrim and reduce stroke cadence. Ergonomic interventions, such as modifying the wheelchair backrest or pushrims, could also be explored for their effect on joint forces during propulsion. Manual wheelchair users should also be advised of the association between body mass and shoulder pathology. Weight loss could be a complementary strategy to other interventions that reduce wheelchair propulsive force. However, we propose that changes in wheelchair setup or propulsion technique may be an alternative, or complement, to losing weight. For some individuals, alternative means of mobility including pushrim activated power assist or powered wheelchairs, should be explored. We were able to find significant relationships and trends between shoulder kinetics and pathology among the 33 subjects recruited for this study, a larger sample size would provide more statistical power to describe the association of shoulder kinetics and pathology.

5. Conclusions

Individuals who experienced higher directional shoulder forces and moments were more likely to exhibit coracoacromial arch pathology on MRI and discomfort during a physical exam. Therefore, researchers and clinicians should work to reduce the forces and moments experienced at the shoulder during wheelchair propulsion. The potential exists to modify wheelchair design and set up or propulsion biomechanics in order to reduce the force required to propel the wheelchair, thereby reducing the loading experienced at the shoulder.

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References

Andersen, J., Kaergaard, A., Frost, P., Thomsen, J., Bonde, J., Fallentin, N., Borg, V., Mikkelsen, S., 2002. Physical, psychosocial, and individual risk factors for neck/shoulder pain with pressure tenderness in the muscles among workers performing monotonous, repetitive work. Spine 27 (6), 660–667.

Andersen, J., Kaergaard, A., Mikkelsen, S., Jensen, U., Frost, P., Bonde, J., Fallentin, N., Thomsen, J., 2003. Risk factors in the onset of neck/shoulder pain in a prospective study of workers in industrial and service companies. Occupational and Environmental Medicine 60, 649–654.

Ballinger, D., Rintala, D., Hart, K., 2000. The relation of shoulder pain and range-of-motion problems to functional limitations, disability, and perceived health of men with spinal cord injury: a multifaceted longitudinal study. Archives of Physical Medicine and Rehabilitation 81, 1575–1581.

Bayley, J.C., Cochran, T.P., Sledge, C.B., 1987. The weight-bearing shoulder. The impingement syndrome in paraplegics. Journal of Bone and Joint Surgery—American Volume 69, 676–678.

Boninger, M.L., Cooper, R.A., Baldwin, M.A., Shimada, S.D., Koontz, A.M., 1999. Wheelchair pushrim kinetics: body weight and median nerve function. Archives of Physical Medicine and Rehabilitation 80, 910–915.

Boninger, M., Baldwin, M.A., Cooper, R., Koontz, A., Chan, L., 2000. Manual wheelchair pushrim biomechanics and axle position. Archives of Physical Medicine and Rehabilitation 81 (5), 608–613.

Boninger, M.L., Towers, J.D., Cooper, R.A., Dicianno, B.E., Munin, M.C., 2001. Shoulder imaging abnormalities in individuals with paraplegia. Journal of Rehabilitation R&D 38 (4), 401–408.

Boninger, M.L., Dicianno, B.E., Cooper, R.A., Towers, J.D., Koontz, A.M., Souza, A.L., 2003. Shoulder magnetic resonance imaging abnormalities, wheelchair propulsion and gender. Archives of Physical Medicine and Rehabilitation 84, 1615–1620.

Boninger, M., Koontz, A., Sisto, S., Dyson-Hudson, T., Chang, M., Price, R., Cooper, R., 2005. Pushrim biomechanics and injury prevention in spinal cord injury: recommendations based on CULP-SCI investigations. Journal of Rehabilitation Research and Development 42 (3, Suppl. 1), 9–20.

Carson, R., 1994. Reducing cumulative trauma disorders: use of proper workplace designs. American Association of Occupational Health Nurses Journal 42, 270–276.

Chatterjee, D., 1992. Workplace upper limb disorders: a prospective study with intervention. Occupational Medicine 42, 129–136.

Cooper, R.A., Robertson, R.N., VanSickle, D.P., Boninger, M.L., Shimada, S.D., 1997. Methods for determining three-dimensional wheelchair pushrim forces and moments— a technical note. Journal of Rehabilitation Research and Development 34 (2), 162–170.

Cooper, R., Boninger, M., Shimada, S., Lawrence, B., 1999. Glenohumeral joint kinematics and kinetics for three coordinate system representations during wheelchair propulsion. Amercian Journal of Physical Medicine and Rehabilitation 78 (5), 435–446.

Curtis, K.A., Roach, K.E., Applegate, E.B., Amar, T., Benbow, C.S., Genecco, T.D., Gualano, J., 1995. Development of the Wheelchair User's Shoulder Pain Index (WUSPI). PARAPLEGIA 33 (5), 290–293.

DiGiovine, C.P., Cooper, R.A., Boninger, M.L., 2001. Dynamic calibration of a wheelchair dynamometer. Journal of Rehabilitation Research and Developement 38 (1), 41–55.

Dyson-Hudson, T., Kirshblum, S., 2004. Shoulder pain in chronic spinal cord injury, Part I: Epidemiology, etiology, and pathomechanics. The Journal of Spinal Cord Medicine 27 (1), 4–17.

Escobedo, E., Hunter, J., Hollister, M., Patten, R., Goldstein, B., 1997. MR imaging of rotator cuff tears in individuals with paraplegia. American Journal of Roentology 168, 919–923.

Farley, T., Neumann, C.H., Steinbach, L.S., Petersen, S.A., 1994. The coracoacromial arch: MR evaluation and correlation with rotator cuff pathology. Skeletal Radiology 23 (8), 641–645.

Finley, M., Rasch, E., Keyser, R., Rodgers, M., 2004. The biomechanics of wheelchair propulsion in individuals with and without upper-limb impairment. Journal of Rehabilitation Research and Development 41 (3B), 395–402.

Frost, P., Bonde, J., Mikkelsen, S., Andersen, J., Fallentin, N., Kaergaard, A., Thomsen, J., 2002. Risk of shoulder tendinitis in relation to shoulder loads in monotonous repetitive work. American Journal of Industrial Medicine 41, 11–18.

Hanavan, E., 1964. A mathematical model of the human body, Aerospace Medical Research Laboratories, Wright–Patterson Air Force Base, Ohio, AMRL-TR-64-102.

Hosmer, D., Lemeshow, S., 1989. Applied Logistic Regression. John Wiley and Sons, New York.

Kjellin, I., Ho, C.P., Cervilla, V., Haghighi, P., Kerr, R., Vangness, C.T., Trudell, D., Resnick, D., 1991. Alterations in the supraspinatus tendon at MR imaging: correlation with histopathologic findings in cadavers. Radiology 181, 837–841.

Kulig, K., Rao, S., Mulroy, S., Newsam, C.J., Gronley, J.K., Bontrager, E.L., Perry, J., 1998. Shoulder joint kinetics during the push phase of wheelchair propulsion. Clinical Orthopaedics and Related Research 354 (1), 132–143.

Lal, S., 1998. Premature degenerative shoulder changes in spinal cord injury patients. Spinal Cord 36, 186–189.

Leclerc, A., Chastang, J.-F., Niedhammer, I., Landre, M.-F., Roquelaure, Y. Study Group on Repetitive Work, 2004. Incidence of shoulder pain in repetitive work. Occupational and Environmental Medicine 61, 39–44.

Liou, J.T., Wilson, A.J., Totty, W.G., Brown, J.J., 1993. The normal shoulder: common variations that simulate pathologic conditions at MR imaging. Radiology 186, 435–441.

Little, R., Rubin, D., 2002. Statistical analysis with missing data, second ed. Wiley, New York.

Lundqvist, C., Siosteen, A., Blomstrand, C., Lind, B., Sullivan, M., 1991. Spinal cord injuries. Clinical, functional, and emotional status. Spine 16 (1), 78–83.

McMaster, W., Long, S., Caiozzo, V., 1991. Isokinetic torque imbalances in the rotator cuff of the elite water polo player. The American Journal of Sports Medicine 19 (1), 72–75.

Mulroy, S., Newsam, C.J., Gutierrez, D., Requejo, P., Gronley, J.K., Haubert, L.L., Perry, J., 2005. Effect of fore-aft seat position on shoulder demands during wheelchair propulsion: Part 1. A Kinetic Analysis. The Journal of Spinal Cord Medicine 28 (3), 214–221.

Nichols, P.J., Norman, P.A., Ennis, J.R., 1979. Wheelchair user's shoulder? Shoulder pain in patients with spinal cord lesions. Scandinavian Journal of Rehabilitation Medicine 11, 29–32.

NIOSH 1997. Shoulder musculoskeletal disorders: evidence for work-relatedness, in musculoskeletal disorders and workplace factors: a critical review of epidemiology for work related musculoskeletal disorders of the neck, upper extremity, and low back, National Institute for Occupational Safety and Health, Publications Dissemination, Cincinnati, OH, pp. 122–194.

Orgel, D., Milliron, M., Frederick, L., 1992. Musculoskeletal discomfort in grocery express checkstand workers an ergonomic intervention study. Journal of Occupational Medicine 34 (8), 815–818.

Pentland, W.E., Twomey, L.T., 1991. The weight-bearing upper extremity in women with long term paraplegia. PARAPLEGIA 29, 521–530.

Reyes, M., Gronley, J., Newsam, C., Mulroy, S., Perry, J., 1995. Electromyographic analysis of shoulder muscles of men with low-level paraplegia during a weight relief raise. Archives of Physical Medicine and Rehabilitation 76, 433–439.

Rodgers, M., Gayle, G.W., Figoni, S., Kobayashi, M., Lieh, J., Glaser, R., 1994. Biomechanics of wheelchair propulsion during fatigue. Archives of Physical Medicine and Rehabilitation 75, 85–93.

Sie, I.H., Waters, R.L., Adkins, R.H., Gellman, H., 1992. Upper extremity pain in the postrehabilitation spinal cord injured patient. Archives of Physical Medicine and Rehabilitation 73, 44–48.

Subbarao, J., Klopfstein, J., Turpin, R., 1995. Prevalence and impact of wrist and shoulder pain in patients with spinal cord injury. The Journal of Spinal Cord Medicine 18 (1), 9–13.

Wu, G., van der Helm, F., Veeger, H.E.J., Makhsous, M., Van Roy, P., Anglin, C., Nagels, J., Karduna, A., McQuade, K., Wang, X., Werner, F., Bucholz, B., 2005. ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion—Part II: shoulder, elbow, wrist and hand. Journal of Biomechanics 38, 981–992.