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Variability of peak shoulder force during wheelchair propulsion in manual wheelchair users with and without shoulder pain

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ABSTRACT

Background: Manual wheelchair users report a high prevalence of shoulder pain. Growing evidence shows that variability in forces applied to biological tissue is related to musculoskeletal pain. The purpose of this study was to examine the variability of forces acting on the shoulder during wheelchair propulsion as a function of shoulder pain.

Methods: Twenty-four manual wheelchair users (13 with pain, 11 without pain) participated in the investigation. Kinetic and kinematic data of wheelchair propulsion were recorded for 3 min maintaining a constant speed at three distinct propulsion speeds (fast speed of 1.1 m/s, a self-selected speed, and a slow speed of 0.7 m/s). Peak resultant shoulder forces in the push phase were calculated using inverse dynamics. Within individual variability was quantified as the coefficient of variation of cycle to cycle peak resultant forces.

Findings: There was no difference in mean peak shoulder resultant force between groups. The pain group had significantly smaller variability of peak resultant force than the no pain group ($P<0.01$, $\eta^2=0.18$).

Interpretation: The observations raise the possibility that propulsion variability could be a novel marker of upper limb pain in manual wheelchair users.

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1. Introduction

At least 2.6% (6.8 million) of the US population use assistive devices and nearly a quarter of those using assistive devices utilize a manual wheelchair for mobility (Laplante and Kaye, 2010). Manual wheelchair users depend on their upper limbs for mobility and most functional activities. Unfortunately, the human upper limb is not specialized for the repetitive loading required for wheelchair propulsion. This requirement predisposes manual wheelchair users to upper limb pathology. Indeed, up to 70% of manual wheelchair users report upper limb pain (Finley et al., 2004), which is mainly manifested in the shoulder (Curtis et al., 1999) and wrist (Gellman et al., 1988). Furthermore, even in manual wheelchair users who do not report pain, there is evidence of degenerative changes in the shoulder (Mercer et al., 2006) suggesting that it is just a matter of time before these asymptomatic individuals will experience pain.

Upper limb pain in manual wheelchair users has been linked to difficulty performing activities of daily living, decreased physical activity and decreased quality of life (Gutierrez et al., 2007). Overall, any loss of upper limb function due to pain adversely impacts the independence and mobility of manual wheelchair users. It has been speculated that a decrease in independence and mobility results in greater health care costs and an increased risk for secondary morbidity (cardiovascular disease, obesity, etc.) (Hardy et al., 2011). Subsequently, it is imperative to understand the mechanisms that contribute to upper limb pathology in manual wheelchair users so that appropriate interventions can be developed to prevent or minimize the effect of pain on function and reduce the risk of long-term upper extremity disability.

It is frequently speculated that propulsion biomechanics contributes to the pathogenesis of shoulder pathology (Desroches et al., 2008; Koontz et al., 2002; Kulig et al., 1998; Mercer et al., 2006). This speculation is based on simple Newtonian mechanics that forces applied to the wheelchair hand rim resulted in reactive forces acting on the shoulder that may over time lead to musculoskeletal damage. Indeed, research has shown that greater tangential forces acting on the hand rim generates greater net moment at the shoulder and is indicative of higher risk of shoulder injury (Desroches et al., 2008; Koontz et al., 2002). Investigations utilizing inverse dynamics have revealed that posterior and superior forces are the main forces applied to the shoulder during the push phase (Collinger et al., 2008; Gil-Agudo et al., 2010; Koontz et al., 2002; Mercer et al., 2006; Van Drongelen et al., 2005). It is proposed that greater posterior forces might be associated with coracoacromial ligament edema (Mercer et al., 2006) while greater inferior force might cause compression of the rotator cuff leading

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to shoulder impingement (Koontz et al., 2002; Kulig et al., 1998; Mercer et al., 2006).

Despite the logic of these claims, there has been limited evidence of a relation between forces acting on the shoulder during propulsion and shoulder pain or pathology (Collinger et al., 2008; Mercer et al., 2006). For instance, Collinger et al. (2008) found that there was no difference in mean of peak propulsion force acting on the shoulder between persons with and without shoulder pain.

A potential limitation of the previous research is that it has almost exclusively focused on mean propulsion parameters. This approach essentially ignores the importance of movement variability as a factor related to musculoskeletal injury. There is growing evidence from experimental and clinical studies that motor variability is related to musculoskeletal injury and pain (Srinivasan and Mathiassen, 2012). These investigations have suggested two theories regarding pain and reduced movement variability. Some research has suggested that participants with pain implement an adaptive strategy of less variable motion to avoid pain (Hamill et al., 1999; Heiderscheit et al., 2002; Madeleine and Madsen, 2009; van den Hoorn et al., 2012). There is also evidence that less motor variability may actually contribute to the development of chronic pain (James, 1996; Madeleine et al., 2008; Mathiassen et al., 2003).

To date there has been minimal examinations of variability in manual wheelchair propulsion and its association to shoulder pain. The main goal of this investigation was to examine the relationship between shoulder pain and variability of peak force on shoulder during wheelchair propulsion. It was hypothesized that variability of peak shoulder resultant force will be smaller in individuals with pain than individuals with no pain.

2. Methods

2.1. Participants

Twenty-four manual wheelchair users (10 females, 14 males, age = 24.3 (SD: 10.1) years) who resided in the local community and who were recruited through advertisements participated in the investigation. Inclusion criteria included (1) more than one year of manual wheelchair experience; (2) use of a manual wheelchair for greater than 80% of their daily mobility; and (3) between 18 and 64 years of age. All procedures were approved by local institutional review board. Participants were separated into groups based on their self-report of current shoulder pain (pain = 13, no pain = 11).

2.2. Procedures

Upon arrival at the laboratory participants were informed of the research procedures, were given an opportunity to ask questions concerning the research and were then asked to provide written informed consent. After providing informed consent, participants provided demographic information and self-reported current shoulder pain. Participants also completed the Wheelchair User's Shoulder Pain Index (WUSPI). The WUSPI was designed to measure the severity of shoulder pain related to functional activity of manual wheelchair users in daily living (Curtis et al., 1995). It is composed of 15 items relating to pain in everyday activities of wheelchair users including loading a wheelchair into a car, transferring, etc. Participants reported their pain during each functional activity between zero and ten with a higher score indicating greater pain. Overall WUSPI score is the sum of the 15 items with a maximum possible score of 150. This measure was found to be valid and reliable in manual wheelchair users (Curtis et al., 1995).

Participants were asked to propel at constant speeds of 1.1 m/s (fast speed condition), 0.7 m/s (slow speed condition), as well as a self-selected speed, for 3 min. The sequence of trial at different speeds was randomized for each subject. A speedometer was used to provide real-time velocity feedback to the participants during the three minute propulsion trials. The self-selected speed was determined by asking the subject to push on the roller at a comfortable pace, as if they were pushing in a hallway. When the speed reached a steady state, the speed was recorded as the self-selected speed. Speed feedback was not given to the subject during this process of self-selecting a comfortable speed.

2.3. Data collection and instrumentation

2.3.1. Anthropometric data

Based on previous research (Collinger et al., 2008), segment length and upper extremity circumferences of all participants were measured. The anatomical measures were used as input to Hanavan's mathematical model for human anthropometry (Hanavan, 1964) which calculated the inertial properties of each body segment used in the inverse dynamic model. Body weight was measured with the use of a calibrated force platform (AMTI, Inc., Watertown, MA, USA).

2.3.2. Kinetic data

Each participant's wheelchair was fitted bilaterally with SMARTwheels (Three Rivers Holdings, LLC, Mesa, AZ, USA) and secured to a single roller dynamometer system using a four-point tie-down system and a flywheel system (Digiovine et al., 2001) which has been suggested to be similar to over ground propulsion (Koontz et al., 2012). The SMARTwheels measures three-dimensional forces and torques applied to the push rim. Attaching the SMARTwheels to the subject's own wheelchair does not change the wheel placement alignment or camber (Mercer et al., 2006). All subjects acclimated themselves to the dynamometer setup prior to testing. Kinetic data were collected at 100 Hz and digitally filtered with an eighth-order, zero-phase, low-pass Butterworth filter with 10 Hz cutoff frequency (Collinger et al., 2008). The initial 10 cycles of each trial were removed in order to ensure that steady state performance was analyzed. A total of 65 propulsion cycles from each individual trial was included in data analysis. This number of cycles was based on the minimum number produced by the participants.

2.3.3. Kinematic data

A 10 camera motion capture system (Raptor Digital RealTime System, Motion Analysis Co., Santa Rosa, CA, USA) was used to collect kinematic data by tracking attached reflective markers on the participant's upper body bony landmarks. Based on previous research (Collinger et al., 2008), 18 markers were attached bilaterally, at specific bony landmarks on the following locations: third metacarpophalangeal joint, radial styloid, ulnar styloid, olecranon process, lateral epicondyl, acromion, sternal notch, C7 vertebrae, T3 vertebrae, T6 vertebrae and jaw. The kinematic data were collected at 100 Hz and digitally filtered with fourth-order, zero-phase, low-pass Butterworth filter with 7 Hz cutoff frequency (Collinger et al., 2008).

2.4. Data analysis

2.4.1. Propulsion speed

Actual propulsion speed at each speed condition (fast, slow, and self-selected) was determined by SMARTwheel software.

2.4.2. Definition of push and recovery phase

Each cycle was defined to consist of push and recovery phases. Propulsive moment on hand rim was calculated by the SMARTwheels software. The onset of push phase was defined as the point at which the propulsive moment applied to the push rim deviated from baseline by 5% (Koontz et al., 2002; Kwarciak et al., 2009). The end of push phase was defined as the point when the propulsive moment returned to baseline and remained within 5% (Koontz et al., 2002; Kwarciak et al., 2009) (Fig. 1). Consistent with previous research, only the push phase

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