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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.

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

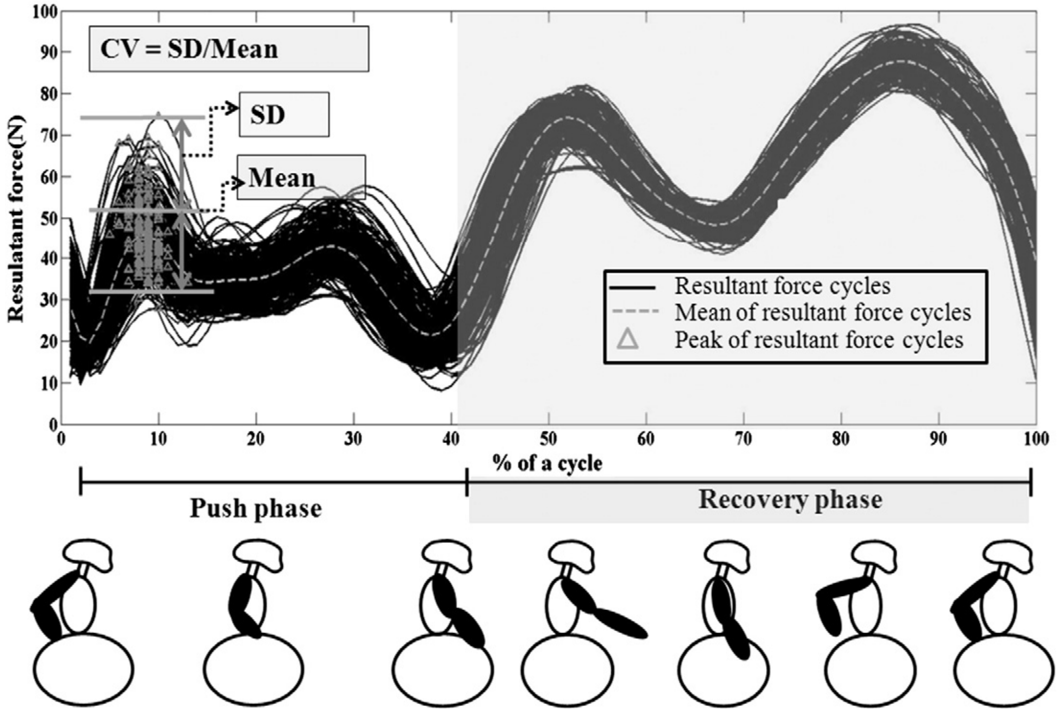

Fig. 1. Cycle to cycle shoulder resultant force as a function of propulsion phase for a single individual over a 3 minute trial at fast speed.

of propulsion was analyzed (Collinger et al., 2008; Koontz et al., 2002; Kulig et al., 2001; Mercer et al., 2006).

2.4.3. Inverse dynamics and shoulder forces

We used an inverse dynamic model described previously (Cooper et al., 1999). The model was used to compute subject-specific 3-D joint forces and moments from segment kinematics, forces acting on the push rim, and subject anthropometrics. Custom developed MATLAB code (The MathWorks, Natick, MA, USA) was used to estimate 3-D and resultant shoulder joint forces using the inverse dynamic analysis. Peak value of shoulder resultant force during the push phase of each cycle was identified for each test condition.

2.4.4. Variability of peak shoulder resultant force

For each individual participant, the mean, standard deviation (SD), and coefficient of variation (CV = SD/Mean) of peak values of shoulder resultant force in push phase over every stroke was calculated (Fig. 1).

2.4.5. Statistical analysis

Statistical analysis was performed using SPSS for Windows, version 19.0 (IBM, Inc., Chicago, IL, USA). Normality of data was tested using Shapiro–Wilk test. Pain group differences in normally distributed demographic variables were assessed with independent t-test, and non-normally distributed variables were analyzed with independent-samples Mann–Whitney U Tests.

Actual propulsion speed, mean, SD and CV of peak values of shoulder resultant force during the push phase were each assessed relative to pain group and propulsion speed using multivariate analysis of covariance (MANCOVA) with weight as a covariant when appropriate Bonferroni correction was used for post-hoc analysis. Weight was entered as a covariant since it has consistently been found to influence peak shoulder forces during wheelchair propulsion (Collinger et al., 2008; Koontz et al., 2002; Mercer et al., 2006;). Significance was set at ≤0.05.

3. Results

3.1. Demographics

Demographic characteristics of the participants are reported in Table 1. There was no difference between groups in gender composition [$P = 0.64$], age [$P = 0.20$], weight [$t(22.0) = -1.52$, $P = 0.14$], and years of wheelchair usage [$P = 0.17$]. Spinal cord injury (pain = 38%, no pain = 55%) and spina bifida (pain = 38%, no pain = 27%) were the predominant types of disability in both groups. Per design, the pain group had significantly greater shoulder pain than the no pain group as indicated by higher WUSPI scores [$t(14.1) = -3.10$, $P < 0.01$].

3.2. Propulsion speed

A main effect of speed condition on propulsion speed was observed [$F(2, 65) = 67.4$, $P < 0.01$, $\eta^2 = 0.68$]. Post-hoc analysis revealed that each speed condition was distinct (P's ≤ 0.01). However, there was no pain group effect ($F(1,65) = 0.03$, $P = 0.87$, $\eta^2 = 0.00$) nor an interaction

Table 1
Participant demographics as a function of pain group.

	Pain	No pain	*P*-value
n	13	11	–
Gender	Male = 7 Female = 6	Male = 7 Female = 4	0.64
Injury type	Spinal cord = 5 (T1–T12 = 4) (L1–L5 = 1) Spina bifida = 5 Amputation = 1 Spinal cyst = 1 Sacral agenesis = 1	Spinal cord = 6 (T1–T12 = 6) Spina bifida = 3 Amputation = 1 Arthrogryposis = 1	–
Age (year)	28 (13)	20 (2)	0.20
Weight (kg)	75.5 (25.4)	61.8 (17.2)	0.14
Wheelchair usage (year)	17.2 (10.2)	12.3 (4.9)	0.17
WUSPI	22.8 (21.3)	3.7 (5.9)	<0.01*

Note: Values are mean (standard deviation). * indicates $p < 0.05$.

between pain group and speed ($F(2,65) = 0.16$, $P = 0.86$, $\eta^2 = 0.00$; see Table 2).

3.3. Peak resultant force

Statistical analysis revealed that there was no significant difference in mean peak resultant force between pain (with and without) groups [$F(1, 65) = 2.21$, $P = 0.14$, $\eta^2 = 0.03$] (Fig. 2). No other significant effects or interactions were observed ($Ps > 0.05$).

Descriptive analysis revealed that SD was 19%, 12% and 12% less in the pain group than the no pain group in slow, self-selected and fast speed conditions, respectively (Fig. 3). Statistical analysis found that there was a trend towards significant differences between the pain and the no pain group [$F(1, 65) = 3.62$, $P = 0.06$, $\eta^2 = 0.05$]. No other significant effects or interactions were observed ($Ps > 0.05$).

Descriptive analysis found that CV was 21%, 18% and 17% less in the pain group than the no pain group in slow, self selected and fast speed conditions, respectively (Fig. 4). Statistical analysis revealed that there was significantly less CV in the pain group than the no pain group [$F(1, 65) = 14.3$, $P < 0.01$, $\eta^2 = 0.18$]. No other significant effects or interactions were observed ($Ps > 0.05$).

4. Discussion

The purpose of this investigation was to examine the relationship between shoulder pain in manual wheelchair users and variability of peak resultant force on the shoulder during wheelchair propulsion. Overall it was found that shoulder pain was associated with smaller variability of peak shoulder resultant force across a range of propulsion speeds. This study is one of the first to demonstrate an association between symptomatic shoulder pain and variability in wheelchair propulsion biomechanics.

While shoulder pain is quite common in manual wheelchair users and propulsion biomechanics are believed to play a significant role in the development of upper extremity pain (Laplante and Kaye, 2010), no previous research has documented an association between forces acting on the shoulder and symptomatic shoulder pain (Collinger et al., 2008; Mercer et al., 2006). It is important to note that these previous investigations have focused on *mean* propulsion parameters. Consistent with these studies, there was no difference in mean peak shoulder resultant force as a function of shoulder pain observed here. Yet, an examination of variability of peak shoulder resultant forces revealed significant difference between those with and without pain. The current observation highlighted that movement variability in and of itself is a sensitive marker of musculoskeletal pain in manual wheelchair users.

Although this association between variability and self-reported pain is novel within wheelchair propulsion research, it is consistent with motor control/biomechanics research that has demonstrated that variability can play a functional role in the prevention or development of injury (James, 1996; Srinivasan and Mathiassen, 2012). There are at least two potential reasons that the no pain group demonstrated greater variability in peak shoulder force than the pain group (Srinivasan and Mathiassen, 2012). First, it is possible that the presence of shoulder pain caused individuals to constrain their shoulder movement in order to avoid pain which resulted in decreased variability of peak shoulder forces. Research has demonstrated that chronic musculoskeletal pain is

Table 2
Propulsion speed as a function of pain group and target propulsion speed.

	Overall (m/s)	Pain (m/s)	No pain (m/s)
Slow speed (0.7 m/s)	0.73 (0.04)	0.73 (0.04)	0.73 (0.04)
Self selected speed	0.94 (0.19)	0.94 (0.15)	0.93 (0.24)
Fast speed (1.1 m/s)	1.14 (0.06)	1.13 (0.05)	1.16 (0.07)

Note: Values are mean (standard deviation).

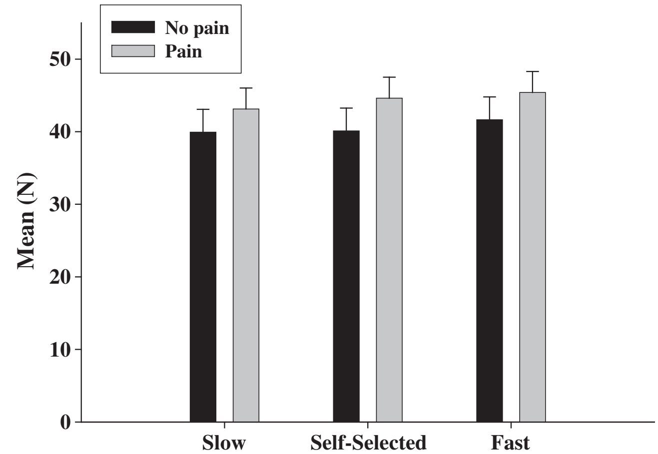

Fig. 2. Mean peak shoulder resultant force as a function of group and propulsion speed.

associated with decreased motor variability in several repetitive motor tasks (Hamill et al., 1999; Heiderscheit et al., 2002; Madeleine and Madsen, 2009; van den Hoorn et al., 2012).

Secondly, it is possible that lower amounts of variability of peak shoulder force could be an underlying mechanism that led to the development of shoulder pain by demanding relatively constant load acting on the shoulder. It has been suggested that a lack of motor variability coincides with a relatively constant force being applied to musculoskeletal tissue and ultimately results in chronic overuse injury (James, 1996; Srinivasan and Mathiassen, 2012). However, due to the cross-sectional nature of this study, no conclusion regarding the directional association between peak shoulder force variability and shoulder pain in manual wheelchair users can be made.

Another unresolved question is where the variability in shoulder kinetics stems from. Within the current investigation, an inverse dynamic model, which depended on the forces acting on the push rim and kinematic data of the upper limbs, was used to determine net resultant shoulder forces. Consequently, it is possible that the variability in resultant force comes from either fluctuations in forces at the hand rim, upper limb movements or a combination of the two. Preliminary data from our laboratory indicates lower variability of forces at the hand rim in persons self-reporting shoulder pain compared to those without shoulder pain during wheelchair propulsion (Rice et al., 2012). Future research needs to examine the underlying contribution to variability in resultant shoulder forces. This is not only of theoretical importance but also has rehabilitative implications. Simply put, the factors driving resultant force variability could be

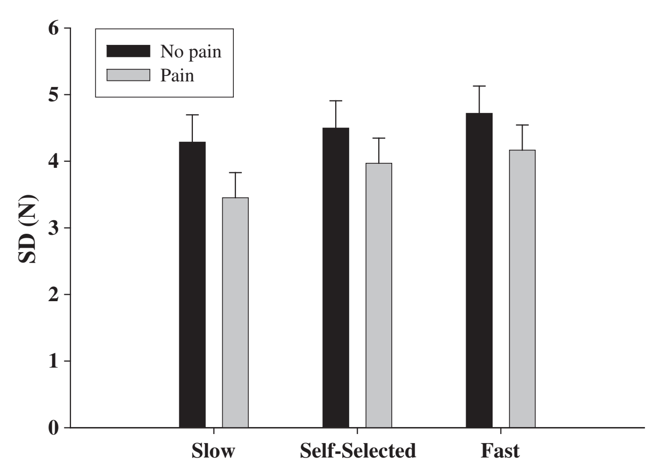

Fig. 3. Standard deviation (SD) of peak shoulder resultant force as a function of group and propulsion speed.

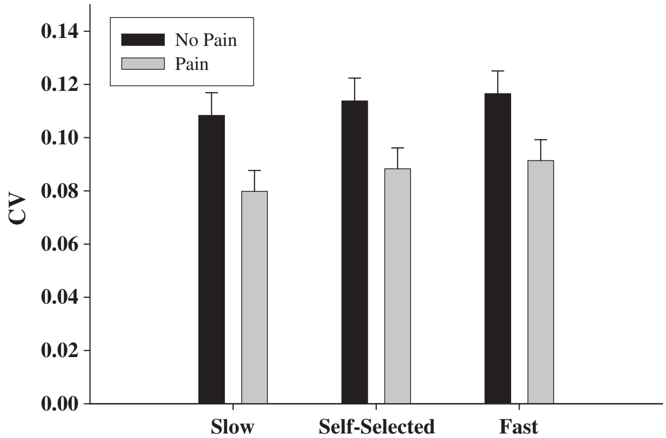

Fig. 4. Coefficient of variation (CV) of peak shoulder resultant force as a function of group and propulsion speed.

targeted for interventions. For instance, it is well known that wheelchair propulsion can be altered with various training interventions (De Groot et al., 2008; Rice et al., 2013). It is within the realm of possibility that wheelchair users could be trained to propel in a more variable movement pattern (Newell et al., 2000). In contrast, it is also potentially possible to mechanically alter the variability of the force applied to the hand rim. For example, it is likely that flexible hand rims which alter propulsion mechanics (Richter et al., 2006) could also result in an increase in the variation in the forces applied to the rim.

The average WUSPI scores of the pain group reported here (22.8 (SD: 21.3)) were lower than previous reports (Brose et al., 2008; Curtis et al., 1999; Finley and Rodgers, 2007; Kemp et al., 2011). This suggests the subjective experience of pain was significantly lower in the current sample. However, it is promising that differences in variability were observed even with this relatively low level of pain. Further work is necessary to determine if variability of propulsion is sensitive to greater amounts of shoulder pain.

Despite the novel observations of this current investigation, it is not without limitations. A major limitation of the study design is its cross-sectional nature. Consequently, this investigation cannot address whether decreased variability results from or contributes to the development of shoulder pain in manual wheelchair users. Further work may investigate the relationship between pain and variability of shoulder forces with a longitudinal design or an experimental model of induced shoulder pain. Another limitation was that the study did not include a thorough examination of upper limb function which might influence variability of movement. To our knowledge there was no participant with upper limb paralysis or impairment. Additionally, the generalizability of the current findings to the general wheelchair population is suspect — given that the current sample was primarily physically active college-age students. It is also important to note that the majority of our samples were child onset wheelchair users — which may impact the pathogenesis of musculoskeletal pain (Sawatzky et al., 2005). Also the current investigation focused on the amount of variability, this approach provides no information concerning the time sequential structure of the movement fluctuations (Hausdorff, 2009). Further research should investigate the time sequential structure of variability in wheelchair propulsion — since the structure of variability is often more sensitive to impairment that distributional measures (Hausdorff, 2009). Lastly, while the current investigation examined resultant force applied to the shoulder, it is possible that the examination of uni-dimensional forces and moments could yield complimentary observations.

5. Conclusion

To our knowledge, this is one of the first investigations to identify an association between variability of forces acting on the shoulder during wheelchair propulsion and symptomatic shoulder pain in manual wheelchair users. This observation supports the proposal that variability is a unique marker of movement that is related to musculoskeletal pain. Further, it raises the possibility that variability of wheelchair propulsion could be used as a clinical measure of shoulder pathology and manipulations of propulsion variability could potentially be a novel prevention strategy for upper limb pain in manual wheelchair users.

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References

Brose, S.W., Boninger, M.L., Fullerton, B., Mccann, T., Collinger, J.L., Impink, B.G., et al., 2008. Shoulder ultrasound abnormalities, physical examination findings, and pain in manual wheelchair users with spinal cord injury. Arch. Phys. Med. Rehabil. 89 (11), 2086–2093.

Collinger, J.L., Boninger, M.L., Koontz, A.M., Price, R., Sisto, S.A., Tolerico, M.L., et al., 2008. Shoulder biomechanics during the push phase of wheelchair propulsion: a multisite study of persons with paraplegia. Arch. Phys. Med. Rehabil. 89 (4), 667–676.

Cooper, R.A., Boninger, M.L., Shimada, S.D., Lawrence, B.M., 1999. Glenohumeral joint kinematics and kinetics for three coordinate system representations during wheelchair propulsion. Am. J. Phys. Med. Rehabil. 78 (5), 435–446.

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

Curtis, K.A., Drysdale, G.A., Lanza, R.D., Kolber, M., Vitolo, R.S., West, R., 1999. Shoulder pain in wheelchair users with tetraplegia and paraplegia. Arch. Phys. Med. Rehabil. 80 (4), 453–457.

De Groot, S., De Bruin, M., Noomen, S.P., Van, D.W., 2008. Mechanical efficiency and propulsion technique after 7 weeks of low-intensity wheelchair training. Clin. Biomech. 23 (4), 434–441.

Desroches, G., Aissaoui, R., Bourbonnais, D., 2008. Relationship between resultant force at the pushrim and the net shoulder joint moments during manual wheelchair propulsion in elderly persons. Arch. Phys. Med. Rehabil. 89 (6), 1155–1161.

Digiovine, C.P., Cooper, R.A., Boninger, M.A., 2001. Dynamic calibration of a wheelchair dynamometer. J. Rehabil. Res. Dev. 38 (1), 41–55.

Finley, M.A., Rodgers, M.M., 2007. Effect of 2-speed geared manual wheelchair propulsion on shoulder pain and function. Arch. Phys. Med. Rehabil. 88 (12), 1622–1627.

Finley, M.A., Rasch, E.K., Keyser, R.E., Rodgers, M.M., 2004. The biomechanics of wheelchair propulsion in individuals with and without upper-limb impairment. J. Rehabil. Res. Dev. 41 (3 B), 385–394.

Gellman, H., Chandler, D.R., Petrasek, J., SIE, I., Adkins, R., Waters, R.L., 1988. Carpal tunnel syndrome in paraplegic patients. J. Bone Joint Surg. Ser. A 70 (4), 517–519.

Gil-Agudo, A., Del Ama-Espinosa, A., Pérez-Rizo, E., Pérez-Nombela, S., Pablo Rodríguez-Rodríguez, L., 2010. Upper limb joint kinetics during manual wheelchair propulsion in patients with different levels of spinal cord injury. J. Biomech. 43 (13), 2508–2515.

Gutierrez, D.D., Thompson, L., Kemp, B., Mulroy, S.J., Winstein, C.J., Gordon, J., et al., 2007. The relationship of shoulder pain intensity to quality of life, physical activity, and community participation in persons with paraplegia. J. Spinal Cord Med. 30 (3), 251–255.

Hamill, J., Van Emmerik, R.E.A., Heiderscheit, B.C., LI, L., 1999. A dynamical systems approach to lower extremity running injuries. Clin. Biomech. 14 (5), 297–308.

Hanavan Jr., E.P., 1964. A mathematical model of the human body. AMRL-TR-64-102. AMRL-TR.Aerospace Medical Research Laboratories (6570th), pp. 1–149.

Hardy, S.E., Kang, Y., Studenski, S.A., Degenholtz, H.B., 2011. Ability to walk 1/4 mile predicts subsequent disability, mortality, and health care costs. J. Gen. Intern. Med. 26 (2), 130–135.

Hausdorff, J.M., 2009. Gait dynamics in Parkinson's disease: common and distinct behavior among stride length, gait variability, and fractal-like scaling. Chaos 19 (2), 026113.

Heiderscheit, B.C., Hamill, J., Van Emmerik, R.E.A., 2002. Variability of stride characteristics and joint coordination among individuals with unilateral patellofemoral pain. J. Appl. Biomech. 18 (2), 110–121.

James, C.R., 1996. Effects of overuse injury proneness and task difficulty on joint kinetic variability during landing. In: Stergiou, N. (Ed.), Innovative Analyses of Human Movement. Human Kinetics, Champaign, IL, pp. 29–62.

Kemp, B.J., Bateham, A.L., Mulroy, S.J., Thompson, L., Adkins, R.H., Kahan, J.S., 2011. Effects of reduction in shoulder pain on quality of life and community activities among people living long-term with SCI paraplegia: a randomized control trial. J. Spinal Cord Med. 34 (3), 278–284.

Koontz, A.M., Cooper, R.A., Boninger, M.L., Souza, A.L., Fay, B.T., 2002. Shoulder kinematics and kinetics during two speeds of wheelchair propulsion. J. Rehabil. Res. Dev. 39 (6), 635–649.

Koontz, A.M., Worobey, L.A., Rice, I.M., Collinger, J.L., Boninger, M.L., 2012. Comparison between overground and dynamometer manual wheelchair propulsion. J. Appl. Biomech. 28 (4), 412–419.

Kulig, K., RAO, S.S., Mulroy, S.J., Newsam, C.J., Gronley, J.K., Bontrager, E.L., et al., 1998. Shoulder joint kinetics during the push phase of wheelchair propulsion. Clin. Orthop. Relat. Res. 354, 132–143.

Kulig, K., Newsam, C.J., Mulroy, S.J., Rao, S., Gronley, J.K., Bontrager, E.L., et al., 2001. The effect of level of spinal cord injury on shoulder joint kinetics during manual wheelchair propulsion. Clin. Biomech. 16 (9), 744–751.

Kwarciak, A.M., Sisto, S.A., Yarossi, M., Price, R., Komaroff, E., Boninger, M.L., 2009. Redefining the manual wheelchair stroke cycle: identification and impact of nonpropulsive pushrim contact. Arch. Phys. Med. Rehabil. 90 (1), 20–26.

Laplante, M.P., Kaye, H.S., 2010. Demographics and trends in wheeled mobility equipment use and accessibility in the community. Assist. Technol. 22 (1), 3–17.

Madeleine, P., Madsen, T.M.T., 2009. Changes in the amount and structure of motor variability during a deboning process are associated with work experience and neck–shoulder discomfort. Appl. Ergon. 40 (5), 887–894.

Madeleine, P., Mathiassen, S.E., Arendt-Nielsen, L., 2008. Changes in the degree of motor variability associated with experimental and chronic neck–shoulder pain during a standardised repetitive arm movement. Exp. Brain Res. 185 (4), 689–698.

Mathiassen, S.E., Möller, T., Forsman, M., 2003. Variability in mechanical exposure within and between individuals performing a highly constrained industrial work task. Ergonomics 46 (8), 800–824.

Mercer, J.L., Boninger, M., Koontz, A., Ren, D., Dyson-Hudson, T., Cooper, R., 2006. Shoulder joint kinetics and pathology in manual wheelchair users. Clin. Biomech. 21 (8), 781–789.

Newell, K.M., Deutsch, K.M., Morrison, S., 2000. On learning to move randomly. J. Mot. Behav. 32 (3), 314–321.

Rice, I.M., Hsu, I.M.K., Jayaraman, C., Culp, S., Hsiao-Wecksler, E.T., Sosnoff, J.J., 2012. Handrim force variability during two speeds of wheelchair propulsion. Unpublished poster presentation at the Rehabilitation Engineering and Assistive Technology Society of North America, Jun 30–July 2, 2012, Baltimore, MD, USA.

Rice, I.M., Pohlig, R.T., Gallagher, J.D., Boninger, M.L., 2013. Handrim wheelchair propulsion training effect on overground propulsion using biomechanical real-time visual feedback. Arch. Phys. Med. Rehabil. 94 (2), 256–263.

Richter, W.M., Rodriguez, R., Woods, K.R., Karpinski, A.P., Axelson, P.W., 2006. Reduced finger and wrist flexor activity during propulsion with a new flexible handrim. Arch. Phys. Med. Rehabil. 87 (12), 1643–1647.

Sawatzky, B.J., Slobogean, G.P., Reilly, C.W., Chambers, C.T., Hol, A.T., 2005. Prevalence of shoulder pain in adult- versus childhood-onset wheelchair users: a pilot study. J. Rehabil. Res. Dev. 42 (3 Suppl. 1), 1–8.

Srinivasan, D., Mathiassen, S.E., 2012. Motor variability in occupational health and performance. Clin. Biomech. 27 (10), 979–993.

Van Drongelen, S., Van Der Woude, L.H., Janssen, T.W., Angenot, E.L., Chadwick, E.K., Veeger, D.H., 2005. Mechanical load on the upper extremity during wheelchair activities. Arch. Phys. Med. Rehabil. 86 (6), 1214–1220.

van den Hoorn, W., Bruijn, S.M., Meijer, O.G., Hodges, P.W., Van Dieën, J.H., 2012. Mechanical coupling between transverse plane pelvis and thorax rotations during gait is higher in people with low back pain. J. Biomech. 45 (2), 342–347.