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Predictors of Shoulder Pain in Manual Wheelchair Users

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Predictors of Shoulder Pain in Manual Wheelchair Users

by

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Dedication

This thesis is dedicated to my loving husband, JT Walford.

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Abstract

Predictors of Shoulder Pain in Manual Wheelchair Users

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Manual wheelchair users rely on their upper limbs to provide independent mobility, which leads to high muscular demand on their upper extremities. This increased demand often results in shoulder pain and injury. However, the specific causes of shoulder pain are unknown. Previous work has shown that decreased shoulder muscle strength is predictive of shoulder pain onset, and others have analyzed joint kinetics, joint kinematics, propulsion technique (e.g. cadence, contact percentage) and intra-individual variability for their relation to shoulder pain or injury. However, one challenge to such studies is that the demand placed on the upper extremity cannot be measured directly, and therefore the causal mechanisms leading to pain and injury are unknown. The purpose of this study was to build upon this previous work and determine in a longitudinal setting whether there are specific kinetic, kinematic, spatiotemporal and intra-individual variability measures that predict whether a manual wheelchair user is likely to develop shoulder pain. All participants were asymptomatic for shoulder pain at the time of initial data collection and were categorized into pain and no pain groups based on who developed shoulder pain at either the 18-month or the 36-month follow-up assessment.

Shoulder strength measures, handrim and joint kinetics, kinematics, spatiotemporal measures, individual standard deviations (SDs) and coefficients of variation (CVs) of the aforementioned parameters were evaluated as predictors of shoulder pain using a logistic regression model. The most important predictors of shoulder pain included shoulder adductor strength, positive shoulder joint work during the recovery phase and maximum trunk angle. Individuals who developed shoulder pain had weaker shoulder adductors, higher positive shoulder joint work during recovery, and less trunk flexion than those who did not develop pain. In addition, relative intra-individual variability (CV) was a better predictor of shoulder pain than absolute variability (SD), however future work is needed to determine when increased versus decreased intra-individual variability is more favorable for preventing shoulder pain. Thus, these predictors may provide insight into how to improve rehabilitation training and outcomes for manual wheelchair users and ultimately decrease their likelihood of developing shoulder pain and injuries.

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Chapter 1: Introduction

There are currently 3.7 million wheelchair users in the United States (Brault, 2012), with 90% of these using manual wheelchairs (Kaye et al., 2000). Manual wheelchair users (MWCU) rely on their upper limbs to provide independent mobility, which leads to high muscular demand on their upper extremity. This increased demand often results in shoulder pain and injury (Waring and Maynard, 1991), which can lead to decreased physical activity and quality of life (Gutierrez et al., 2007). However, the specific causes of shoulder pain are unknown. Some studies suggest that demographic variables such as body weight, age and time from injury can predict the presence of shoulder pain, although these variables and their relation to pain are not consistent across studies (Boninger et al., 2001; Ferrero et al., 2015; Mulroy et al., 2015; Nichols et al., 1979; Pentland and Twomey, 1994; van Drongelen et al., 2006; Vogel et al., 2002). In addition, two studies have analyzed the role of shoulder muscle strength and found that decreased strength was a predictor of shoulder pain onset (Mulroy et al., 2015; van Drongelen et al., 2006). Others have looked at various biomechanical measures such as joint kinetics, joint kinematics and propulsion technique (e.g., contact angle, cadence) and their relation to shoulder pain or injury (Eriks-Hoogland et al., 2014; Mercer et al., 2006; Mulroy et al., 2006, 2015). However, one challenge to such studies is that the demand placed on the upper extremity cannot be measured directly, and therefore causal mechanisms cannot be identified.

Studies have used joint kinetics as a surrogate measure for muscle demand (Desroches et al., 2010; Kulig et al., 2001, 1998; Price et al., 2007; Sabick et al., 2004) and found that higher shoulder joint forces and moments are associated with upper limb pain (Mulroy et al., 2006) and pathology (Mercer et al., 2006) in MWCU. However, only one of those studies was longitudinal and therefore able to suggest a link between a higher vertical component of the shoulder joint force and the development of shoulder pain (Mulroy et al., 2006). In addition, higher joint power is hypothesized to contribute to upper limb pain and injury (Guo et al., 2003; Price et al., 2007). Although joint power has not yet been investigated for its relationship to shoulder pain or pathology, joint power captures both the moment occurring at the joint as well as its angular velocity, with lower angular joint velocity and acceleration being considered more favorable for preventing the development of injuries (Shimada et al., 1998). Joint work is another measure used to quantify muscle demand that is often analyzed in gait to provide insight into mechanical efficiency as well as to identify compensatory mobility strategies in various patient populations (DeVita and Hortobagyi, 2000; Olney et al., 1991; Ventura et al., 2011). To our knowledge, the only study that has analyzed joint work in wheelchair propulsion assessed the difference between two methods for calculating upper extremity work and found that the upper extremity supplies more power than is necessary for wheelchair propulsion (Guo et al., 2003), which may be a contributing factor to the development of shoulder pain. Thus, joint work may be useful in assessing upper extremity effort as it incorporates both the shoulder kinetics (joint moment) and kinematics (angular velocity) to determine joint power, which is integrated over time, and thus quantifies sustained upper limb effort.

The recovery hand patterns used by MWCU influences biomechanical variables such as cadence, contact angle and contact percentage, which may be risk factors for developing shoulder pain (Figure 1). The literature suggests that using lower cadence (Mulroy et al., 2006; Rankin et al., 2012), increased contact angle (Mulroy et al., 2006; Paralyzed Veterans of America Consortium for Spinal Cord Medicine, 2005) and increased contact percentage (Paralyzed Veterans of America Consortium for Spinal

Cord Medicine, 2005) can lead to decreased demand on the upper extremity, and therefore decreased likelihood of developing pain and injuries. A number of studies have shown that semi-circular (SC) and double-loop (DL) patterns utilize decreased cadence and increased contact angle (Boninger et al., 2002; Kwarciak et al., 2012; Qi et al., 2014). SC has also been shown to have increased contact percentage compared to the other hand patterns (Boninger et al., 2002; Shimada et al., 1998). In addition, the SC and DL patterns have recently been shown to require less muscle power and stress than arcing (ARC) and single loop (SL) patterns (Slowik et al., 2016b). Clinical practice guidelines recommend that health-care professionals encourage their patients to use a more under-rim pattern, specifically the SC pattern (Paralyzed Veterans of America Consortium for Spinal Cord Medicine, 2005). However, the propulsion pattern used has not yet been directly related to the development of shoulder pain and injury.

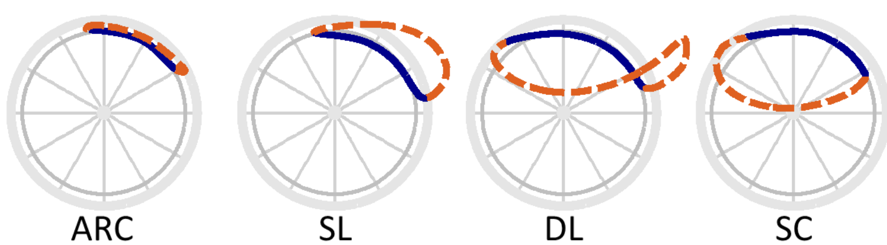

Figure 1: The four primary hand patterns used during manual wheelchair propulsion: arcing (ARC), single loop (SL), double loop (DL), and semi-circular (SC) (Slowik et al., 2016b).

In highly repetitive tasks such as wheelchair propulsion, analyzing intra-individual variability might distinguish those who are at a higher risk of developing pain or injuries. Recent studies have shown that MWCU with shoulder pain have significantly lower cycle-to-cycle variability in peak total shoulder joint force (Moon et al., 2013), peak total handrim force and push time (Rice et al., 2014). Other work has found that kinematic spatial variability in the wrist motion is higher at the beginning of the recovery phase in those with shoulder pain than those without pain (Jayaraman et al., 2014). These differences in variability between pain and no pain groups are consistent with studies of non-MWCU populations where lower intra-individual variability was present in those who had upper limb pain performing a repetitive task (Madeleine et al., 2008a, 2008b). Similarly, able-bodied participants have been shown to have increased mechanical efficiency in manual wheelchair propulsion when propulsion was taught in a more variable manner, specifically in a wheelchair basketball game (Leving et al., 2016), which is consistent with the idea that increased task variability may improve motor learning (Srinivasan and Mathiassen, 2012). Although recent work has shown differences in intra-individual variability between groups of MWCU with and without shoulder pain, it is not yet known if variability is a reaction to the presence of pain or part of the pathomechanics causing shoulder pathology and pain and potentially as a predictor of those who will develop shoulder pain over time.

The purpose of this study was to determine whether there are specific biomechanical measures that predict whether a manual wheelchair user is likely to develop shoulder pain over time. Specifically, we hypothesize that individuals with higher shoulder joint work, higher shoulder joint moments, higher handrim forces, decreased cycle-to-cycle variability and those using a more over-rim hand pattern will be more likely to develop shoulder pain. We also expect that the effect of shoulder strength on the shoulder joint kinetics will better predict shoulder pain development than joint kinetics alone. If these modifiable risk factors can be identified, we can further investigate them as potential targets for interventions to prevent the development of shoulder pain and injury.

Chapter 2: Methods

SUBJECTS

Experimental data were collected and analyzed from 102 individuals with paraplegia (93 men, 9 women; age: 36.2 ± 9.6 years; time from injury: 9.5 ± 6.5 years; height: 1.74 ± 0.09 m; mass: 73.9 ± 15.9 kg). Participants were recruited from outpatient clinics from the Rancho Los Amigos National Rehabilitation Center and gave informed written consent in accordance with the Institutional Review Board. All participants were asymptomatic for shoulder pain at the time of the initial baseline data collection and were categorized into either the pain (P) or no pain (NP) group based on whether they experienced an increase of ≥ 10 points on the Wheelchair User's Shoulder Pain Index (Curtis et al., 1995) from baseline at either the 18-month or 36-month follow-up assessment.

DATA COLLECTION

Participants propelled their own wheelchair on a stationary ergometer (Figure 2) at their comfortable, self-selected speed (avg: 1.00 ± 0.26 m/s) for a 40-second trial with data collected during the last 10 seconds. Three-dimensional kinetic data were measured from the handrim on the right side using an instrumented wheel (SmartWheel; Three Rivers Holdings, Mesa, AZ, USA) at 200 Hz. Kinematic data were collected from the trunk, right-side upper extremity and wheel using a CODA motion analysis system (Charnwood Dynamics Ltd., Leicestershire, UK) at 100 Hz with 15 active markers placed on body segment landmarks and right wheel (e.g. Lighthall-Haubert et al., 2009). Shoulder strength was measured as peak maximal isometric torque of the shoulder flexors, extensors, abductors, adductors, internal rotators and external rotators using a Biodex System 3 Pro dynamometer (Biodex Medical Systems Inc., Shirley, NY, USA).

Strength, kinetic and kinematic data were collected from the participants at three time points: baseline, 18 months after baseline, and 36 months after baseline.

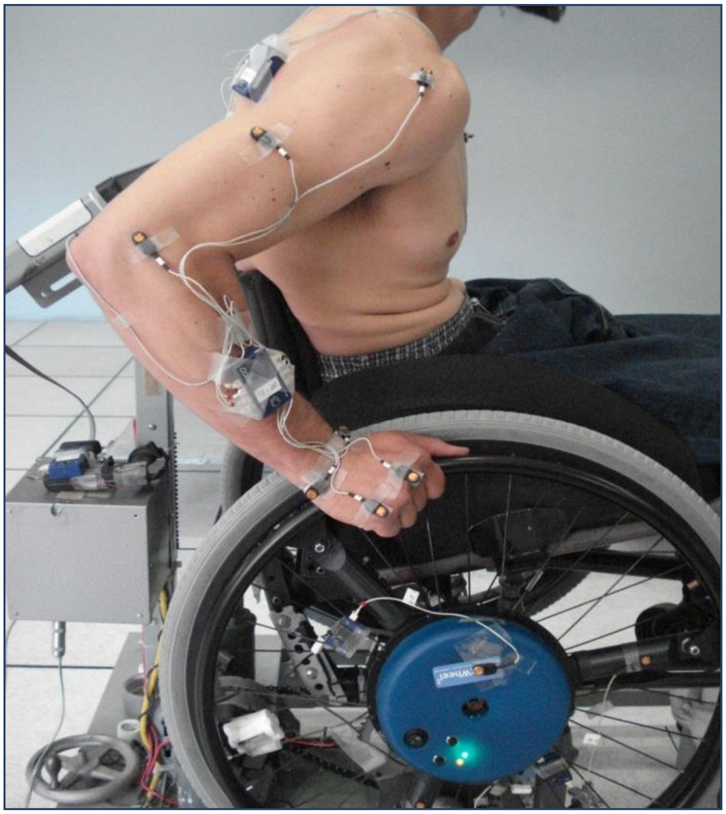

Figure 2: Experimental setup used to collect kinematics and kinetics on a custom-built wheelchair ergometer.

DATA PROCESSING

Kinematic and kinetic data were processed using a low-pass, fourth-order, zero-lag Butterworth filter with cutoff frequencies of 6 and 10 Hz, respectively, in Visual 3D (C-Motion, Inc., Germantown, MD, USA). A threshold of 5 N for the resultant handrim force was used to indicate the beginning and end of the contact and recovery phases. Trunk angle, shoulder plane-of-elevation, shoulder elevation angle and shoulder internal/external rotation angles were calculated in Visual 3D using the International Society of Biomechanics recommendations (Wu et al., 2005). Ranges of motion (ROM), average, maximum and minimum for these angles, peak and average handrim forces, cadence, contact time, contact angle and contact percentage were calculated for each cycle and averaged across cycles for each subject (Table 1). The third metacarpophalangeal joint center (MCP3) was located using a method described previously (Rao et al., 1996), and the path of the MCP3 was projected onto the handrim plane and averaged across cycles. This closed-curve hand path was then used to calculate two quantitative parameters to characterize the hand pattern: net radial thickness (NRT) and total radial thickness (TRT), which quantify the hand's displacement above the handrim and the absolute distance between the hand and handrim, respectively, using a previously described method (Slowik et al., 2015). Peak and average shoulder joint moments (i.e., flexion/extension, adduction/abduction and internal/external rotation) were calculated for the contact and recovery phases using inverse dynamics in Visual 3D for each cycle, and averaged across cycles for each subject. Shoulder joint power was calculated in Visual 3D and exported to custom code in Matlab (Mathworks Inc., Natick, MA, USA) in order to calculate shoulder joint work (i.e., the time integral of shoulder joint power, see Figures 5-6 in Appendix D). Positive and negative shoulder joint work were calculated for the contact and recovery phases for each cycle and averaged across cycles for each subject. Cycle-to-cycle variability was measured for each subject by calculating the absolute variability as the standard deviation (SD) and calculating the relative variability as the coefficient of variation (CV=SD/|mean|) across all cycles for each of the aforementioned parameters (Table 1). All dependent measures were calculated for the right side only because all subjects were asymptomatic for shoulder pain at the time of data collection, and were assumed to have bilateral symmetry (Soltau et al., 2015).

Variable Name	Abbreviation	Definition
Kinematics: ROM, max and min:		
Trunk Angle [º]		Angle of trunk position relative to the lab vertical axis
Plane-of-elevation [º]		Humerus angle relative to the trunk about the vertical axis
Elevation Angle [º]		Humerus angle about its forward axis
Internal/External Rotation [º]		Humerus angle about its longitudinal axis
Handrim kinetics: Peak and average:		
Tangential Force [N]	F_{tan}	Tangential force applied to the handrim
Radial Force [N]	F_{rad}	Radial force applied to the handrim
Lateral Force [N]	F_{lat}	Lateral force applied to the handrim
Total Force [N]	F_{tot}	$F_{tot} = \sqrt{F_{tan}^2 + F_{rad}^2 + F_{lat}^2}$
Spatiotemporal measures:		
Contact Time [s]		Amount of time spent in the contact phase (based on handrim force threshold of 5N)
Cycle Time [s]		Amount of time spent in the full propulsion cycle
Contact Percentage [%]		Percentage of the propulsion cycle spent in the contact phase
Contact Angle [º]	θ	Angle between the positions of the hand between the start and end of the contact phase
Net Radial Thickness [m]	*NRT*	Displacement of the hand above the handrim
Total Radial Thickness [m]	*TRT*	Distance between the hand and handrim

Table 1: Definition of parameters analyzed.

Joint kinetics: for contact, recovery and full cycle		
Peak and Average Flexion/Extension Moment [mm]	M_x	Peak and average reaction moment at the shoulder joint about the mediolateral axis, normalized by body weight
Peak and Average Ad/Abduction Moment [mm]	M_y	Peak and average reaction moment at the shoulder joint about the forward axis, normalized by body weight
Peak and Average Int/External Rotation [mm]	M_z	Peak and average reaction moment at the shoulder joint about the humerus longitudinal axis, normalized by body weight
Positive Shoulder Joint Work [mm]	W_{pos}	Integral of the shoulder joint power (dot product of the shoulder joint moment and angular velocity) for Power > 0
Negative Shoulder Joint Work [mm]	W_{neg}	Integral of the shoulder joint power (dot product of the shoulder joint moment and angular velocity) for Power < 0

Table 1, cont.

STATISTICAL ANALYSES

To determine which parameters were predictors of shoulder pain development, variables were organized into nine groups and a stepwise logistic regression procedure was performed in Matlab. Sub-model groups included 1) Strength measures, 2) Handrim kinetics, 3) Joint kinetics, 4) Handrim and joint kinetics, 5) Handrim kinetics, joint kinetics and strength, 6) Kinematics, 7) Spatiotemporal variables, 8) SDs, and 9) CVs as potential predictors. Strength measures included in sub-model 1 included shoulder flexors, adductors and external rotators in order to represent all functional muscle groups without introducing multicollinearity (e.g., Mulroy et al., 2015). In each sub-model, variables that were highly correlated (e.g., variance inflation factor, VIF ≥ 5) to another variable were removed as potential predictors from the model. Forward stepping was conducted with a p-value of < 0.1 to enter and ≥ 0.1 to remove based on the F-test of the change in deviance as the criterion for adding or removing a variable. Once all regression models were complete, predictors from all models with a p-value < 0.1 were added to a final model. Predictors with VIF ≥ 5 were removed, and then a logistic regression was run on the final predictors using k-fold cross-validation with k=5 subject groups. The five subject groups were chosen using a random number generator, and for each of the five iterations of the cross-validation procedure, the four groups that were used to train the model will be referred to as "training groups" while the group that was left out of the regression model will be referred to as the "testing group".

Chapter 3: Results

Sub-model results yielded a total of sixteen potential predictors to be added to the final model (Appendix B, Tables 6-13). Only sub-model 7 (spatiotemporal variables) did not produce any predictors. Values of all measures are provided in Appendix A. Of the final predictors, one term (average internal/external rotation moment during recovery) was correlated to other joint kinetic terms and was therefore removed as a potential predictor. Remaining predictors had VIFs < 3, indicating little to no multicollinearity (Table 2). The 5-fold cross-validation yielded overall model results that were statistically significant (all $p<0.005$) and on average, explained 47.6 ± 3.3% (mean ± SD) of the variance in the data (Tjur R^2; Tjur, 2009). The average Tjur R^2 value for the testing groups was 27.3 ± 8.1% (mean ± SD). Final model coefficients and p-values are provided in Appendix C.

Variable Name
Strength measures:
Adductor Torque [N-m]
Handrim Kinetics:
Average Tangential Force [N] (+ forward at top dead center, TDC / - backward at TDC) **Average Radial Force [N]** (+ outward / - inward)
Joint Kinetics:
Average Flexion/Extension Moment during Contact [mm] (+ flexion / - extension) **Average Adduction/Abduction Moment during Recovery [mm]** (+ adduction / - abduction) **Negative Shoulder Joint Work during Contact [mm]** **Positive Shoulder Joint Work during Recovery [mm]**
Kinematics:
Maximum Trunk Angle [º] (+ backward / - forward)
SDs
Minimum Trunk Angle SD [º] (+ extension / - flexion)
CVs
Average Tangential Force CV [%] **Average Radial Force CV [%]** **Contact Angle CV [%]** **Average Adduction/Abduction Moment during Contact CV [%]**
Interactions
Adduction Torque [N-m] * Average Flexion/Extension Moment during Contact [mm]
Average Tangential Force [N] * Average Flexion/Extension Moment during Contact [mm]
Average Flexion/Extension Moment during Contact [mm] * Negative Shoulder Joint Work during Contact [mm]

Table 2: Variables included as predictors in the final model. CV variables are represented as a percentage (SD/|mean|*100). Shoulder joint work and moments were normalized by body weight (N-mm/N).

Maximum trunk angle and adductor torque were significant predictors of shoulder pain in all of the 5-fold cross-validation models (all $p<0.03$, Appendix C, Tables 14-18). Positive shoulder joint work during recovery and the adductor torque and average flexion/extension moment during contact interaction were significant predictors in four of the five models ($p<0.05$ in Models 2-5, and $p<0.02$ in Models 1, 2, 4 and 5, respectively). CV of the average adduction/abduction moment during contact and CV of the contact angle were significant predictors of shoulder pain in two models ($p<0.04$ in Models 1 and 5, and $p<0.03$ in Models 4 and 5, respectively). Finally, the variables that were significant predictors in only one of models included the negative shoulder joint work during contact ($p=0.038$, Model 2), the SD of the minimum trunk angle ($p=0.038$, Model 5), CV of the average radial force ($p=0.046$, Model 4), the interaction of average tangential force and average flexion/extension moment during contact ($p=0.035$, Model 4) and the interaction of the average flexion/extension moment during contact and negative shoulder joint work during contact ($p=0.040$, Model 2). None of the spatiotemporal measures were predictors of shoulder pain development.

Regarding these predictors, individuals who developed shoulder pain had more trunk extension, less strength in the shoulder adductors, more positive shoulder joint work during recovery and less negative shoulder joint work during contact. Relative variability was larger for the contact angle, larger for the average adduction/abduction moment during contact and smaller in the average radial force in the pain group while absolute variability was smaller for the minimum trunk angle in the pain group (Table 3). The first interaction term indicates that as the strength of the shoulder adductors decreases, the influence of the average flexion/extension moment during contact on pain development decreases, which is lower in the individuals who develop pain. The second interaction term indicates that as the average flexion/extension moment during contact decreases, the negative shoulder joint work during contact has more of an influence on pain development, which was smaller in the pain group. The final interaction indicates that as the average tangential force decreases, the influence of the flexion/extension moment during contact on pain development increases, which was lower in those who develop shoulder pain.

Predictor	Occurrences	No Pain Group Average (n=74)	Pain Group Average (n=28)	% Difference
Individual Predictors				
Maximum Trunk Angle [°]	5	-7.81 ± 7.32	-3.55 ± 8.91	-74.89
Adductor Torque [N-m]	5	71.50 ± 19.54	63.90 ± 20.57	11.23
Positive Shoulder Joint Work during Recovery [mm]	4	3.96 ± 1.19	4.11 ± 1.03	3.72
CV Contact Angle [%]	2	4.49 ± 2.44	5.31 ± 2.43	
CV Average Add/Abd Moment during Contact [%]	2	131.7 ± 394.9	213.2 ± 544.4	
CV Average F_{rad} [%]	1	16.83 ± 12.76	11.27 ± 5.77	
SD Minimum Trunk Angle [°]	1	0.841 ± 0.476	0.621 ± 0.336	30.17
Negative Shoulder Joint Work during Contact [mm]	1	-0.486 ± 0.466	-0.338 ± 0.263	-35.85

Table 3: Variables that were significant ($p<0.05$) predictors in at least one model during cross-validation (n=102). Group averages are mean ± SD. ‘Occurrences’ indicates how many of the 5-fold cross-validation models included that predictor. Percent difference is given for all measures excluding relative variability (CVs).

Interaction Predictors				
Adductor Torque [N-m] **:** Average Flex/Ext Moment during Contact [mm]	4			
Average Flex/Ext Moment during Contact [mm] **:** Negative Shoulder Joint Work during Contact [mm]	1			
Average F_{tan} [N] **:** Average Flex/Ext Moment during Contact [mm]	1			

Table 3, cont.

Chapter 4: Discussion

Manual wheelchair users commonly experience shoulder pain and injury, which is often related to rotator cuff impingement or tears (Bayley et al., 1987; Waring and Maynard, 1991). However, the specific causes of shoulder pain development are unknown because most studies examining biomechanical measures between pain and no pain groups are often not longitudinal, but rather examine wheelchair users who presently do and do not have shoulder pain (Collinger et al., 2008; Dysterheft et al., 2017; Jayaraman et al., 2016, 2015, 2014; Moon et al., 2013; Rice et al., 2014). The present study is unique in that at the initial assessment, none of the participants experienced shoulder pain, and were later grouped into P and NP groups based on who developed shoulder pain over the next three years. Categorizing participants based on future pain development allowed for the use of a regression model to identify biomechanical measures of strength, kinetics, kinematics and variability that are important predictors of shoulder pain development in MWCU.

STRENGTH

Strength of the shoulder adductors was a predictor of shoulder pain in all five cross-validation models (Appendix C, Tables 14-18, Models 1-5), with individuals who develop pain having an 11.2% difference in shoulder adductor strength than those who do not. The adductor strength values (Table 3, NP=1.00±0.29 N-m/kg, P=0.90±0.37 N-m/kg) in this study were similar to those found in previous work (Gagnon et al., 2016; Sabick et al., 2004). Our finding that shoulder adductor weakness is predictive of pain is consistent with previous work that found shoulder strength plays a role in shoulder pain development (Mulroy et al., 2015; van Drongelen et al., 2006). The shoulder adductors provide adequate depression of the humeral head during weight-bearing tasks such as during transfers (Perry et al., 1996). Therefore, if adductor weakness is present, it may not allow for the required unweighting of the rotator cuff which in turn could cause impingement (Burnham et al., 1993; Mulroy et al., 2015). In addition, others have suggested strengthening the shoulder adductors as well as the abductors and external rotators to avoid shoulder pain and injuries (Curtis et al., 1999; Mulroy et al., 2004; Qi et al., 2012; Slowik et al., 2016a). Therefore, therapies that target shoulder strength, especially in the shoulder adductors, may help decrease the likelihood of shoulder pain development (Curtis et al., 1999; Mulroy et al., 1996), especially when included in rehabilitation programs that address additional factors that contribute to shoulder pain in MWCU.

KINETICS

The joint kinetic measures included in the cross-validation regression models were negative shoulder joint work during contact (Appendix C, Table 15, Model 2) and positive shoulder joint work during recovery (Appendix C, Tables 15-18, Models 2-5). The net work done over the full propulsion cycle (NP=6.73±2.43 N-m, P=6.45±2.40 N-m) was less than that reported in previous work (Guo et al., 2003) because the present study only calculated shoulder joint work rather than the work done by all upper limb segments.

Contrary to our hypothesis that shoulder joint work would be higher in those who develop shoulder pain, the P group performed less negative work during the contact phase than the NP group. During contact, the shoulder extensors start to become active with the shoulder flexors in order to decelerate the arm (Rankin et al., 2011). At the end of contact, shoulder joint work is negative (Figure 3) when the arm is decelerating and changing direction (Mulroy et al., 1996). Even though negative shoulder joint work, which indicates eccentric muscle contraction, is higher in the NP group during contact, the P group had higher negative shoulder joint work over the full propulsion cycle. Therefore, lower negative shoulder joint work over the full cycle in the NP group may be indicative of more efficient use of the shoulder muscles by beginning eccentric muscle action earlier (in contact) than the P group without producing more negative shoulder joint work in the full propulsion cycle. However, future work is needed to determine the primary contributors to this difference in negative shoulder joint work. It is also important to note that negative shoulder joint work during contact was only significant in one of the five final models (Appendix C, Table 15, Model 2) and has a small magnitude (Table 3). Therefore, whether or not negative shoulder joint work during contact is a clinically meaningful measure should be investigated in future studies.

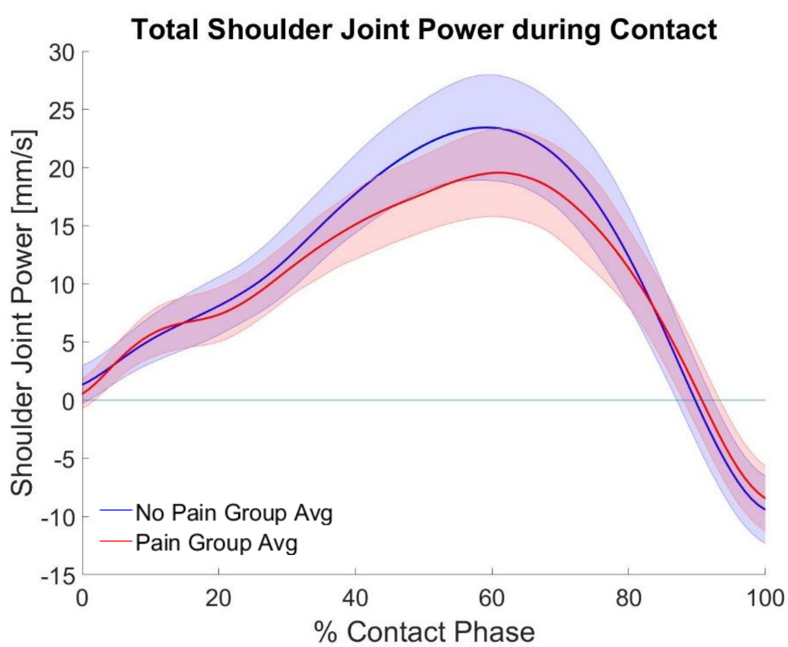

Figure 3: Total shoulder joint power during the contact phase with one standard deviation from the mean shaded. Joint power is normalized by body weight (N-mm/N-s).

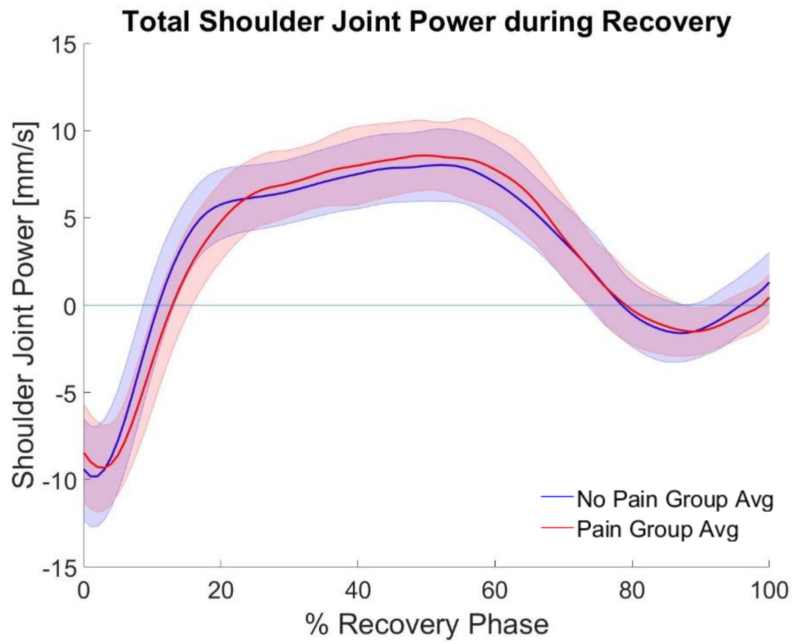

Figure 4: Total shoulder joint power during the recovery phase with one standard deviation from the mean shaded. Joint power is normalized by body weight (N-mm/N-s).

Consistent with our hypothesis that shoulder joint work would be higher in individuals who develop shoulder pain, the P group had higher positive shoulder joint work during recovery than the NP group. Shoulder joint work is positive in the middle of the recovery phase (Figure 4) when the shoulder delivers power to accelerate the arm backward (Mulroy et al., 1996; Rankin et al., 2011). The P group delivered more power to the shoulder than the NP group in the recovery phase where delivering power is not critically needed (Price et al., 2007; Rankin et al., 2011), which suggests that the NP group conserves power production during the recovery phase in order to conserve upper limb effort that will be needed during contact. However, future work should utilize modeling and simulation to identify the underlying mechanisms that contribute to the differences between positive and negative shoulder joint work in those who do and do not develop shoulder pain that are not identifiable in an experimental setting.

Three interaction terms between kinetic variables were significant in the final models. However, interpretation of the interaction terms becomes difficult to apply in a clinical setting. Two of the interaction terms are only significant predictors in one model (Table 3). The third interaction was between the adductor torque and average flexion/extension shoulder joint moment during contact, which was significant in four of the five final models (Appendix C, Tables 14-15, 17-18, Models 1-2, 4-5). However, the average flexion/extension moment during contact was not a significant predictor of pain on its own or in the final models, and therefore may only be a predictor of pain development for those with lower moments during contact despite stronger shoulder adductor muscles.

KINEMATICS

Maximum trunk angle was a significant predictor of shoulder pain in every cross-validation model (Appendix C, Tables 14-18, Models 1-5). The maximum trunk angle for the NP group was more forward leaning than the P group by 4.26°, yielding a 74.9% difference between groups (Table 3). The population average was similar to previously reported trunk angles (Rao et al., 1996). Maximum trunk angle was not indicative of increased trunk range of motion (Table 4), which may put individuals at higher risk for injury (Rodgers et al., 2000), but rather the NP group had more trunk flexion than the P group for very similar trunk ranges of motion. The literature contains conflicting evidence regarding the magnitude of the trunk contribution to propulsion power (Guo et al., 2006, 2003; Rankin et al., 2011). Therefore, future work should analyze how muscle contributions change with increased trunk flexion without a change in trunk range of motion and determine if these muscle contributions change significantly with only 4-5° of added trunk flexion. Though a more flexed trunk posture may reduce the likelihood of shoulder pain development, the ability to adopt a more forward leaning posture may depend on the level of spinal cord injury. Therefore, trunk angle may not always be a modifiable rehabilitation target for all patients.

		Range of Motion	Average	Maximum	Minimum
Shoulder Rotation [°]	No Pain	67.2 ± 20.6	53.1 ± 10.9	80.6 ± 10.7	13.3 ± 19.1
	Pain	63.8 ± 17.3	60.0 ± 15.2	85.7 ± 13.3	21.9 ± 23.3
Trunk Angle [°]	No Pain	5.70 ± 3.00	-10.9 ± 7.62	-7.81 ± 7.32	-13.5 ± 8.14
	Pain	5.59 ± 2.68	-6.57 ± 9.48	-3.55 ± 8.91	-9.15 ± 10.1

Table 4: ROMs, average, maximum and minimum angles for the trunk and shoulder rotation. Values represent mean ± SD.

On its own, average shoulder rotation angle over the full propulsion cycle is a significant predictor of shoulder pain (p=0.016) with an overall model p=0.013. However, due to multicollinearity, average shoulder rotation angle was removed from sub-model 6 and could not be considered as a potential predictor in the final model. Average shoulder rotation was higher in the P group, indicating a more internally rotated arm posture than the NP group with a 12.2% difference in the average shoulder rotation angle (Table 4). The population average was similar to what has been reported in previous work (Collinger et al., 2008). Shoulder pathology such as impingement is more likely to occur when internal rotation is paired with abduction or forward flexion (Hawkins and Kennedy, 1980; Neer, 1983; Newsam et al., 1999). Furthermore, the shoulder joint experiences peak loading during arm extension and internal rotation, leaving it susceptible to injury (Collinger et al., 2008). Therefore, minimizing the shoulder internal rotation angle during propulsion may also decrease the likelihood of shoulder pain development. Future work is needed to understand how muscle contributions to propulsion change with more trunk flexion and less shoulder internal rotation. Musculoskeletal modeling and simulation analyses could provide insight on these alternative propulsion techniques that may help inform improved rehabilitation programs for MWCU to minimize their chances of developing shoulder pain and injuries.

VARIABILITY

Consistent with our hypothesis that intra-individual variability would be lower in individuals who develop pain, absolute variability (SD) in the minimum trunk angle (i.e. trunk flexion) was lower in the P group and was a significant predictor of pain in one of the five final models (Appendix C, Table 18, Model 5). Therefore, more cycle-to-cycle variability in the trunk flexion angle may help decrease the likelihood of shoulder pain development, which is consistent with previous work analyzing butchers performing a repetitive task that found cycle-to-cycle trunk variability was lower in the presence of shoulder pain even though variability in the arm motion posture increased (Madeleine et al., 2008b). However, another study that analyzed butchers with experimental-induced shoulder pain found that variability in kinematic measures can increase in the presence of shoulder pain (Madeleine et al., 2008a). Therefore, the effect of whether increased or decreased kinematic variability can cause shoulder pain development is not consistent and requires further investigation.

Although cycle-to-cycle variability of both handrim and joint kinetics has been found to be lower in individuals with shoulder pain (Moon et al., 2013; Rice et al., 2014), our results for the radial force CV (handrim kinetic variability) and adduction moment CV (joint kinetic variability) were not consistent with one another. Supporting our hypothesis that individuals who develop pain would have lower cycle-to-cycle variability, relative variability (CV) of the radial handrim force was lower in the P group and was a predictor of shoulder pain in one of the five final models (Appendix C, Table 17, Model 4). Thus, a more variable radial component of the handrim force may help decrease the likelihood of an individual developing shoulder pain, consistent with previous work that found that the peak resultant handrim force CV was smaller in individuals with shoulder pain than those without (Rice et al., 2014). The lower radial handrim force CV may also be related to the lower CV of the peak resultant shoulder joint force in the pain group than the no pain group found in other work (Moon et al., 2013). Because these variables were collected prior to the onset of shoulder pain, our results suggest that lower variability in handrim force may be part of the wheelchair propulsion mechanics that lead to shoulder joint pain. However, inconsistent with our hypothesis that individuals who develop pain would have lower cycle-to-cycle variability, we also found that the average adduction/abduction shoulder joint moment during contact CV was higher in the P group. One study suggests that the difference between when increased or decreased relative variability is more favorable depends on whether the variable is measured during contact versus recovery (Sosnoff et al., 2015). However, in this case, we would expect that the average adduction/abduction moment during contact CV would be lower in the P group as it is for the radial force CV, which is both a kinetic measure and contact phase measure. Variability of shoulder joint moments has not been analyzed for its relationship to shoulder pain in previous work. Thus, the present study shows that kinetic relative variability, while predictive of shoulder pain, should be further analyzed to understand when using increased or decreased cycle-to-cycle relative variability is favorable for avoiding shoulder pain.

The contact angle CV in the present study was similar to that reported in previous work (Rice et al., 2014), but contrary to our hypothesis, the relative variability of the contact angle was higher in the P group. Higher contact angle CV for the P group is consistent with previous work that showed structured variability (i.e., sample entropy, SampEn) of the contact angle is higher in individuals with shoulder pain than those without pain (Jayaraman et al., 2016). Higher intra-individual variability in the contact angle, a spatial measure, may also be related to other work that found higher spatial variability of the wrist motion during the beginning of the recovery phase in MWCU who presently have shoulder pain than those without pain (Jayaraman et al., 2014). However, these two variables quantify variability differently and during two different phases of the propulsion cycle, and therefore future work should analyze the relationship between these two measures. Though contact angle CV was higher in the P group, other work found that the CV of contact time was smaller in individuals with shoulder pain than without (Rice et al., 2014), suggesting that whether increased or decreased variability is favorable may vary between spatial versus temporal measures. Therefore, future work should compare methods for quantifying spatiotemporal variability (i.e., using CV vs. SampEn) as well as study how variability of spatial and temporal measures differ in order to better understand whether increased or decreased cycle-to-cycle variability in spatiotemporal measures is favorable.

Another observation to note is that only one absolute variability (SD) measure was a predictor of shoulder pain, but three relative variability (CV) measures were significant predictors. In addition, the CV sub-model had six more predictors than the SD sub-model did, with nearly a 20% increase in the Tjur R^2 value (Appendix B, Tables 12 and 13, Sub-models 8 and 9). Thus, relative cycle-to-cycle variability is likely a better predictor of shoulder pain than absolute variability. Finally, higher cycle-to-cycle variability may not prevent shoulder pain development, but rather future work is needed to determine how variability during contact versus recovery in kinetic, kinematic and spatiotemporal measures plays a role in the development of pain in MWCU.

LIMITATIONS

A potential limitation of this study was that the outcome measure, pain, was represented as a binary rather than as a continuous variable. Grouping subjects categorically into pain or no pain groups makes it difficult to see how certain biomechanical measures may influence shoulder pain development more heavily than others. However, because studies have yet to determine definitive predictors for shoulder pain development, using two groups for this study was sufficient in order to determine more generally how MWCU might avoid developing any level of shoulder pain. Future work should look into how these predictors might affect the degree of shoulder pain. Although using categorical groups for pain makes the results of the present study more generalizable, this is not the case for all MWCU with paraplegia. Another limitation was that the large majority (>90%) of the population was male. Future work is needed to determine whether the predictors of shoulder pain identified in this study are generalizable to females as well.

The P and NP groups were determined based on pain occurring within a three-year period, which is a relatively short time period. This is a potential limitation as some patients in our NP group may develop shoulder pain in future years. However, important biomechanical predictors of pain identified in this study including shoulder adductor strength, positive shoulder joint work during recovery, maximum trunk angle and relative cycle-to-cycle variability were identified shortly before the onset of pain in the P group. Therefore, future studies may have the ability to determine who is likely to develop shoulder pain in the relatively near future so that precautions can be taken to avoid shoulder pain and injury.

In this study, cycle-to-cycle variability was measured over a period of ten seconds, which also could be a limitation in that trends of variability may not be detectable in such a short time period. Future work should analyze variability over a longer time to determine whether these same relationships are observed or if variability changes when considering a longer continuous period of propulsion or multiple trials.

A few potential limitations regarding the experimental design of this study are that only level propulsion was analyzed, only one instrumented wheel was used for data collection and data were measured using a stationary ergometer rather than overground propulsion. MWCU encounter a variety of turns and bouts of propulsion in executing activities of daily living (Sonenblum et al., 2012). However, understanding first how user preferences during level ground propulsion influence pain development is needed before being able to understand the implications of alternative propulsion conditions. Using one instrumented wheel likely did not alter study results because trials were randomly selected, and therefore minimized systematic differences that might have occurred from fatigue effects. In addition, previous work from our group has shown that with one instrumented wheel, side-to-side differences are low and not clinically significant (Soltau et al., 2015). Finally, though ergometers cannot perfectly replicate overground conditions, trends in kinetic, kinematic and spatiotemporal measures are consistent in MWCU between propulsion using an ergometer and overground propulsion (Koontz et al., 2012), and therefore the conclusions of this study were likely not altered. However, future work should examine how intra-individual variability changes during overground propulsion when compared to an ergometer.

Chapter 5: Conclusions

This study identified predictors of shoulder pain development in manual wheelchair propulsion by analyzing a population of MWCU who were asymptomatic for shoulder pain at baseline and were categorized into P and NP groups based on who developed shoulder pain 18-36 months later. The most influential predictors of shoulder pain included strength of the shoulder adductors, the maximum trunk angle and the positive shoulder joint work during the recovery phase. In addition, relative cycle-to-cycle variability in kinetic and spatiotemporal measures was predictive of shoulder pain, although whether increased or decreased variability was more favorable for avoiding shoulder pain was not consistent. Thus, the predictors identified in this study provide insight for future work to improve the rehabilitation and biomechanical analysis of MWCU, and ultimately decrease their likelihood of developing shoulder pain.

Appendices

APPENDIX A

Predictor		No Pain Group Average (n=74)	Pain Group Average (n=28)	% Difference
Strength measures:				
Flexion Torque [N-m]		61.24 ± 17.40	57.88 ± 18.94	5.65
Adduction Torque [N-m]		71.50 ± 19.54	63.90 ± 20.57	11.23
External Rotation Torque [N-m]		34.03 ± 9.55	32.16 ± 10.95	5.66
Kinematics:				
Trunk Angle [°]	ROM	5.70 ± 3.00	5.59 ±2.68	1.79
	Average	-10.86 ± 7.62	-6.57 ± 9.48	-49.17
	Minimum	-13.50 ± 8.14	-9.15 ± 10.05	-38.45
	Maximum	-7.81 ± 7.32	-3.55 ± 8.91	-74.89
Plane-of-Elevation [°]	ROM	76.27 ± 21.72	70.10 ± 20.67	8.43
	Average	-26.65 ± 11.35	-29.44 ± 10.57	-9.95
	Minimum	-57.13 ± 10.09	-56.95 ± 10.26	-0.316
	Maximum	19.14 ± 20.10	13.15 ± 20.61	37.09
Elevation Angle [°]	ROM	25.07 ± 7.34	28.19 ± 8.32	11.72
	Average	40.18 ± 6.62	41.27 ± 5.13	2.67
	Minimum	27.56 ± 7.04	26.76 ± 4.71	2.95
	Maximum	52.63 ± 7.61	54.95 ± 7.57	4.31
Shoulder Rotation [°]	ROM	67.24 ± 20.58	63.79 ± 17.34	5.26
	Average	53.10 ± 10.90	59.98 ± 15.21	12.18
	Minimum	13.34 ± 19.12	21.87 ± 23.27	48.44
	Maximum	80.58 ± 10.65	85.66 ± 13.33	6.11

Table 5: Group averages for the NP and P groups for all biomechanical measures excluding SDs and CVs. Group averages are mean ± SD.

Predictor			No Pain Group Average (n=74)	Pain Group Average (n=28)	% Difference
Handrim Kinetics:					
Tangential Force [N]		Average	19.48 ± 4.86	18.35 ± 4.55	5.93
		Maximum	32.39 ± 9.58	29.69 ± 8.28	8.69
Radial Force [N]		Average	-16.29 ± 6.33	-18.57 ± 5.88	-13.09
		Minimum	-30.65 ± 10.30	-33.12 ± 11.66	-7.75
Lateral Force [N]		Average	-4.84 ± 3.48	-5.62 ± 3.53	-14.95
		Minimum	-9.61 ± 4.56	-10.09 ± 4.62	-4.93
		Maximum	1.59 ± 2.03	1.31 ± 0.942	19.37
Total Force [N]		Average	28.05 ± 6.52	28.45 ± 7.08	1.41
		Maximum	42.87 ± 12.05	41.91 ± 12.85	2.27
Joint Kinetics:					
Flexion/ Extension Moment [mm]	Avg.	Contact	7.69 ± 3.55	7.13 ± 3.06	7.54
		Recovery	-2.32 ± 1.49	-2.18 ± 1.58	-5.99
		Full Cycle	2.69 ± 1.93	2.48 ± 1.78	8.23
	Peak	Contact	14.30 ± 5.79	13.06 ± 4.89	9.03
		Recovery	3.51 ± 2.44	3.42 ± 2.31	2.59
		Full Cycle	14.30 ± 5.78	13.06 ± 4.89	9.03
Adduction/ Abduction Moment [mm]	Avg.	Contact	1.38 ± 3.50	2.01 ± 3.05	36.76
		Recovery	-6.03 ± 1.02	-6.34 ± 0.967	-5.04
		Full Cycle	-2.32 ± 1.91	-2.17 ± 1.75	-6.94
	Peak	Contact	6.36 ± 5.17	6.71 ± 4.23	5.30
		Recovery	-3.05 ± 1.21	-3.09 ± 1.19	-1.35
		Full Cycle	6.40 ± 5.10	6.72 ± 4.17	5.00
Internal/ External Rotation Moment [mm]	Avg.	Contact	5.84 ± 2.21	5.59 ± 1.64	4.28
		Recovery	-1.38 ± 0.968	-0.859 ± 0.994	-46.84
		Full Cycle	2.23 ± 1.22	2.37 ± 0.916	6.08
	Peak	Contact	9.67 ± 3.74	8.86 ± 2.79	8.75
		Recovery	4.06 ± 2.00	4.54 ± 2.13	11.20
		Full Cycle	9.68 ± 3.72	8.90 ± 2.77	8.41

Table 5, cont.

Predictor		**No Pain Group Average (n=74)**	**Pain Group Average (n=28)**	**% Difference**
Joint Kinetics, cont.:				
Positive Shoulder Joint Work [mm]	Contact	7.35 ± 3.82	6.45 ± 2.84	13.10
	Recovery	3.96 ± 1.19	4.11 ± 1.03	3.72
	Full Cycle	11.31 ± 4.28	10.55 ± 3.51	6.90
Negative Shoulder Joint Work [mm]	Contact	-0.486 ± 0.466	-0.338 ± 0.263	-35.85
	Recovery	-1.16 ± 0.659	-1.31 ± 0.829	-12.77
	Full Cycle	-1.64 ± 0.847	-1.65 ± 0.913	-0.514
Spatiotemporal measures:				
Contact Time [s]		0.444 ± 0.105	0.460 ± 0.099	3.57
Release Time [s]		0.716 ± 0.199	0.716 ± 0.152	0.119
Cycle Time [s]		1.16 ± 0.267	1.18 ± 0.228	1.31
Cadence [Hz]		0.907 ± 0.207	0.882 ± 0.170	2.80
Contact Percentage [%]		0.387 ± 0.056	0.393 ± 0.045	1.57
Start Angle [°]		-29.12 ± 9.79	-30.77 ± 8.66	-5.53
End Angle [°]		48.56 ± 10.88	46.02 ± 10.89	5.38
Contact Angle [°]		77.68 ± 12.84	76.79 ± 14.83	1.15
NRT [m]		-0.019 ± 0.053	-0.017 ± 0.057	-7.41
TRT [m]		0.046 ± 0.038	0.049 ± 0.036	5.16
NRT/TRT		0.044 ± 0.887	0.091 ± 0.914	70.40

Table 5, cont.

APPENDIX B

Strength		
	Predictor	**p-value**
	Intercept	0.640
1	Adductor Torque [N-m]	0.090

Table 6: Sub-model 1 results. Potential predictors included strength measures with collinear terms removed. All predictors with p<0.1 were added to the final model. Tjur R^2 = 3.0%, overall model p = 0.082.

Handrim Kinetics		
	Predictor	**p-value**
	Intercept	0.493
1	Average F_{tan} [N]	0.093
2	Average F_{rad} [N]	0.040

Table 7: Sub-model 2 results. Potential predictors included handrim kinetics with collinear terms removed. All predictors with p<0.1 were added to the final model. Tjur R^2 = 5.3%, overall model p = 0.055.

Joint Kinetics		
	Predictor	**p-value**
	Intercept	0.069
1	Average Adduction/Abduction Moment during Recovery [mm]	0.096
2	Average Internal/External Rotation Moment during Recovery [mm]	0.013

Table 8: Sub-model 3 results. Potential predictors included joint kinetics with collinear terms removed. All predictors with p<0.1 were added to the final model. Tjur R^2 = 9.2%, overall model p = 0.012.

All Kinetics		
	Predictor	**p-value**
	Intercept	0.028
1	Average F_{rad} [N]	0.004
2	Average Flexion/Extension Moment during Contact [mm]	0.019
3	Average Internal/External Rotation Moment during Recovery [mm]	0.002
4	Negative Shoulder Joint Work during Contact [mm]	0.018
5	Positive Shoulder Joint Work during Recovery [mm]	0.007

Table 9: Sub-model 4 results. Potential predictors included handrim and joint kinetics with collinear terms removed. All predictors with $p<0.1$ were added to the final model. Tjur R^2 = 22.5%, overall model $p < 0.001$.

All Kinetics + Strength		
	Predictor	**p-value**
	Intercept	0.993
1	Adductor Torque [N-m]	0.005*
2	Average F_{tan} [N]	0.080*
3	Average F_{rad} [N]	0.008*
4	Average Flexion/Extension Moment during Contact [mm]	0.547
5	Average Internal/External Rotation Moment during Recovery [mm]	0.001*
6	Negative Shoulder Joint Work during Contact [mm]	0.032*
7	Positive Shoulder Joint Work during Recovery [mm]	0.006*
8	Adductor Torque **:** Average Flexion/Extension Moment during Contact Interaction	0.010*
9	Average F_{tan} **:** Average Flexion/Extension Moment during Contact Interaction	0.031*
10	Average Flexion/Extension Moment during Contact **:** Negative Shoulder Joint Work during Contact Interaction	0.080*

Table 10: Sub-model 5 results. Potential predictors included handrim kinetics, joint kinetics and strength measures with collinear terms removed. All predictors with $p<0.1$ are indicated using '*' and were added to the final model. Tjur R^2 = 35.4%, overall model $p < 0.001$.

Kinematics		
	Predictor	**p-value**
	Intercept	0.034
1	Maximum Trunk Angle [º]	0.019

Table 11: Sub-model 6 results. Potential predictors included shoulder and trunk kinematics with collinear terms removed. All predictors with $p<0.1$ were added to the final model. Tjur R^2 = 6.0%, overall model p = 0.016.

SDs		
	Predictor	**p-value**
	Intercept	0.980
1	SD Minimum Trunk Angle [º]	0.032

Table 12: Sub-model 8 results. Potential predictors included SDs with collinear terms removed. All predictors with $p<0.1$ were added to the final model. Tjur R^2 = 5.0%, overall model p = 0.018.

CVs		
	Predictor	**p-value**
	Intercept	0.617
1	CV Minimum Shoulder Rotation [%]	0.144
2	CV Average F_{tan} [%]	0.033*
3	CV Average F_{rad} [%]	0.002*
4	CV Contact Angle [%]	0.037*
5	CV Average Adduction/Abduction Moment during Contact [%]	0.096*
6	CV Average Adduction/Abduction Moment during Recovery [%]	0.118
7	CV Average Flexion/Extension Moment over Full Cycle [%]	0.109

Table 13: Sub-model 9 results. Potential predictors included CVs with collinear terms removed. All predictors with $p<0.1$ are indicated using ‘*’ and were added to the final model. Tjur R^2 = 24.9%, overall model $p < 0.001$.

APPENDIX C

Final Predictor	β-estimate	p-value	No Pain Group Average (n=58)	Pain Group Average (n=23)
Intercept	6.905	0.346		
Adductor Torque [N-m]	-0.145	0.014*****	73.57 ± 19.41	62.57 ± 19.11
Maximum Trunk Angle [º]	0.187	0.011*****	-7.78 ± 7.53	-3.74 ± 9.22
Average F_{tan} [N]	0.251	0.393	19.64 ± 4.66	18.13 ± 4.19
Average F_{rad} [N]	-0.093	0.280	-15.99 ± 5.91	-19.03 ± 5.60
Average Flex/Ext Moment during Contact [mm]	-0.401	0.634	7.85 ± 3.58	7.35 ± 3.09
Average Ab/Adduction Moment during Recovery [mm]	0.423	0.407	-5.99 ± 1.03	-6.19 ± 0.894
Negative Shoulder Joint Work during Contact [mm]	4.376	0.185	-0.482 ± 0.475	-0.319 ± 0.245
Positive Shoulder Joint Work during Recovery [mm]	1.017	0.064†	3.96 ± 1.27	3.94 ± 0.866
SD Minimum Trunk Angle [º]	-2.184	0.105	0.881 ± 0.485	0.600 ± 0.316
CV Average F_{tan} [%]	-0.037	0.637	12.58 ± 5.50	12.06 ± 4.80
CV Average F_{rad} [%]	-0.137	0.105	17.86 ± 13.76	10.19 ± 3.83
CV Average Ab/Adduction Moment during Contact [%]	0.003	0.037*****	151.2 ± 444.5	252.5 ± 595.6
CV Contact Angle [%]	0.286	0.215	4.64 ± 2.58	5.12 ± 2.08
Adductor Torque [N-m] **:** Average Flex/Ext Moment during Contact [mm]	0.018	0.016*****		
Average F_{tan} [N] **:** Average Flex/Ext Moment during Contact [mm]	-0.060	0.145		
Average Flex/Ext Moment during Contact [mm] **:** Negative Shoulder Joint Work during Contact [mm]	-0.497	0.159		
Overall Model		1.13E-4		
		Tjur R^2		
Training Group (Groups 1-4)		49.5%		
Test Group (Group 5)		15.6%		

Table 14: Final model values for Model 1 of the 5-fold cross validation procedure. Groups 1-4 (n=81) were used to train the model. "*" indicates a significant correlation (p<0.05). "†" indicates approaching significance (p<0.1).

Final Predictor	β-estimate	p-value	No Pain Group Average (n=60)	Pain Group Average (n=22)
Intercept	3.140	0.632		
Adductor Torque [N-m]	-0.176	0.012*	71.20 ± 20.22	66.20 ± 20.41
Maximum Trunk Angle [°]	0.190	0.008*	-8.04 ± 7.37	-2.77 ± 9.29
Average F_{tan} [N]	0.367	0.225	19.44 ± 4.88	18.72 ± 4.39
Average F_{rad} [N]	-0.128	0.123	-16.24 ± 6.33	-18.71 ± 6.20
Average Flex/Ext Moment during Contact [mm]	-0.989	0.156	7.59 ± 3.51	7.15 ± 3.01
Average Ab/Adduction Moment during Recovery [mm]	0.183	0.721	-6.01 ± 1.03	-6.41 ± 0.962
Negative Shoulder Joint Work during Contact [mm]	12.427	0.038*	-0.458 ± 0.440	-0.308 ± 0.247
Positive Shoulder Joint Work during Recovery [mm]	1.645	0.010*	3.94 ± 1.16	4.10 ± 1.10
SD Minimum Trunk Angle [°]	-0.776	0.465	0.812 ± 0.468	0.651 ± 0.357
CV Average F_{tan} [%]	0.070	0.326	11.83 ± 5.54	11.80 ± 5.08
CV Average F_{rad} [%]	-0.126	0.079†	16.03 ± 11.26	11.98 ± 6.10
CV Average Ab/Adduction Moment during Contact [%]	0.003	0.100	95.16 ± 167.4	258.1 ± 609.2
CV Contact Angle [%]	0.322	0.136	4.50 ± 2.51	5.30 ± 2.61
Adductor Torque [N-m] **:** Average Flex/Ext Moment during Contact [mm]	0.023	0.014*		
Average F_{tan} [N] **:** Average Flex/Ext Moment during Contact [mm]	-0.071	0.125		
Average Flex/Ext Moment during Contact [mm] **:** Negative Shoulder Joint Work during Contact [mm]	-1.428	0.040*		
Overall Model		2.64E-4		
		Tjur R^2		
Training Group (Groups 1-3, 5)		47.5%		
Test Group (Group 4)		29.4%		

Table 15: Final model values for Model 2 of the 5-fold cross validation procedure. Groups 1-3, 5 (n=82) were used to train the model. "*" indicates a significant correlation (p<0.05). "†" indicates approaching significance (p<0.1).

Final Predictor	β-estimate	p-value	No Pain Group Average (n=62)	Pain Group Average (n=19)
Intercept	0.411	0.953		
Adductor Torque [N-m]	-0.151	0.024*	72.95 ± 19.04	58.24 ± 20.74
Maximum Trunk Angle [°]	0.217	0.007*	-8.13 ± 7.42	-4.30 ± 8.75
Average F_{tan} [N]	0.450	0.130	19.33 ± 5.04	18.39 ± 4.59
Average F_{rad} [N]	-0.100	0.196	-16.20 ± 6.63	-18.49 ± 6.46
Average Flex/Ext Moment during Contact [mm]	-0.306	0.637	7.60 ± 3.71	7.16 ± 3.29
Average Ab/Adduction Moment during Recovery [mm]	0.004	0.994	-6.02 ± 0.950	-6.53 ± 0.913
Negative Shoulder Joint Work during Contact [mm]	6.293	0.062†	-0.493 ± 0.479	-0.404 ± 0.262
Positive Shoulder Joint Work during Recovery [mm]	1.002	0.048*	4.02 ± 1.21	4.22 ± 1.02
SD Minimum Trunk Angle [°]	-1.103	0.367	0.818 ± 0.488	0.675 ± 0.365
CV Average F_{tan} [%]	0.054	0.494	11.75 ± 5.46	12.27 ± 5.67
CV Average F_{rad} [%]	-0.086	0.246	16.61 ± 13.11	11.35 ± 6.56
CV Average Ab/Adduction Moment during Contact [%]	0.002	0.058†	120.9 ± 405.2	226.2 ± 613.7
CV Contact Angle [%]	0.236	0.292	4.62 ± 2.50	5.23 ± 2.40
Adductor Torque [N-m] **:** Average Flex/Ext Moment during Contact [mm]	0.016	0.058†		
Average F_{tan} [N] **:** Average Flex/Ext Moment during Contact [mm]	-0.070	0.103		
Average Flex/Ext Moment during Contact [mm] **:** Negative Shoulder Joint Work during Contact [mm]	-0.698	0.058†		
Overall Model		2.05E-3		
		Tjur R^2		
Training Group (Groups 1-2, 4-5)		41.8%		
Test Group (Group 3)		39.7%		

Table 16: Final model values for Model 3 of the 5-fold cross validation procedure. Groups 1-2, 4-5 (n=81) were used to train the model. "*" indicates a significant correlation ($p<0.05$). "†" indicates approaching significance ($p<0.1$).

Final Predictor	β-estimate	p-value	No Pain Group Average (n=57)	Pain Group Average (n=25)
Intercept	5.250	0.463		
Adductor Torque [N-m]	-0.206	0.007*	69.20 ± 18.49	64.47 ± 21.64
Maximum Trunk Angle [°]	0.150	0.027*	-7.70 ± 6.49	-3.91 ± 8.30
Average F_{tan} [N]	0.462	0.131	19.60 ± 5.03	17.95 ± 4.65
Average F_{rad} [N]	-0.012	0.920	-16.64 ± 6.11	-17.78 ± 5.05
Average Flex/Ext Moment during Contact [mm]	-0.774	0.303	7.55 ± 3.35	6.70 ± 2.68
Average Ab/Adduction Moment during Recovery [mm]	0.187	0.708	-6.04 ± 1.06	-6.30 ± 1.01
Negative Shoulder Joint Work during Contact [mm]	3.572	0.274	-0.508 ± 0.468	-0.318 ± 0.268
Positive Shoulder Joint Work during Recovery [mm]	1.130	0.029*	4.02 ± 1.15	4.08 ± 1.06
SD Minimum Trunk Angle [°]	-1.394	0.265	0.859 ± 0.488	0.612 ± 0.340
CV Average F_{tan} [%]	0.008	0.907	12.45 ± 5.83	12.48 ± 5.01
CV Average F_{rad} [%]	-0.222	0.046*	17.38 ± 13.04	11.34 ± 6.02
CV Average Ab/Adduction Moment during Contact [%]	0.004	0.069†	144.9 ± 446.8	120.61 ± 244.7
CV Contact Angle [%]	0.590	0.018*	4.39 ± 2.44	5.38 ± 2.53
Adductor Torque [N-m] **:** Average Flex/Ext Moment during Contact [mm]	0.033	0.004*		
Average F_{tan} [N] **:** Average Flex/Ext Moment during Contact [mm]	-0.096	0.035*		
Average Flex/Ext Moment during Contact [mm] **:** Negative Shoulder Joint Work during Contact [mm]	-0.390	0.312		
Overall Model		3.03E-5		
		Tjur R^2		
Training Group (Groups 1, 3-5)		51.7%		
Test Group (Group 2)		22.4%		

Table 17: Final model values for Model 4 of the 5-fold cross validation procedure. Groups 1, 3-5 (n=82) were used to train the model. "*" indicates a significant correlation ($p<0.05$). "†" indicates approaching significance ($p<0.1$).

Final Predictor	β-estimate	p-value	No Pain Group Average (n=59)	Pain Group Average (n=23)
Intercept	1.892	0.772		
Adductor Torque [N-m]	-0.187	0.029*	70.47 ± 20.28	67.07 ± 20.14
Maximum Trunk Angle [°]	0.231	0.003*	-7.36 ± 7.72	-3.11 ± 9.12
Average F_{tan} [N]	0.639	0.153	19.38 ± 4.70	18.64 ± 4.96
Average F_{rad} [N]	-0.131	0.125	-16.40 ± 6.62	-18.91 ± 6.20
Average Flex/Ext Moment during Contact [mm]	-0.439	0.490	7.88 ± 3.61	7.36 ± 3.30
Average Ab/Adduction Moment during Recovery [mm]	0.476	0.365	-6.10 ± 1.04	-6.34 ± 1.03
Negative Shoulder Joint Work during Contact [mm]	6.764	0.154	-0.489 ± 0.469	-0.355 ± 0.285
Positive Shoulder Joint Work during Recovery [mm]	1.324	0.017*	3.84 ± 1.17	4.20 ± 1.09
SD Minimum Trunk Angle [°]	-3.306	0.038*	0.839 ± 0.452	0.577 ± 0.304
CV Average F_{tan} [%]	-0.034	0.792	11.83 ± 4.83	11.88 ± 4.41
CV Average F_{rad} [%]	-0.063	0.429	16.33 ± 12.58	11.55 ± 6.07
CV Average Ab/Adduction Moment during Contact [%]	0.002	0.025*	148.1 ± 441.2	220.8 ± 599.3
CV Contact Angle [%]	0.481	0.023*	4.32 ± 2.15	5.50 ± 2.52
Adductor Torque [N-m] **:** Average Flex/Ext Moment during Contact [mm]	0.024	0.017*		
Average F_{tan} [N] **:** Average Flex/Ext Moment during Contact [mm]	-0.092	0.086†		
Average Flex/Ext Moment during Contact [mm] **:** Negative Shoulder Joint Work during Contact [mm]	-0.744	0.141		
Overall Model		1.67E-4		
		Tjur R^2		
Training Group (Groups 2-5)		47.6%		
Test Group (Group 1)		29.7%		

Table 18: Final model values for Model 5 of the 5-fold cross validation procedure. Groups 2-5 (n=82) were used to train the model. "*" indicates a significant correlation (p<0.05). "†" indicates approaching significance (p<0.1).

APPENDIX D

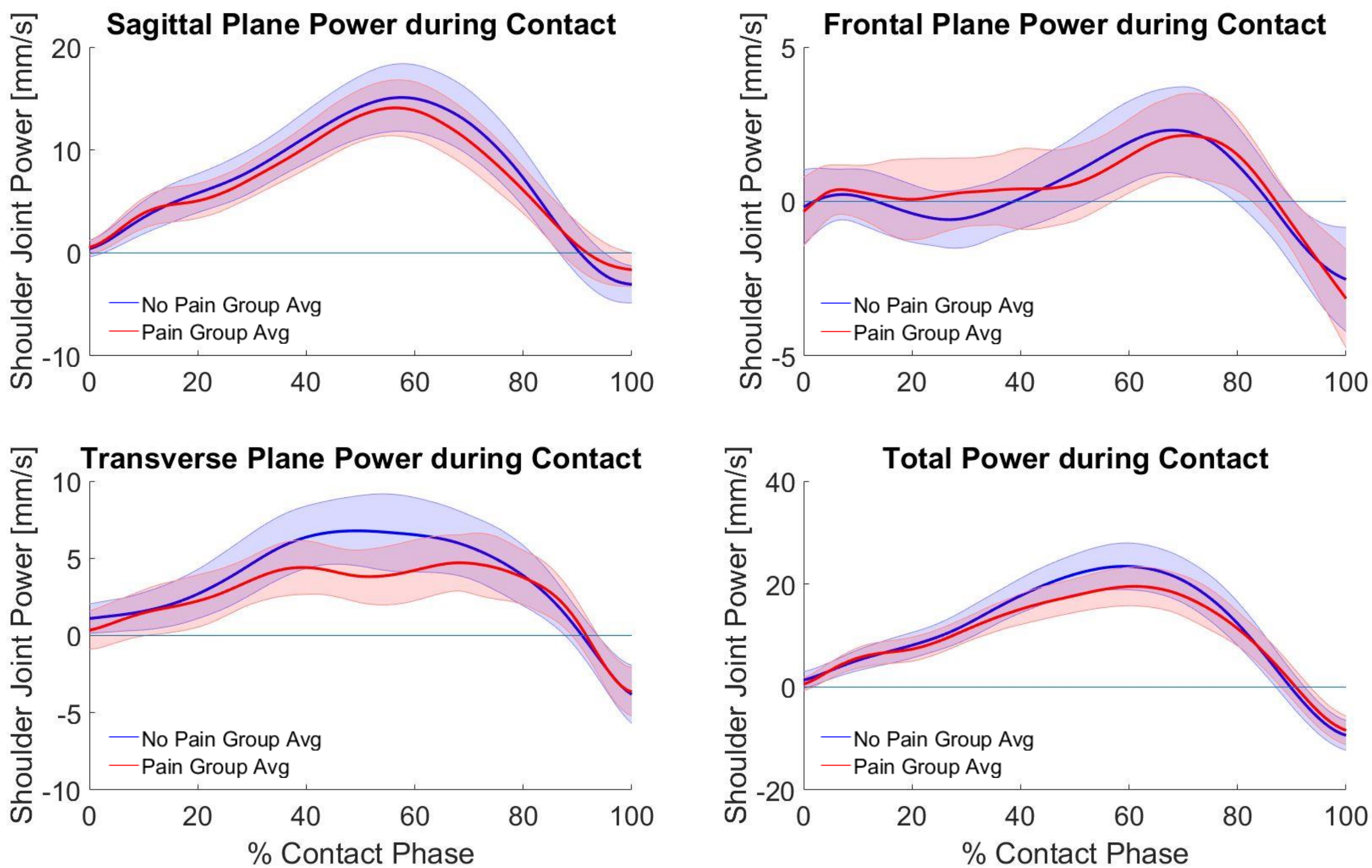

Figure 5: Shoulder joint power in each direction during the contact phase with one standard deviation from the mean shaded. Joint power is normalized by body weight (N-mm/N-s).

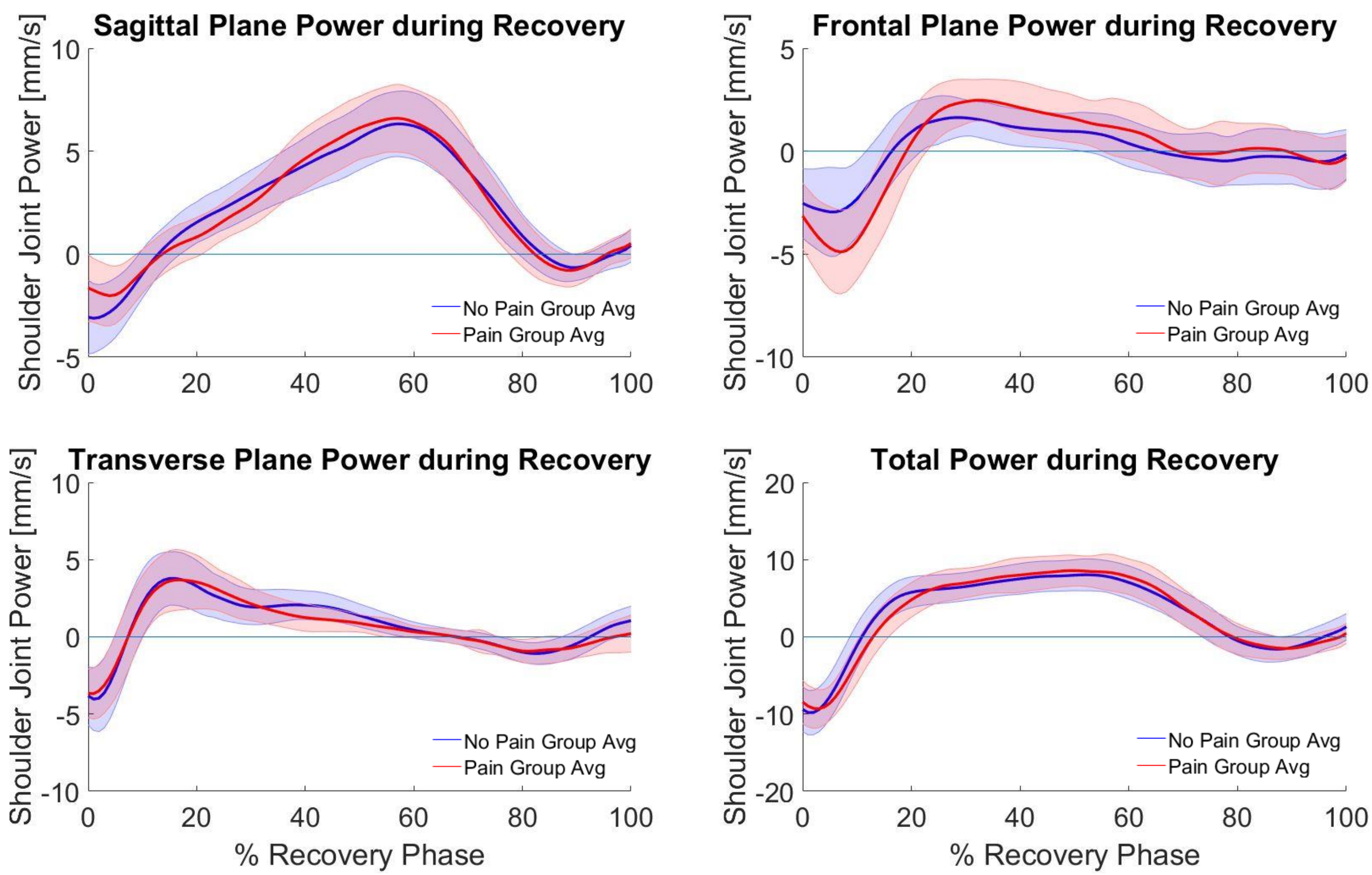

Figure 6: Shoulder joint power in each direction during the recovery phase with one standard deviation from the mean shaded. Joint power is normalized by body weight (N-mm/N-s).

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