SHOULDER PAIN IN MANUAL WHEELCHAIR USERS: TOWARDS A MULTI-DISCIPLINARY SOLUTION FOR A MULTI-FACETED PROBLEM

BY

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DISSERTATION

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# ABSTRACT

It is estimated that there are over 2 million manual wheelchair users in the United States. Up to 70% of manual wheelchair users report upper limb pain, which is mainly manifested in the shoulder and wrist. Shoulder pain in wheelchair users is linked to difficulty performing activities of daily living, decreased physical activity and decreased quality of life.

The main focus of this dissertation is to identify biomarkers from wheelchair propulsion data that are potentially related to shoulder pain in manual wheelchair users. Three biomarkers that distinguish between manual wheelchair users with and without shoulder pain are identified. The acceptability of the identified biomarkers are subjected to hypothesis testing using data collected from a sample of 30 experienced adult manual wheelchair users with and without shoulder pain. The results and their implications will be discussed. In this dissertation we will also discuss the interpretation and the physical significance of each of the results, a summary of limitations for the approaches adopted, and suggestions on the future course of research to address these limitations.

While the past two decades of research on shoulder pain and wheelchair propulsion has led to the development of important clinical guidelines, it has failed to identify specific biomarkers that may be related to shoulder pain in manual wheelchair users. This could be in part due to employing a binary approach by focusing on just (1) the pure bio-mechanical

aspects, and (2) wheelchair design aspects (ergonomics). The originality of this dissertation is in the adoption of a multidisciplinary approach. Methodologies integrating theories and analyses from fields related to human movement science such as human motor control theory, non-linear dynamics and human factors (occupational ergonomics) are adopted to identify potential biomarkers that relate to shoulder pain in manual wheelchair users.

This dissertation concludes with preliminary results from a prototype wearable device, custom developed for manual wheelchair users. Wheelchair propulsion data obtained from the device will be benchmarked with data from the currently available technologies for tracking manual wheelchair propulsion (SMARTWheel and motion capture). This dissertation also proposes a framework for incorporating the research findings into the custom developed wearable technology for home-based rehabilitation training purposes.

*To my grandmother, mother, & brother
for their continued support all along,
And
Shri. Gopalakrishnan for being a
constant source of moral guidance.*

आचार्यात् पादमादत्ते पादं शिष्यः स्वमेधया ।
सब्रह्मचारिभ्यः पादं पादं कालक्रमेण च ॥

*Sanskrit verse, Rig Veda*

Translates as-

*Knowledge acquisition*

*one fourth from the teacher,*
*one fourth from own intelligence,*
*one fourth from fellow classmates, and*
*one fourth only with time.*

# ACKNOWLEDGEMENTS

# TABLE OF CONTENTS

# Chapter 1

## Introduction and motivation

It is estimated that over 3.6 million Americans use wheelchairs for mobility (LaPlante et al., 2010) with a majority (~90%) using a manual wheelchair (LaPlante et al., 2010). Although wheelchair use has numerous benefits (Hosseini et al., 2012), the repetitive cyclic arm movement required for manual propulsion places a significant demand on the upper extremity, specifically the shoulder (Nichols et al., 1979, Gellman et al., 1988, Curtiset al., 1999, Finley et al., 2004). This increased demand often results in shoulder pain. Indeed up to 70% of manual wheelchair users report shoulder pain (Finley et al., 2004).

Shoulder pain in wheelchair users have been linked to difficulty performing activities of daily living, decreased physical activity and decreased quality of life (Chow et al., 2011). Subsequently, it is imperative to understand the mechanisms that contribute to shoulder pain in manual wheelchair users so that appropriate interventions can be developed to prevent or minimize the effect of shoulder pain on function and thus reduce the risk of long-term upper extremity disability.

Shoulder pain occurring in wheelchair users is a multi-faceted problem (Dyson-Hudson et al., 2004). It has been suggested that upper limb pain is related to a myriads of factors including functional level (Curtis KA et al., 1999), duration of wheelchair use, wheelchair design (van der Woude et al., 2006), body weight  (Sinnott et al., 2000, Collinger et al.,  2008), propulsion mechanics (Koontz et al., 2002, Mercer et al., 2006) muscle coordination (Burhham et al., 1993, Kotajarvi et al., 2002), age (Fullerton et al., 2003), and gender (Lal S, 1998, Gutierrez et al., 2007) . The multi-factorial nature of the possible mechanisms and associated variables creates a daunting task for researchers and clinicians.

Given that the association between shoulder pain and manual wheelchair propulsion is multi-faceted in nature, a multi-disciplinary approach to analyze and address this problem is needed. Such multi-disciplinary approaches can provide better understanding of the pathology and may lead to new knowledge for better monitoring/tracking/ and prevention of shoulder injury in manual wheelchair users.

Investigations of mechanisms contributing to shoulder pain in wheelchair users have mainly examined biomechanical variables of manual wheelchair propulsion (Boninger et al., 2002, Collinger et al., 2008, Rankin et al., 2011, Richter et al., 2007, 2011, Koontz et al., 20009, 2012, de Groot et al., 2003,2004, Kwarciak et al., 2012 )   and wheelchair design aspects

(van der Woude et al., 2006, Boninger et al., 2005, Cowan et al., 2009). While the research on shoulder pain and wheelchair propulsion has provided important information and has led to the development of clinical guidelines (Boninger et al., 2002, Collinger et al., 2008, Rankin et al., 2011, 2012, Richter et al., 2007, 2011, Koontz et al., 20009, 2012, de Groot et al., 2003, 2004, Kwarciak et al., 2012, Mercer et al., 2006), it has several potential limitations.

First, a multi-disciplinary approach to examine wheelchair propulsion has not been adopted. Second, research has mainly focused on the complete propulsion cycle and the push phase, but much less so on the recovery phase. Third, there has been minimal enquiry on the temporal structure of motor variability in wheelchair propulsion in the context of shoulder pain.

**Research objectives of this dissertation**

Consequently the main objectives of this dissertation is to implement a multi-disciplinary approach to identify biomarkers from wheelchair propulsion data that could be potentially related to shoulder pain in manual wheelchair users. Methodologies integrating theories and analyses from fields related to human movement science such as human motor control theory, non-linear dynamics and human factors (occupational ergonomics) are adopted to identify potential biomarkers that relate to shoulder pain in

manual wheelchair users. Three biomarkers that distinguish between manual wheelchair users with and without shoulder pain are identified. These three biomarkers are: (1) differences in kinematic jerk during recovery phase between individuals propelling a manual wheelchair with and without shoulder pain (Chapter 3), (2) trunk kinematic differences between individuals propelling a manual wheelchair with and without shoulder pain (Chapter 4) and (3) the structure in cycle-to-cycle variability of wheelchair propulsion variables between individuals propelling a manual wheelchair with and without shoulder pain (Chapter 5).

The acceptability of the identified biomarkers are evaluated sing hypothesis testing data collected from a sample of 30 experienced adult manual wheelchair users with and without shoulder pain. The results and their implications will be discussed. This dissertation will also discuss the interpretation and the physical significance of each of the results, a summary of limitations for the approaches adopted, and suggestions on the future course of research to address these limitations.

**Dissertation organization and chapter associations**

This dissertation is organized as follows:

*Chapter 2* describes the experimental data collection methodologies. The data derived from the same experimental setup was used to hypotheses test

all three studies reported in Chapters 3, 4 and 5.

*Chapter 3* contains the motivation, methodologies, results and discussion for Study 1: Shoulder pain and jerk during wheelchair propulsion.

*Chapter 4* reports the motivation, methodologies, results and discussion for Study 2: Shoulder pain and trunk kinematics during wheelchair propulsion.

*Chapter 5* contains details about Study 3 which investigates the relationship between the cycle-to-cycle time dependent structure in variability and shoulder pain in manual wheelchair users.

*Chapter 6* takes the insights from the three different studies and proposes a framework for the implementation of these metrics through a wearable technology for day-to-day monitoring of wheelchair user propulsion mechanics.

*Chapter 7* contains the overall conclusion and suggestions for future direction of research. Chapter 8 contains list of intellectual contribution this dissertation to wheelchair propulsion literature.


**Association between chapters** (Figure 1.1)

Chapter's 1 and 2 contain the necessary preliminary background materials for Chapters 3, 4 and 5. The Chapters 3, 4 and 5 are not inter-related and can be read in any order. However, it is recommended that to fully appreciate the contents of Chapters 6, 7 and 8, all the previous chapters be read (Figure 1.1).

**The multi-disciplinary approach adopted**

Contrary to traditional approaches, this dissertation adopts a multi-disciplinary approach to identify biomarkers that related to shoulder pain in manual wheelchair users (Figure 1.2). The multi-disciplinary area that this dissertation relies on were not considered in previous in wheelchair propulsion literature. The previous literature focused on just two aspects namely, propulsion biomechanics and wheelchair ergonomics (design aspects), while this dissertation provides a new dimension to this problem integrating approaches from human movement science disciplines like, human motor control theory, non-linear dynamics and human movement ergonomics (occupational ergonomics).

**Basic wheelchair propulsion terminologies**

Before we elaborate further on each of the biomarkers considered, it is important to define some standard manual wheelchair propulsion terminologies used throughout this dissertation.

A typical manual wheelchair propulsion cycle consists of a push phase (i.e. when the hand is in contact with the hand-rim/wheel) and a recovery phase (when the hand is off the hand-rim/wheel). During the push phase the arms are constrained to follow the hand-rims, while during recovery phase the arms can adopt a variety of different movement patterns (Figure 1.1). Four typical propulsion pattern types have been observed based on the hand

trajectory during the recovery phase of manual wheelchair propulsion. They are a semi-circular (SC) pattern, double loop pattern (DLOP), single loop pattern (SLOP) and an arc pattern (Sanderson et al., 1985, Shimada et al., 1988, Boninger et al., 2002, Richter et al., 2007).

A DLOP pattern is characterized by the hands lifting over the propulsion path and crossing the propulsion path to drop below the hand-rim forming a double loop, while a SC pattern is characterized by the hands dropping below the hand-rim during the recovery phase, the hands rise above the hand rim during the recovery phase for a SLOP pattern and ) the hands follows the hand rim closely during the recovery phase for a ARC pattern forming a pumping action (Figure 1.2) (Sanderson et al., 1985, Shimada et al., 1988, Boninger et al., 2002, Richter et al., 2007).

In this dissertation is goes, to capture, motion data (i.e movement kinematics), a motion capture system was used to capture movement kinematics. The procedures and experimental setup for this are detailed in Chapter 2 of this dissertation.

*Peak resultant force at hand-rim (Newton (N)):* The peak value of the resultant force applied at the hand-rim by the palm to push the wheelchair (Figure 1.4(a, b)).

*Contact angle (degrees ($\theta^o$)):* The angle for which the hand is in contact with the hand-rim during the push phase of wheelchair propulsion (i.e. while

pushing the wheelchair (Figure 1.4 (b)).

*Inter push time interval between peak resultant force* *(seconds (sec)):* The time interval between two consecutive peak resultant forces at hand-rim (Figure 1.4 (b)).

*Push time* *(seconds (sec))*: The time taken from the start to end of a push phase (Figure 1.3 (b)).

*Recovery time* *(seconds (sec))*: The time taken from the end of a push phase to the start of the consecutive push phase (Figure 1.4 (b)).

*Steady state propulsion*: The portion of the wheelchair propulsion trial from the 6th push till the end of the trial (Figure 1.4 (a)). The first 5 pushes are influenced by initial start-up effects in overcoming inertia and research evidence suggests that it usually takes up to five pushes to reach a steady state pushing rate (Koontz et al., 2009).

**List of figures**
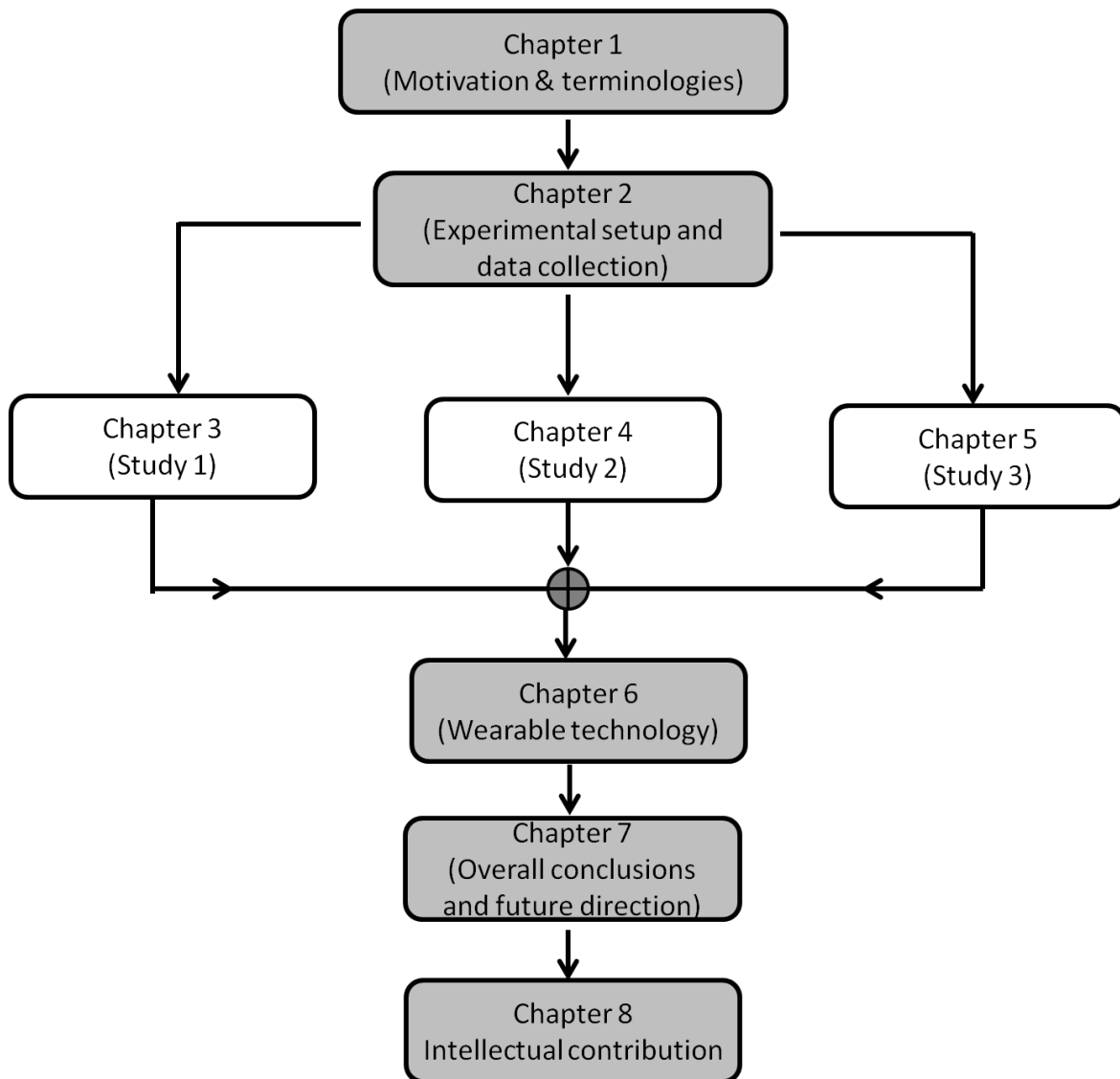


Figure 1.1 Association between chapters
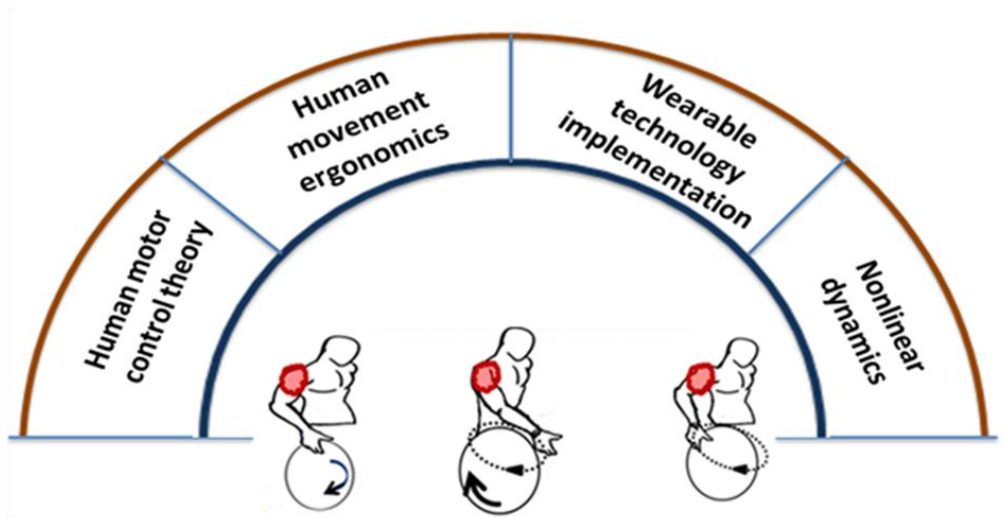
**Multi-disciplinary approach**

Figure 1.2 Various multi-disciplinary approaches this dissertation adopts to investigate shoulder pain in manual wheelchair users
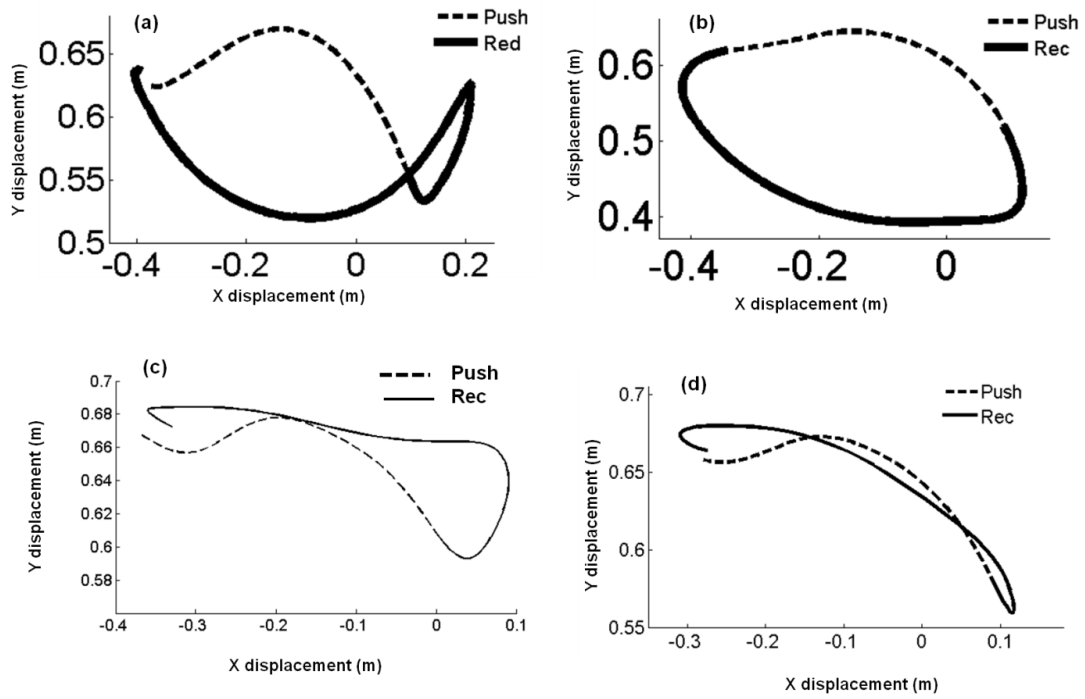
Figure 1.3 Different recovery pattern types. (a) Double loop (DLOP); (b) Semi-circular (SC); (c) Single loop (SLOP) and (d) Arc. The dotted lines denote the push phase and the solid line the recovery phase.
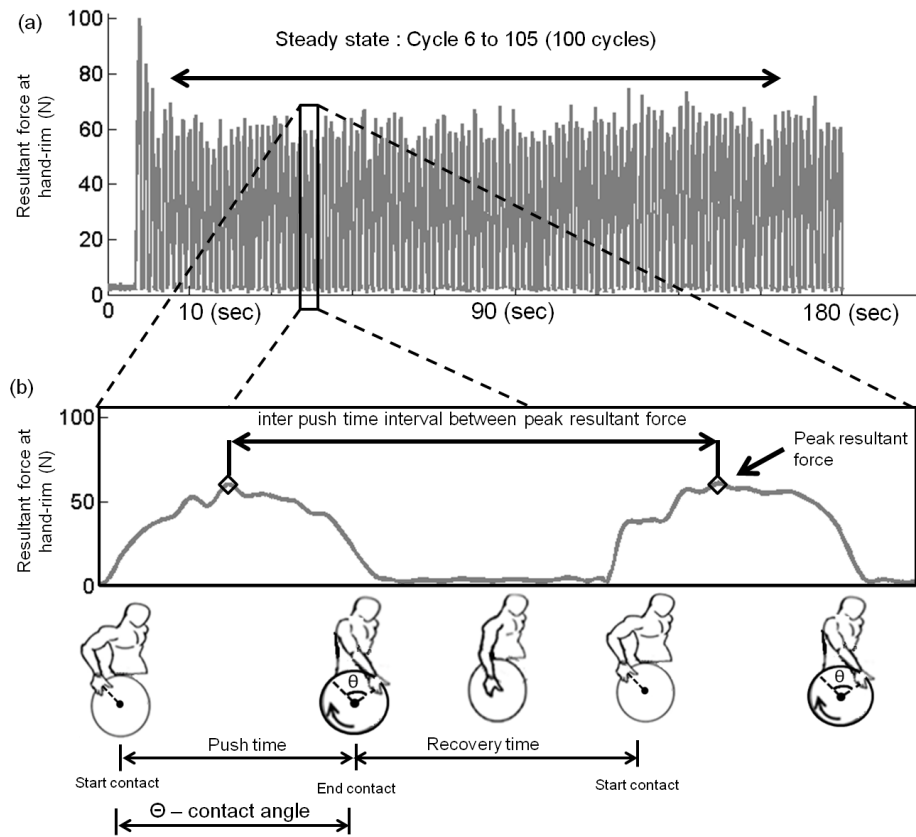
Figure 1.4 Basic terminologies of wheelchair propulsion mechanics used in this dissertation. (a) A plot showing the resultant force at hand-rim for a 3 minute trial with approximately 120 pushes. The steady state portion of the trial is also shown; (b) a magnified view of two sample pushes cycles from the full trial showing all the important terminologies, namely, contact angle, peak resultant force, inter push time interval between peak resultant force, push time and recovery time.

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# Chapter 2

## Experimental data collection procedures

This chapter describes the manual wheelchair propulsion data collection procedures and experimental configuration used in detail. The data analyses reported in Chapters 3, 4 and 5 are all based on this data.

## Participants

Wheelchair propulsion data from 30 individuals between the age of 18 to 65 years, with a range of physical disabilities and a mix of genders (male (n=17) and female (n=13)) from the Urbana-Champaign campus community were collected. Inclusion criteria were: (1) between 18-65 years old and (2) use of a manual wheelchair as their primary means of mobility for 1+ year.

Participants were classified into "with shoulder pain" and "without shoulder pain" groups based on their self-report ("Yes"/"No"- written response) of shoulder pain to our demographic questionnaire provided at the time of data collection.

**Experimental protocol**

All experimental protocols in this study were approved by the University of Illinois at Urbana-Champaign institutional review board. Upon arrival to the laboratory, the experimental procedures were described to the participants and any questions they had regarding the protocol were clarified. Once participants understood the experimental procedures, they voluntarily signed the institutionally approved informed consent form. The participants then provided demographic information (age, height, weight, duration of wheelchair use, diagnosis, pain status, etc). In addition to self reporting their current status of shoulder pain ("Yes"/"No"), participants also rated their current level of shoulder pain on a 10 cm visual analog scale (VAS) (Campbell et al., 1990) and using the wheelchair user shoulder pain index (WUSPI) (Curtis et al., 1995).

A VAS score of 0 indicated that the participant was not experiencing any shoulder pain at the time of data collection and a score of 10 indicated existence of high level of shoulder pain at the time of data collection. The wheelchair user's shoulder pain index (WUSPI) is based on a 15-item questionnaire (Curtis et al., 1995). Each item is rated between 0 to 10, with 0 representing no interference with functional activities and 10 representing complete interference during the past week due to shoulder pain. The total score is the sum of scores of each of the 15 items. Total scores ranged from

0 (no pain) to 150 (maximum limitations to daily activities due to pain) (Curtis et al., 1995). Participants were separated into pain or no pain groups based on self-report of shoulder pain.

Following the voluntary disclosure of the self-reported shoulder pain scores the participant's, wheelchair configuration measurements and upper extremity anthropometry were measured. The wheelchair configuration measurements included, the shoulder X and Z coordinate positions with respect to the axle (XPOS and YPOS; Figure 2.1), camber, and wheel diameter. The upper extremity anthropometry measurements included torso length, upper arm length, forearm length, wrist circumference, elbow circumference, humerus circumference and knuckle circumference.

Following the collection of all volunteer demographic and shoulder pain data, the participants' personal wheelchair fitted bilaterally with 25 inch diameter force sensing SMARTWheels (Three Rivers Holdings LLC; AZ, USA). An individuals' upper extremity kinematics is not significantly affected by attaching to/testing with different SMARTWheel sizes (Mason et al., 2012). Attaching the SMARTWheels to the participant's personal wheelchair does not change the wheel placement alignment or camber (Mason et al., 2012). Each participant's wheelchair was then secured to a single drum dynamometer with a fly wheel and tie-down system (Figure 2.2). The use of force sensing wheels (Figure 2.2) allowed for the determination of temporal-

spatial and kinetic data relating to wheelchair propulsion. The reference axes orientations are shown in Figure 2.2.

**Kinematic data collection: Motion Capture**

Based on the International Society of Biomechanics (ISB) recommendations (Wu et al., 2005), 18 reflective markers were attached at specific bony landmarks to define the trunk, upper arm, forearm, hand, sternum and the jaw: these included sternal notch, C7 vertebrae, T3 vertebrae, T6 vertebrae and bilaterally at the mandible, third metacarpophalangeal joint, radial styloid ulnar styloid, olecronon, lateral epicondyl and the acromion process (Figure 2.2). Two reflective markers, one on the wheel center and the other on the wheel spoke were placed on each of the wheels (Figure 2.2). Kinematic data were collected using a 10 camera motion capture system (Cortex 2.5, Motion Analysis Co.; Santa Rosa, CA, USA) at a sampling rate of 100Hz.

Participants were asked to propel at constant speeds for three separate 3 minute trials at 1.1m/s (fast), 0.7m/s (slow) and self-select (~0.89 m/s) speeds. The sequence of trial speeds was randomized for each participant. A speedometer was used to provide real-time visual feedback to the subjects while kinetic data were collected bilaterally at 100Hz. Sufficient rest and recovery was provided between each trial. Subjects were

given time to acclimate to the dynamometer and propulsion speed before

the beginning of each trial. A force plate was used to measure the weight

of participants (AMTI, Inc., Watertown, MA, USA).
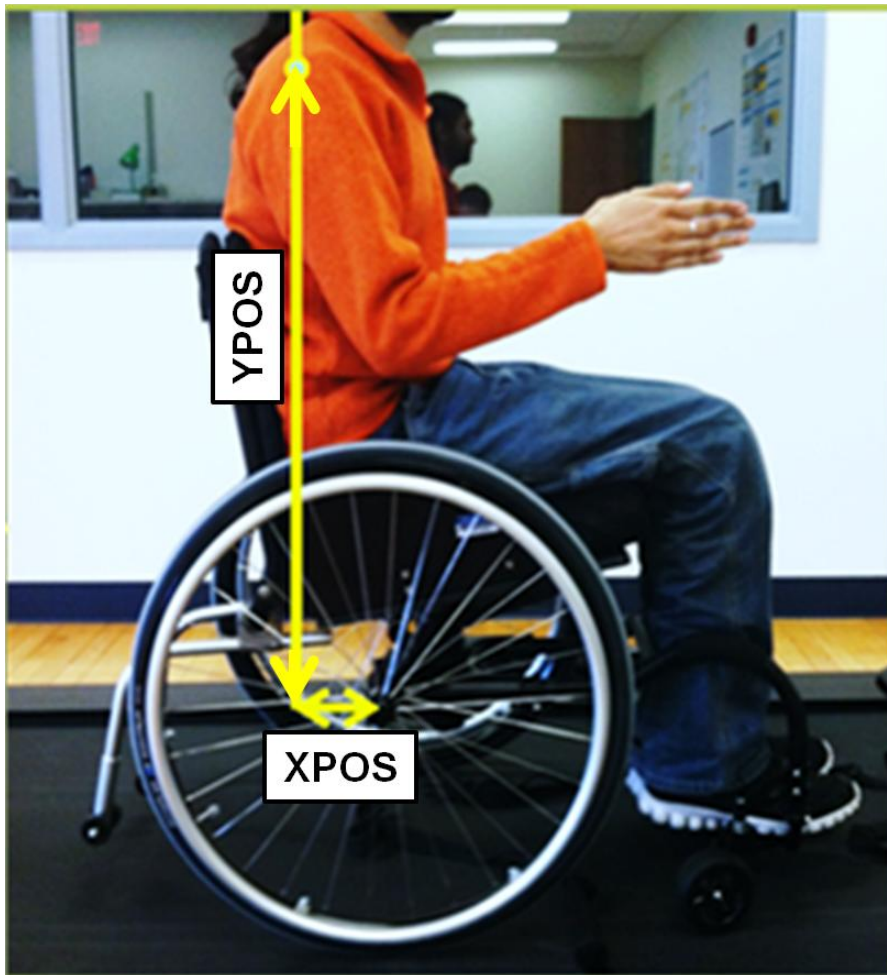
Figure 2.1 The shoulder horizontal and vertical position with respect to the rear axle. XPOS: the horizontal distance between the rear axle and the shoulder, YPOS : the vertical distance between the rear axle and the shoulder.
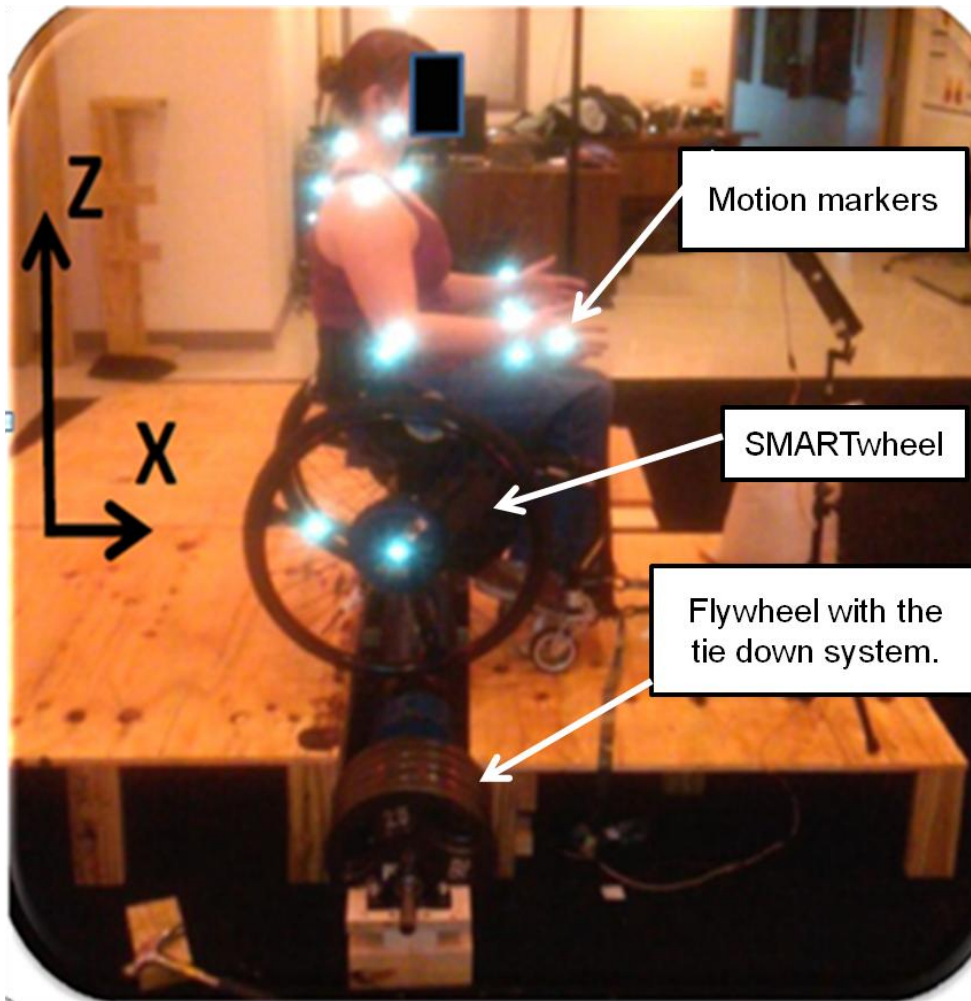
Figure 2.2 Experimental setup in which data was collected for this dissertation. The motion capture marker placements, the instrumented SMARTWheel, the roller dynamometer setup with the flywheel and tie down system and the axes orientations are shown.

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# Chapter 3

## Shoulder pain and jerk during wheelchair propulsion

**Chapter abstract:** Repetitive loading of the upper limb due to wheelchair propulsion plays a leading role in the development of shoulder pain in manual wheelchair users (mWCUs). There has been minimal inquiry on understanding wheelchair propulsion kinematics from a human movement ergonomics perspective. This investigation employs an ergonomic metric, jerk, to characterize the recovery phase kinematics of two recommended manual wheelchair propulsion patterns: semi-circular and the double loop. Further it examines if jerk is related to shoulder pain in manual wheelchair users. Data from 22 experienced adult mWCUs was analyzed for this study (semi-circular: n=12 (pain/without-pain:6/6); double-loop: n=10 (pain/without-pain:4/6)). Participants propelled their own wheelchair fitted with SMARTWheels on a roller dynamometer at 1.1 m/s for 3 minutes. Kinematic and kinetic data of the upper limbs were recorded. Three dimensional absolute jerk experienced at the shoulder, elbow and wrist joint during the recovery phase of wheelchair propulsion were computed. Two-way ANOVAs were conducted with the propulsion pattern type and shoulder pain as between group factors.

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*Findings:* (1) Individuals using a semi-circular pattern experienced lower jerk at their arm joints than those using a double loop pattern ($P<0.05$, $\eta^2=0.32$)$_{wrist}$;($P=0.05$, $\eta^2=0.19$)$_{elbow}$;($P<0.05$, $\eta^2=0.34$)$_{shoulder}$ and (2) individuals with shoulder pain had lower peak jerk magnitude during the recovery phase ($P\leq0.05$, $\eta^2=0.36$)$_{wrist}$;($P\leq0.05$, $\eta^2=0.30$)$_{elbow}$;($P\leq0.05$, $\eta^2=0.31$)$_{shoulder}$. *Conclusions:* Jerk during wheelchair propulsion was able to distinguish between pattern types (semi-circular and double loop) and the presence of shoulder pain. Jerk provides novel insights into wheelchair propulsion kinematics and in the future it may be beneficial to incorporate jerk based metrics into rehabilitation practice.

**Introduction**

Approximately 2 million Americans use a manual wheelchair for mobility (LaPlante et al., 2010). Although the use of a manual wheelchair provides numerous benefits (Hosseini et al., 2012), the repetitive strain encountered by the upper limb during propulsion places significant demand on the tissues (Nichols et al., 1979, Gellman et al., 1988, Curtiset al., 1999, Finley et al., 2004) and has been implicated in upper limb injury (Cooper et al., 1998). Indeed up to 70% of manual wheelchair users (mWCUs) report upper extremity pain (Finley et al., 2004). Upper extremity injury in mWCUs has been linked to difficulty performing activities of daily living, decreased physical activity and decreased quality of life (Chow et al., 2011).

Consequently, wheelchair propulsion research has led to guidelines to minimize over-use injuries (Cooper R et al 1998, Boninger et al., 2002, Koontz et al., 2002, Richter et al., 2007). In general the guidelines suggest that individuals use propulsion patterns such as semi-circular and double loop that maximize contact angle. However these guidelines do not discuss other kinematic markers of movement such as jerk, which has been implicated in overuse injuries (Berret et al., 2008, Srinivasan et al., 2012, William et al., 2008, Mark (2012)).

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Jerk, the third derivative of position has been widely employed in clinical rehabilitation and human motor control research to quantify movement smoothness and evaluate the performance of upper limb tasks (Hogan et al., 1987, Flash., 1990, Chang et al., 2005, Caimmi et al., 2008). Occupational ergonomics research has revealed distinct differences in arm jerk between movements in individuals with and without shoulder pain (Cote et al., 2005). Consequently, the purpose of this investigation is to examine jerk in wheelchair propulsion as a function of recovery pattern and shoulder pain.

To appreciate this research it is important that the reader understands that a typical push-rim wheelchair propulsion has two phases, a push phase (hands in contact with push-rim) and a recovery phase (hands move freely to initiate next push).   Four general categories of recovery patterns widely reported in the literature are semi-circular (SC), single loop (SLOP), double loop (DLOP) and ARC (Shimada et al. 1998, Boninger et al., 2002, Richter et al., 2007, Raina et al., 2012, Slowik et al., 2015).   The magnitude of forces and moments experienced by the shoulder joint during recovery phase of wheelchair propulsion can be as high as that during the push phase (Mercer et al., 2006, Sosnoff et al., 2015). Given this association, it is logical to expect that shoulder pain will influence arm kinematics during the recovery phase of wheelchair propulsion. Indeed recent research shows that mWCUs with shoulder pain employed spatial adaptive strategies to wrist kinematics

during the recovery phase of wheelchair propulsion (Jayaraman et al., 2014).

This analysis examines jerk-based metrics extracted from the three dimensional kinematics of the upper arm joints during wheelchair propulsion. The main goals are (1) to introduce and benchmark a jerk-based framework for wheelchair propulsion and (2) to examine jerk in wheelchair propulsion as a function of recovery pattern and shoulder pain. To accomplish these goals, two recovery pattern types, SC and DLOP patterns were analyzed. A DLOP pattern is characterized by the hands lifting over the propulsion path and crossing the propulsion path to drop below the hand-rim forming a double loop, while a SC pattern is characterized by the hands dropping below the hand-rim during the recovery phase (Sanderson et al., 1985, Shimada et al., 1988, Boninger et al., 2002, Richter et al., 2007).

We postulate two hypotheses: (H1) that individuals using a SC recovery pattern will experience lower jerk magnitudes at their wrists than individuals using a DLOP recovery pattern; (H2) that individuals with shoulder pain will minimize peak jerk magnitude at their upper arm joints during the recovery phase kinematics in an effort to avoid pain. H1 rests on the logical rationale that the arm's movement trajectory during a SC pattern is simpler than a DLOP pattern. H2 is based on the observation that the neuromuscular system avoids large acceleration changes to avoid pain (Berret et al., 2008).

**Methods**

**Participant demographics**

Wheelchair propulsion data from 22 experienced adult mWCUs were analyzed. This data constitutes a subset of data from a larger study (n=27) examining wheelchair propulsion and shoulder pain (Sosnoff et al., 2015) . The total number of participants that employed a SLOP (n=4) or ARC (n=1) were few, hence only SC and DLOP patterns were analyzed.   Inclusion criteria for the larger investigation were: (1) between 18-65 years old and (2) use of a manual wheelchair as their primary means of mobility for 1+ year.   Table 1 lists the participant demographic information. The recovery pattern types were classified using the third metacarpophalangeal joint's sagittal plane motion (Shimada et al., 1988). A sample DLOP and SC pattern are shown in Fig.1 (a1-b1) respectively. Twelve individuals used a SC pattern while ten participants used a DLOP pattern. These propulsion patterns were self-selected and no specific instructions regarding pattern were provided.

**Kinematic and kinetic data collection**

The kinematic and kinetic data collection procedures have been detailed in Chapter 2. Since participants switched the propulsion pattern used for different speeds, only the kinematics at 1.1 m/s speed was

analyzed.

**Kinematic and kinetic data post-processing**

The motion data were post- processed and any missing intermediate marker data points were fit using a cubic interpolation. This post-processing was accomplished with Cortex 2.5 Motion Analysis software. The hand-rim kinetic data from the SMARTWheel was used to identify the push and recovery phases from each propulsion cycle. The push phase's start and end points were located where the moment applied to the hand-rim (Mz) was greater and   lower than 1 Nm, respectively, for at least 10 ms   (Richter et al., 2011).

The post- processed motion data were filtered using a fourth-order low-pass Butterworth filter with 6 Hz cut-off frequency (Bednarczyk et al., 1994) . The wrist motion data was approximated as the mid-point of the radial styloid (RS) and ulnar styloid (US) hand segment marker coordinates. The elbow motion data was approximated as the mid-point of olecronon and lateral epicondyl, while the acromion process was used to represent the shoulder kinematics. To test our hypotheses (H1 & H2) we analyzed three dimensional motion data from wrist, elbow and shoulder joints.

Based on recommendation from previous literature, jerk metrics in this analysis were computed from the Cartesian coordinate motion data (Flash

(1990), Hreljac (2000)). From the three dimensional motion data (i.e. X, Y and Z coordinate) of the joints, the instantaneous resultant velocity along the trajectory was approximated. To accomplish this, first the individual velocity components from X,Y and Z coordinate motion data were calculated. Then the resultant of these individual velocity components was computed to obtain the instantaneous resultant velocity (Winter (2009)). The acceleration and jerk were approximated by obtaining the successive first and second order time derivatives of the resultant velocity respectively (Winter 2009). The absolute magnitude of the jerk was computed and used for further analyses. For all time derivative approximations, a two point central difference scheme was used (Winter 2009). The data was filtered using a fourth-order low-pass Butterworth filter with 6 Hz cut-off frequency before approximating each time derivative   (Cooper et al., 2002, Winter (2009)).

To be consistent across individuals, 50 consecutively occurring recovery phases were extracted from each participant's data set. Each extracted absolute jerk curve was time normalized to 100 points using a shape preserving cubic spline. Two jerk measures were computed from these absolute jerk curves, namely a *jerk cost criteria* ($J_c$) and *peak jerk criteria* ($PJ_c$).

To compute $J_c$, first the area under the absolute jerk curve for each of the 50 extracted recovery cycles was computed. $J_c$ was computed as the

average value of the area under the absolute jerk curve. The scheme used for computing $J_c$ is shown in figure 3.3. The group-wise averaged $J_c$ between the SC and DLOP groups were statistically compared to validate the first hypothesis. For this comparison, data belonging to the dominant hand side was analyzed (right side: n=19(SC=10; DLOP=9); left side: n=3(SC =2, DLOP=1)).

To compute $PJ_c$, first the $P_{max}$ from each of cycle was extracted. $P_{max}$ was defined to be the peak magnitude of absolute jerk that occurred during the recovery trajectory. Two distinct peak jerk magnitude locations were observed, the first between the 0% to 30% (see peak points $P_1$ in figure 3.2(c) & $P_4$ in figure 3.2(d)) and the other between the 70% to 100% (see peak points $P_2$ in figure 3.2(c) & $P_5$ in figure 3.2(d)) intervals. $PJ_{c\,(0\%\ to\ 30\%)}$ and $PJ_{c\,(70\%\ to\ 100\%)}$ were computed as the average of the peak magnitude of the absolute jerk ($P_{max}$) (averaged over the 50 consecutive cycles). The scheme used for computing $PJ_c$ is shown in Figure 3.4. The group mean $PJ_c$'s from the two intervals were statistically compared between the groups with and without shoulder pain to validate the second hypothesis. The $PJ_c$ belonging to the side of the hand with the greatest shoulder pain was analyzed for the pain group (right side: n=9 (SC=5; DLOP=4); left side: n=1(SC =1)). The $PJ_c$ from the dominant hand side was used for the group without shoulder pain, (right side: n=10 (SC=5; DLOP=5); left side: n=2 (SC=1; DLOP=1)).

**Kinetic data processing**

In addition to the kinematic data, the within individual cycle-wise spatial-temporal propulsion variables were extracted. These included, the mean contact angle, mean push time, mean push speed and mean peak resultant force at the hand-rim, each averaged over the 50 cycles considered. A custom developed MATLAB program was used for all computations.

**Statistical analysis**

All statistical data analyses were conducted using SPSS (version 21, IBM, Inc.). All values are reported as Mean (SD) unless otherwise noted. The significance level was set to $P \leq 0.05$.

**Independent variables**

Participant demographics information (age, body weight, arm length and manual wheelchair propulsion experience) self-reported current level of shoulder pain (VAS scores) and WUSPI scores were treated as independent variables. A series of two tailed independent t-tests with propulsion pattern (SC or DLOP) and the shoulder pain (pain vs. no pain) as the between subject factors were conducted to check if statistically significant group differences existed in demographic variables. Two-tailed Mann–Whitney U

tests were used to identify if statistical significant difference in VAS and WUSPI scores existed between the pain groups.

## Dependent variables

Mean contact angle, mean peak resultant force at hand-rim during push, mean push speed, Jc's and $PJ_c$'s were treated as dependent variables. To test if statistically significant group differences existed in the dependent variables, a series of two-way analysis of variances with propulsion pattern (SC or DLOP) and the shoulder pain status (pain vs. no pain) as the between subject factors were conducted.

## Results

## Demographics

No statistically significant between group differences in demographics information as a function of recovery pattern type or shoulder pain status were observed, ($P$'s>0.05; Table 3.1). Per design, the group with shoulder pain reported higher pain than the no pain group (VAS: [$U$=11,P<0.05]; WUSPI:[$U$=11,P<0.05]). No statistically significant difference in shoulder pain was observed as a function of recovery pattern ($P$>0.05).

**Spatial-temporal propulsion variables at hand-rim**

The group-wise mean (SD) of spatial-temporal propulsion variables are reported in Table 3.2. No statistically significant between group differences in peak resultant force, push speed or contact angle were observed (*P's*>0.05) as a function of pattern type. Push time was significantly different between the SC and DLOP groups [F(1,18)=4.63,*P<0.05*,$\eta^2$=0.20] with the SC group having a greater push time. No statistical significant differences were observed in mean spatial-temporal propulsion variables as a function of shoulder pain (*P's*>0.05).

**Recovery kinematic and jerk metrics**

A representative plot of the resultant velocity, acceleration and jerk at the wrist for a DLOP and SC pattern are shown in figure 3.1 (a2-a4) and figure 3.1 (b2-b4), respectively. A time normalized (0% to 100%) absolute jerk curve computed for the wrist for SC and DLOP pattern are show in figure 3.2(c-d), respectively. Figure 3.3 (a-b) shows a sample area under the curve for SC and DLOP pattern types. The number of peak jerk points for a DLOP pattern appears higher than that for a SC pattern. The area under the curve for a DLOP pattern is larger than that of a SC pattern.

A sample recovery trajectory comparing peak jerk magnitudes ($P_{max}$) between the groups with and without shoulder pain for the SC and DLOP pattern types are shown in figure 3.4(a-b) respectively. It is clear from the sample data that irrespective of the pattern type, $P_{max}$ magnitude for the individual with shoulder pain is lower than that for the individual without shoulder pain.

## Jerk criteria ($J_c$)

A statistically significant main effect of recovery pattern (SC and DLOP) was observed for $J_c$ at the wrist, elbow and shoulder joint; $[F(1,18)=8.49, P<0.05, \eta^2=0.32]_{wrist}$; $[F(1,18)=4.3, P=0.05, \eta^2=0.19]_{elbow}$ ; $[F(1,18)=9.28, P<0.05, \eta^2=0.34]_{shoulder}$. The SC group experienced lower mean $J_c$'s than the DLOP group during the recovery phase (See figure 3.5(a-c)). No statistically significant between group differences in $J_c$ was observed as a function of shoulder pain ($P>0.05$).

## Peak jerk criteria ($PJ_c$)

A statistically significant main effect of shoulder pain was observed for $PJ_{c\ (0\%\ to\ 30\%)}$; $[F(1,18)=10.01, P<0.05, \eta^2=0.36]_{wrist}$ ; $[F(1,18)=7.8, P<0.05, \eta^2=0.30]_{elbow}$ and $[F(1,18)=8.16, P<0.05, \eta^2=0.31]_{shoulder}$. The shoulder pain group had lower $PJ_{c\ (0\%\ to\ 30\%)}$ magnitude at all the three joints

than the no pain group (See figure 3.6(a-c)). No statistically significant main effect of shoulder pain was observed for $PJ_{c(70\% \text{ to } 100\%)}$ ($P > 0.05$).

**Discussion**

In this investigation, the jerk characteristics of the upper limb during the recovery phase of manual wheelchair propulsion as a function of propulsion pattern and shoulder pain were examined. In agreement with our postulated hypotheses the SC recovery pattern experienced lower $J_c$ and individuals with shoulder pain had less $PJ_c$ regardless of propulsion style. Overall, our results suggest that, utilizing a jerk metric while analyzing manual wheelchair propulsion provides novel insights.

The mean spatial-temporal wheelchair propulsion parameters observed in this investigation were consistent with previous literature (Boninger et al., 1997, Shimada et al., 1998, Boninger et al., 2002, Collinger et al., 2008, Richter et al., 2011). This benchmarking was essential to suggest that the observations from our investigation are generalizable and qualitative comparison of our results with previous literature is acceptable.

The logical reason for the DLOP recovery pattern to incur greater $J_c$ is a result of the joints undergoing sharp directional turns, leading to frequent

switching between acceleration and deceleration during the recovery

trajectory. In contrast, when executing a SC pattern the arm undergoes less

directional change resulting in lower jerk. Additionally, the relatively complex

DLOP kinematics requires the upper extremity musculature to do additional

work to overcome the inertia and gravity. From a jerk minimization

perspective, it appears that the SC pattern appears to be superior.

The second novel observation from this investigation is that,

individuals with shoulder pain minimized $PJ_{c(0\% \text{ to } 30\%)}$ at all the three joints,

namely, the wrist, elbow and shoulder compared to the group without

shoulder pain. This observation is consistent with research observation

reported in occupational biomechanics. For instance, individuals with back

pain lift a box with less jerk than those without back pain (Slaboda et al.,

2005) and it was suggested that those with pain adopt a pain minimizing

strategy, characterized by lower jerk. Another investigation revealed that

shoulder pain influenced the kinematics of arm joints (Cote JN et al., 2005)

with those with shoulder pain having a kinematic movement pattern with

lower acceleration magnitudes than those without pain. In the context of our

analysis, it is maintained that individuals with shoulder pain adopt a

smoother arm motion pattern to reduce momentary discomfort at the

shoulder during wheelchair propulsion.

Although the examination of jerk is relatively novel within wheelchair biomechanics research, it is consistent with human motor control research that has implemented jerk based measures to evaluate upper limb movement. Evidence from motor control research suggests that kinematic analysis involving jerk is a viable approach to quantify movement coordination during rehabilitation (Ramos et al., 1997, Teulings et al., 1997, Cozens et al., 2003, Caimmi et al., 2008, Chang et al., 2005). Our results and suggest that integrating jerk metrics with existing analysis procedures can yield an enhanced understanding of wheelchair propulsion mechanics.

For instance, wheelchair propulsion analyses have focused on studying the effect of the different recovery patterns on overall mechanical efficiency (Boninger et al., 2002, de Groot et al., 2004). There are divergent suggestions regarding the overall mechanical efficiency of SC and DLOP patterns (Shimada et al., 1998, Boninger et al., 2002, de Groot et al., 2004), with both   being suggested as the more efficient propulsion technique. However, these studies did not differentiate the influence of different kinematic effects of the recovery phase jerk cost to their overall mechanical efficiency estimates. Perhaps integrating metrics such as jerk with existing wheelchair analysis procedures will yield novel information concerning mechanical efficiency.

**Limitations**

Despite being novel there are limitations that need to be acknowledged. Our sample size was small to investigate the influence of specific injury demographics on the jerk characteristics. However, the diversity of injury could also be viewed as a strength of this study. The results were significant despite having a sample with diverse injury demographics. The sample demographics limited our analysis to SC and DLOP patterns. It is not clear if similar movement characteristics could be identified in other recovery pattern types (ARC and SLOP; figure S1, S2 and S3). However for sake of completeness, the jerk characteristics for a sample SLOP and ARC patterns are provided in the chapter addendum section (Chapter addendum: Figures 3.7–3.9). Information on wrist pain was not collected. It is reasonable to expect that wrist pain is unlikely to influence kinematics during the recovery phase since the wrist experiences minimal forces/moments. A last limitation involves the laboratory based roller dynamometer setup utilized which provides a propulsion environment that does not exactly emulate real life propulsion.

**Conclusions**

This research implemented a novel approach integrating metrics and inferences from human movement ergonomics and motor control to understand kinematics of manual wheelchair users with shoulder pain. The analysis indicates that, adopting jerk based quantification of wheelchair

propulsion kinematics is worthwhile and yields insightful inferences. Overall the recovery phase kinematics of individuals using a SC recovery pattern placed lower jerk magnitudes than those using a DLOP and (2) mWCUs with shoulder pain had lower peak jerk magnitude during the recovery phase of wheelchair propulsion. In the future it may be beneficial to incorporate jerk based metric into rehabilitation practice.
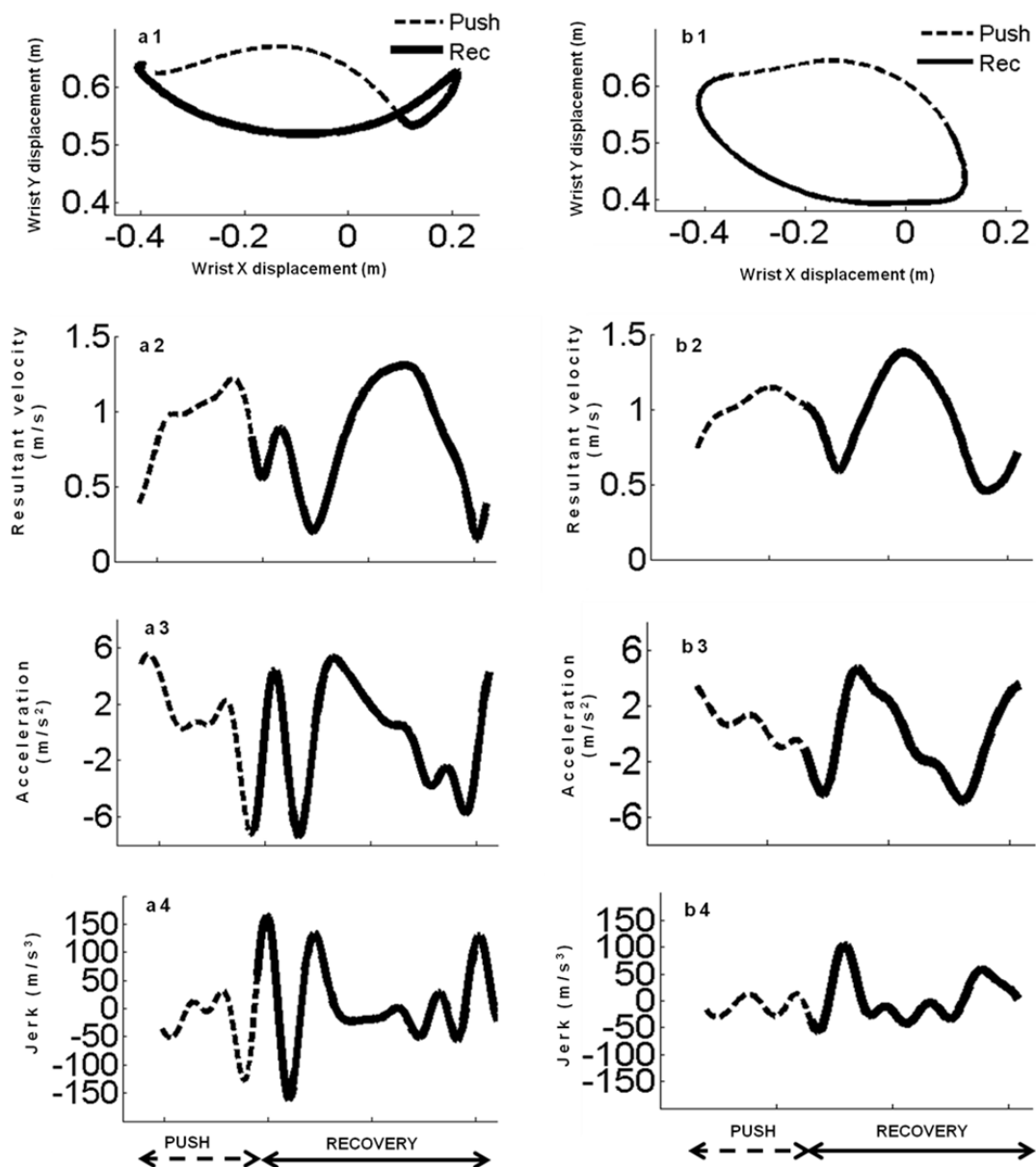
Figure 3.1 The instantaneous <u>resultant velocity,</u> acceleration and jerk at wrist from a sample DLOP (a1-a4) and SC (b1-b4) recovery pattern type. The solid lines in all the plots belong to the recovery phase and the dotted line to the push phase. The resultant velocity plot for wrist during the recovery for a DLOP (solid lines - a2) has two asymmetric velocity profile one for each loop as opposed to a SC pattern (solid line - b2).The rate of change of acceleration and deceleration for a DLOP (solid line – a3) is greater than a SC (solid line – b3) pattern. The jerk magnitude for the DLOP (solid line – a4) is greater than a SC (solid line – b4) pattern.
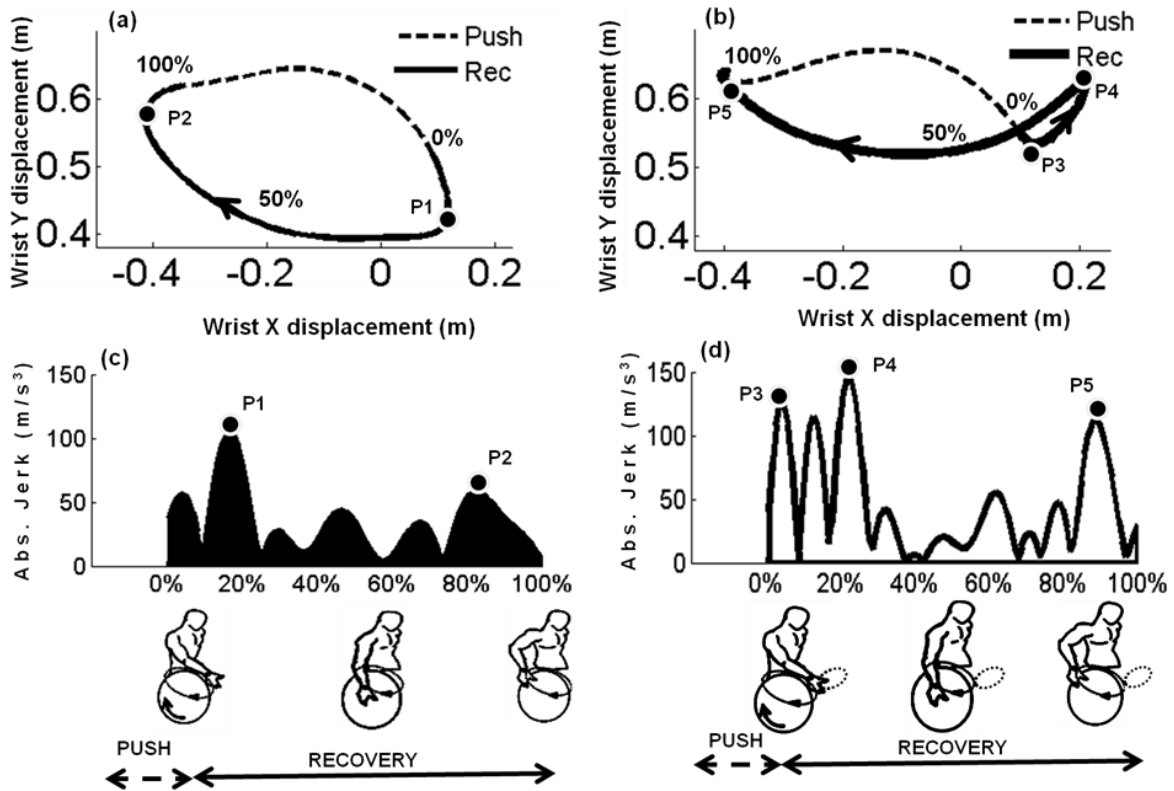
Figure 3.2 A sample time normalized (0% - 100% points) recovery trajectory absolute jerk curve at wrist for SC and DLOP patterns. (a) & (b) Push (dotted lines) and recovery (solid lines) phases for a SC and DLOP patterns respectively . The 0% and 100% points represent the start and end of the recovery phase. There are 100 data points between the 0% to 100% points; (c) & (d): Peak jerk magnitudes at wrist during SC and DLOP recovery patterns respectively. Peak jerk magnitude at wrist occurred between 0% to 30% (P1,P3,P4) and 70% to 100% (P2,P5) intervals along the recovery trajectory. These peak jerk points typically were seen to occur at those intervals along the recovery phase that required steep acceleration and deceleration rate of change.
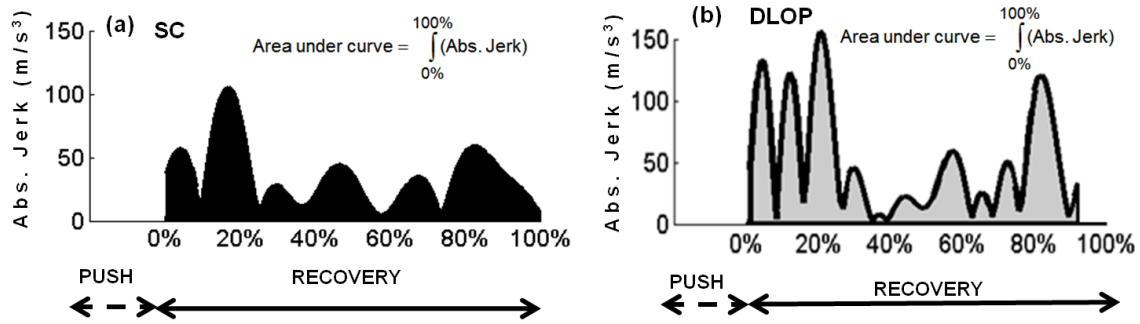
$$J_c = \left(\frac{1}{50}\right)\sum_{C_m-25}^{C_m+24}\left(\text{Area under curve}_{C_m}\right)$$

where $C_m$ is the temporally located mid-cycle of the trial.

Figure 3.3 The jerk cost criteria ($J_c$): (a) & (b) are sample SC and DLOP pattern recovery phase absolute jerk curves depicting the area under the curve as shaded regions respectively. The scheme used for computing $J_c$ is shown below the figures. $C_m$ : the propulsion cycle located closest to the mid of the trial (i.e. 90 seconds from the start of trial).

$$P_{max} = \mathbf{max}(Abs.Jerk)_{0\%-30\%}$$

$$PJ_c = \left(\frac{1}{50}\right) \sum_{C_m-25}^{C_m+24} \left(P_{max\,C_m}\right)$$
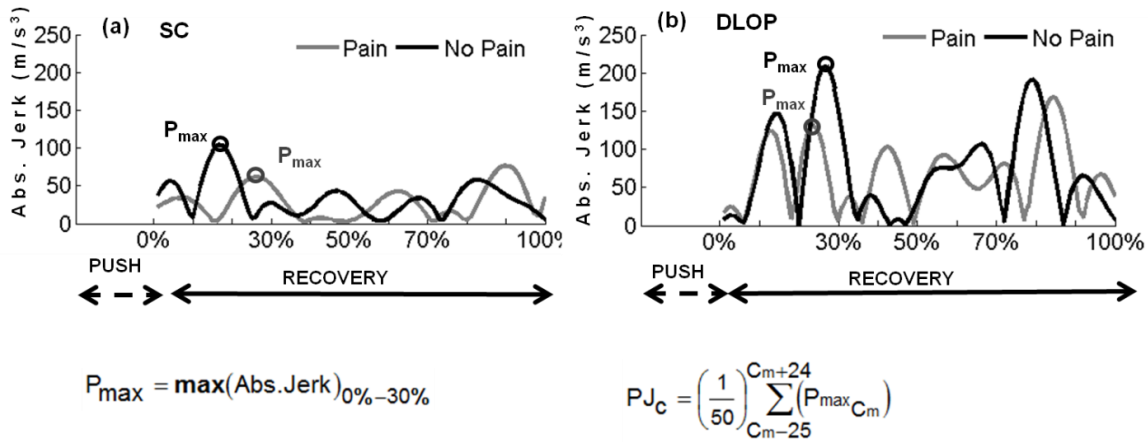
where $C_m$ is the temporally located mid-cycle of the trial.

Figure 3.4 The peak jerk criteria ($PJ_c$). (a) a sample plot from the SC group comparing $P_{max(0\%-30\%)}$ values between two individuals, with and without shoulder pain. (b) a sample plot from the DLOP group comparing $P_{max(0\%-30\%)}$ values between two individuals, with and without shoulder pain. The scheme used for computing $P_{max}$ and $PJ_c$ is shown below the figures. $C_m$ is the propulsion cycle located closest to the midpoint of the trial (i.e. 90 seconds from the start of trial).
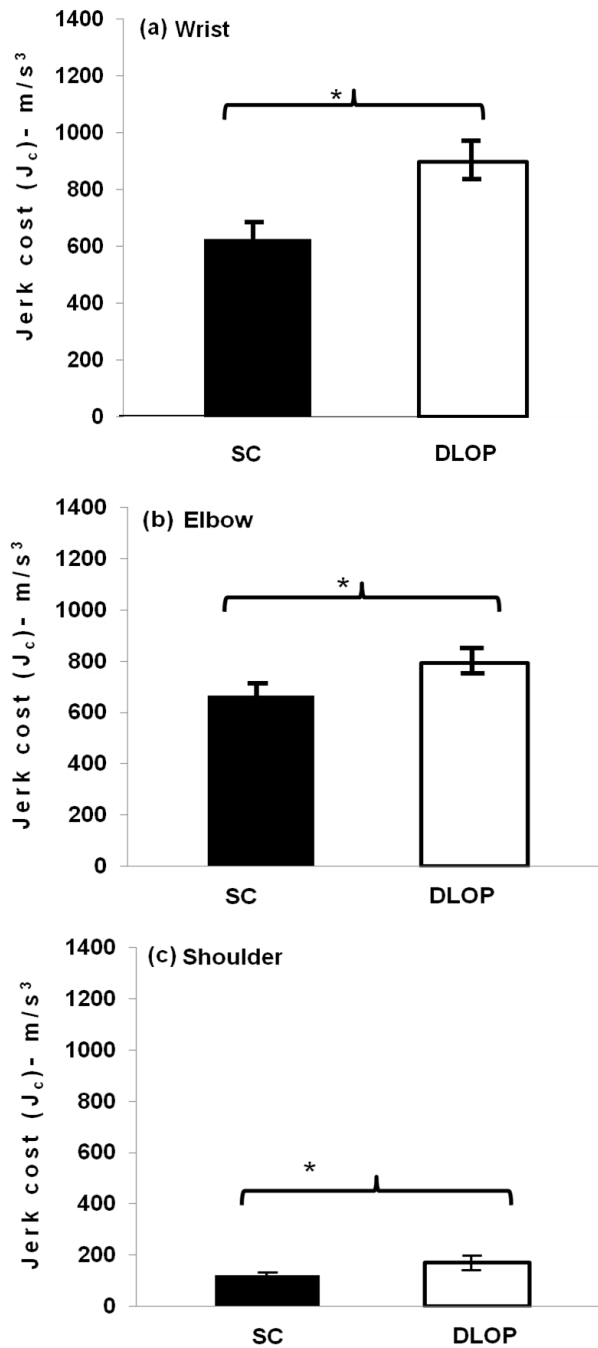
Figure 3.5 Group mean comparison for $J_c$ at the arm joints between SC and DLOP groups. (a) Group mean $J_c$ at wrist joint for SC group is lower than that of the DLOP group; (b) Group mean $J_c$ at elbow joint for SC group is lower than that of the DLOP group; (c) Group mean $J_c$ at shoulder joint for SC group is lower than that of the DLOP group (*$P \leq 0.05$).
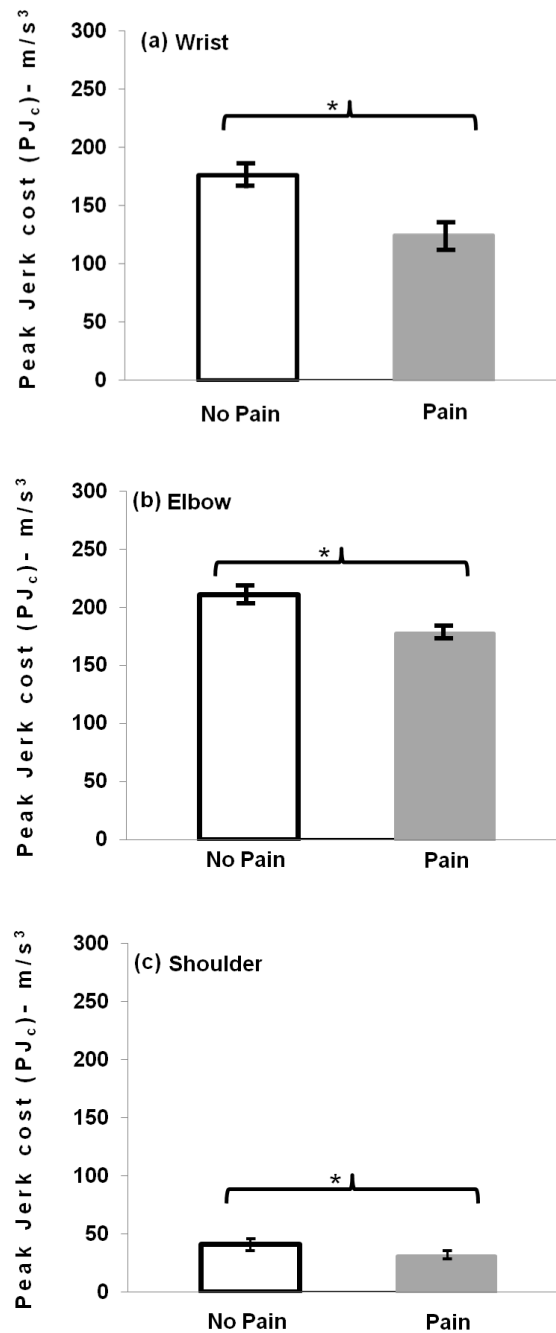
Figure 3.6 Group mean comparison for $PJ_c$ at arm joints between groups with and without shoulder pain. (a) Group mean $PJ_c$ at wrist joint for shoulder pain group is lower than that of the group without shoulder pain; (b) Group mean $PJ_c$ at elbow joint for shoulder pain group is lower than that of the group without shoulder pain; (c) Group mean $PJ_c$ at shoulder joint for shoulder pain group is lower than that of the group without shoulder pain. (*$P \leq 0.05$)

**List of tables**

**Table 3.1** Demographic information

| Characteristics | Recovery pattern type SC (n=12) Mean(SD) | DLOP (n=10) Mean(SD) | Shoulder pain status With pain (n=10) Mean(SD) | No pain(n=12) Mean(SD) |
|---|---|---|---|---|
| Age(years) | 25.5(10.68) | 21.5(4.1) | 25.8(11.11) | 22.00(5.31) |
| Body weight (N) | 608.73(235.22) | 627.24(180.68) | 677.68(216.58) | 590.92(194.47) |
| Arm length (mm) | 644.5 (54.51) | 630.5 (33.76) | 647(47.6) | 630.9(45.8) |
| Current shoulder pain (VAS in cm) | 2.11(3.22) | 1.83(2.64) | 4.16(3.00) * | 0.02(0.09) * |
| WUSPI scores | 14.58(22.77) | 6.00(6.29) | 20.70(22.28)* | 2.33(3.75)* |
| Experience using manual wheelchair (Yrs) | 16.83(5.72) | 13.05(4.71) | 16.45(6.18) | 14.09(4.86) |
| Injury details | T6 and below (n=7); Sacral agenesis(n=1), Spinal cyst T6(n=1), Spina bifidia(n=2), C7(n=1). | T6 and below (n=5); Amputee(n=2), Spina bifidia (n=3). | T6 and below (n=5); Sacral agenesis(n=1), Spinal cyst T6(n=1), Spina bifidia(n=2), Amputee (n=1). | T6 and below (n=7); C7(n=1), Spina bifidia(n=3), Amputee (n=1). |

**Table 3.2** Mean (SD) of propulsion variables

| Spatial-temporal propulsion variables | Recovery pattern type | | Shoulder pain status | |
|---|---|---|---|---|
| | SC (n=12) Mean(SD) | DLOP (n=10) Mean(SD) | With pain (n=10) Mean(SD) | No pain(n=12) Mean(SD) |
| Mean peak resultant force at hand-rim (N) | 59.07 (18.38) | 66.36 (20.81) | 64.56(17.99) | 60.57(21.11) |
| Mean push time  (sec) | 0.54 (0.06)* | 0.47 (0.07)* | 0.54(0.08) | 0.48(0.06) |
| Mean contact angle (deg) | 112.37 (12.21) | 101.45 (13.30) | 111.69(15.80) | 103.84(11.26) |
| Mean push speed (m/sec) | 1.12 (0.04) | 1.17 (0.08) | 1.12(0.05) | 1.16(0.07) |

*P<0.05

**Chapter addendum**
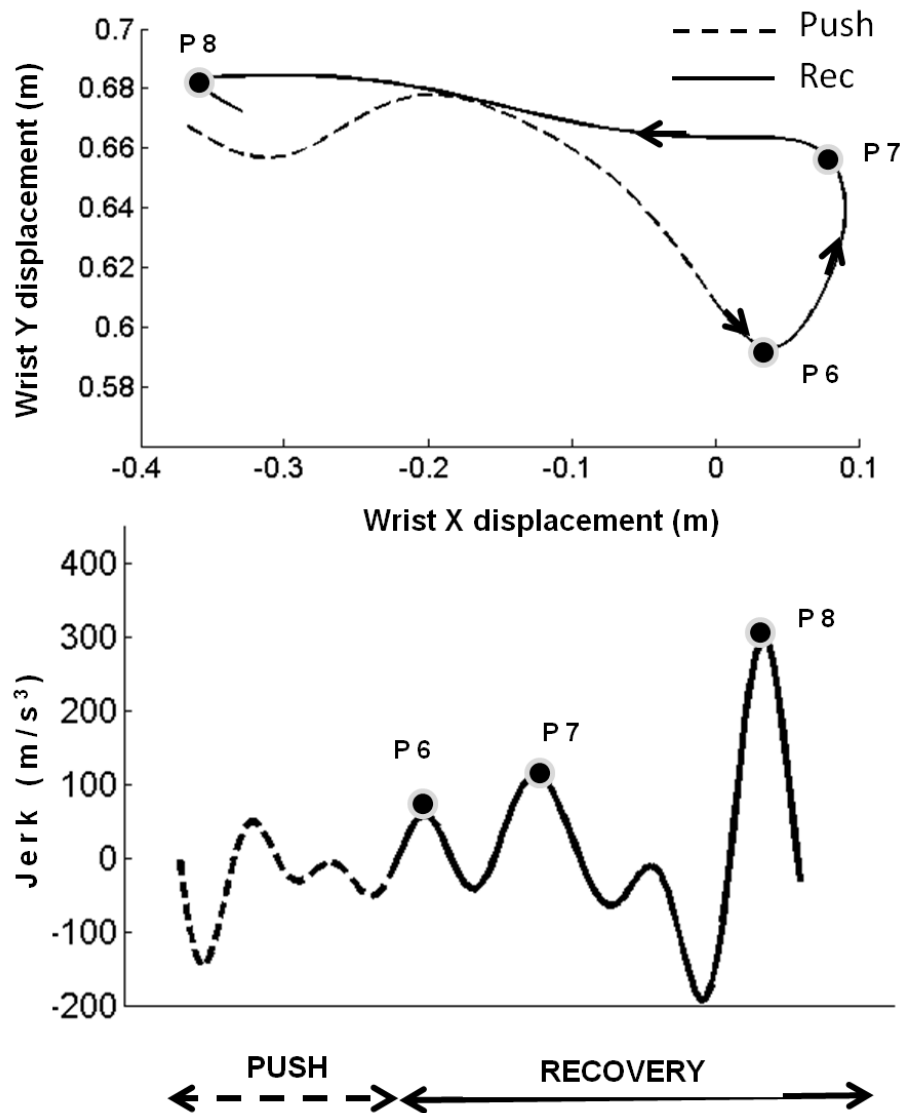


Figure 3.7 Wrist kinematic characteristics of a sample SLOP pattern: the hands rise above the hand rim during the recovery phase for a SLOP pattern; the dotted line lines in all the plots belong to the push phase and the solid lines to the recovery phase. Points P6,P7 and P8denote the jerk magnitude at the turns points made by the wrist during the SLOP recovery pattern.
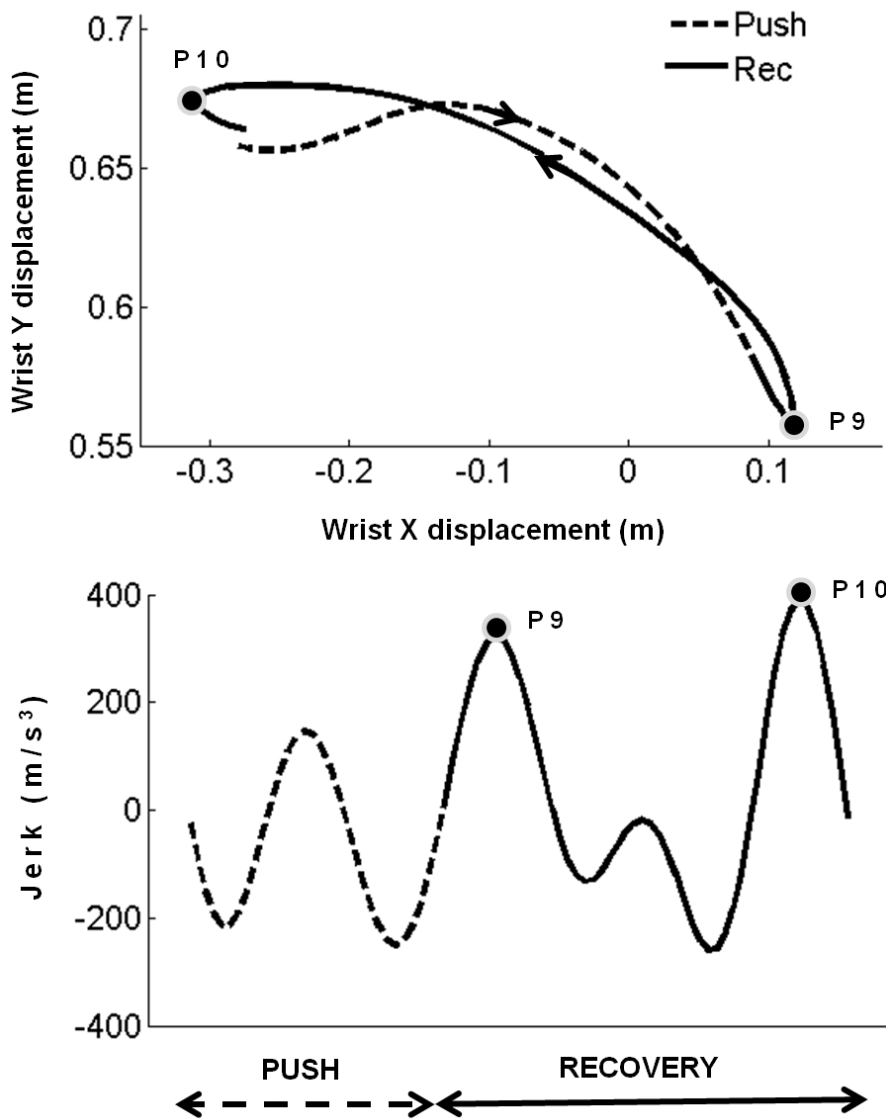
Figure 3.8 Wrist kinematic characteristics of a sample ARC pattern: the hands follows the hand rim closely during the recovery phase for a ARC pattern forming a pumping action; the dotted line lines in all the plots belong to the push phase and the solid lines to the recovery phase. Points P 9and P 10 denote the jerk magnitude at the turns points made by the wrist during the ARC recovery pattern.
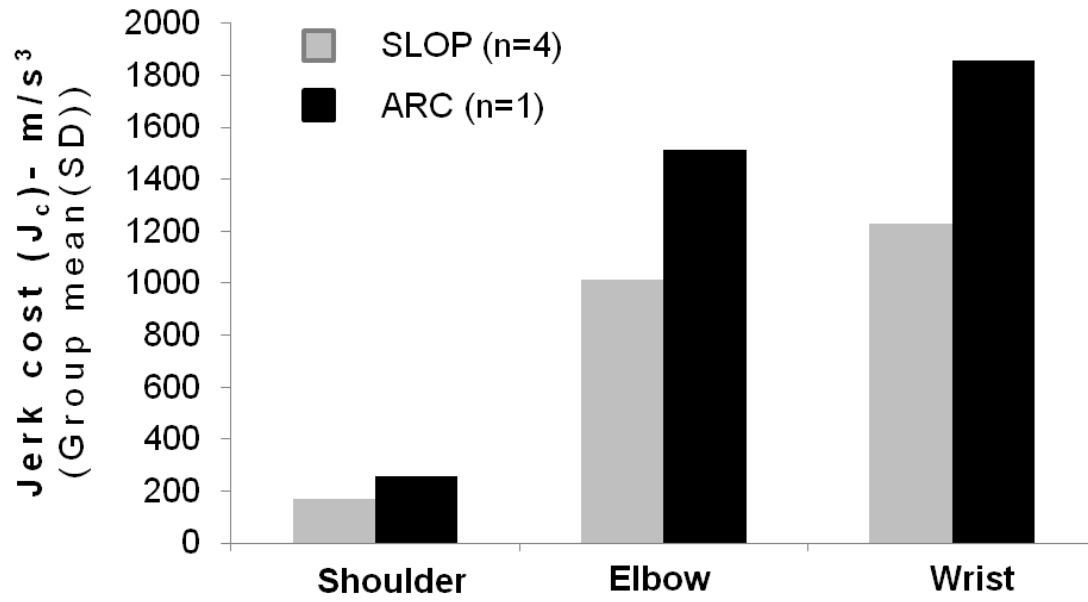
Figure 3.9 Group mean plots for $J_c$ at the upper arm joints for SLOP and ARC patterns.

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# Chapter 4

# Shoulder pain and trunk kinematics in manual wheelchair propulsion

**Chapter abstract:** Trunk kinematics during wheelchair propulsion is known to influence propulsion biomechanics and also bears implication for shoulder injury. The main aim of this investigation was to study the trunk kinematic differences between manual wheelchair users with and without shoulder pain. Data from 25 experienced adult mWCUs was analyzed for this study (pain/without-pain:13/12)). Participants propelled their own wheelchairs fitted with SMARTWheels on a roller dynamometer at 1.1 m/s for 3 minutes. Kinematic and kinetic data of the upper limbs were recorded. The net drift in the trunk position (during steady state propulsion) from the initial reference position (rest) along the sagittal plane (X direction) was computed. Two-way ANOVA was conducted to investigate if statistically significant differences in trunk drift existed as a function of shoulder pain( $p < 0.05$). Overall our results show that the trunk drift along the sagittal plane for manual wheelchair users with shoulder pain was greater than those without pain ($p < 0.05$; $\eta^2 = 0.30$). Incorporating trunk kinematics based measures may provide additional knowledge to understand the adaptive strategies employed by mWCUs with shoulder pain. To our knowledge this is the first work documenting trunk kinematic differences between in mWCUs with and without shoulder pain.

**Introduction**

Approximately 70% of MWCUs report incidence of shoulder pain within the first 12 months of wheelchair use (Finley et al., 2004). Shoulder pain in MWCUs is reported to have significant negative impact on quality of life (Chow et al., 2011). Overall, any loss of upper limb function due to pain adversely impacts the independence and mobility of mWCUs. It has been speculated that a decrease in independence and mobility leads to greater health care costs and an increased risk for secondary morbidity (osteoporosis, obesity, cardiovascular disease etc) (Dyson-Hudson et al., 2004, Gellman et al.,1988., Geronda et al.2004, Gutierrez et al., 2007, Finley et al., 2004). Consequently, over the past two decades many research investigations have focused on a plethora of biometric markers in an attempt to understand the relationship between shoulder injury and manual wheelchair propulsion. (Boninger et al., 2002, Collinger et al., 2008, Rankin et al., 2011, Rankin et al., 2012, Richter et al., 2007, 2011, Koontz et al., 20009, Koontz et al., 2012, de Groot et al., 2003, de Groot et al., 2004, Kwarciak et al., 2012, Mercer et al., 2006).

Evidence from this research highlight that the movement of the trunk during wheelchair propulsion is related to shoulder injury in MWCUs. Guidelines from these studies maintain that propelling a manual wheelchair with a 'trunk flexed forward/ anterior tilt' position exposes mWCUs to secondary injury risk (Rodgers et al., 2000, Rodgers et al., 2001, Rankin et

al., 2011, Gagnon et al., 2015, Gagnon et al., 2009, Sanderson DJ et al., 1985, Vanlandewijck YC et al., 1989, Chow et al., 2009, Yang et al., 2006, Rice et al., 2004).

Despite this large body of research, presently there is very limited information regarding trunk kinematics differences between groups propelling a manual wheelchair with and without shoulder pain. The is due to the reason that most of these studies on trunk kinematics in wheelchair propulsion literature studied user populations with healthy shoulder and/or a task constraint that may simulate a shoulder pain/exhaustion event ( i.e. ramp, loaded propulsion). There is a need to extend this body of research to study the trunk kinematics of experienced mWCUs with shoulder pain. Our investigation seeks to directly address this void in the literature by studying the trunk kinematics differences between experienced manual wheelchair users with and without shoulder pain.

Consequently, this cross-sectional study analyzed the trunk kinematics during manual wheelchair propulsion in a group of experienced adult MWCUs with and without shoulder pain. We hypothesized that the trunk kinematics of individuals propelling a manual wheelchair with shoulder pain will significantly differ from the trunk kinematics of the group without shoulder pain. This being the first study to record trunk kinematic differences between users with and without shoulder pain, the outcome from this analysis   will

potentially improve the knowledgebase and complement existing practices, leading to a better understanding of the role played by the trunk in the context of propulsion biomechanics and shoulder pain.

## Methods

### Protocol

Wheelchair propulsion kinetic and kinematic data from 25 individuals was analyzed in this study. Other aspects of this dataset have already been published (Sosnoff et al., 2015). Inclusion criteria for this secondary analysis were: (1) between 18-65 years old and (2) use of a manual wheelchair as their primary means of mobility for 1+ year and (3) Spinal related disability. Table 1 lists the participant demographic details. Reasons for wheelchair use included traumatic spinal cord injury (n=14), spinal cyst (n=1), spina bifidia(n=9), sacral agenisis (n=1) .

### Kinematic and kinetic data collection

The kinematic and kinetic data collection procedures have been detailed in Chapter 2. Since participants switched the propulsion pattern used for different speeds studying the kinematics at a specific speed was most appropriate. Consequently, only the data belonging to trials conducted with a speed of 1.1 m/s was used for this analysis.

Before starting the trial, the participants were instructed to hold an upright sitting position (i.e. trunk held at 90 deg relative to the horizontal plane), for 5 seconds. The sternum motion marker position recorded for this upright sitting position was treated as the initial reference position for the trunk. Participants then pushed the wheelchair at a target speed of 1.1 m/s for 3 minutes. A speedometer was used to provide real-time visual feedback while kinetic data were collected bilaterally at 100Hz. Kinematic data were collected using a 10-camera motion capture system (Cortex 2.5, Motion Analysis Co.; Santa Rosa, CA, USA) at a sampling rate of 100Hz. Hand-rim force and moment data were collected by the SMARTWheel.

**Kinematic data post-processing**

The post-processing was carried out using Cortex 2.5 Motion Analysis software. During this phase any missing motion-marker data points were fitted using a cubic interpolation. Then the motion data was filtered using a 4<sup>th</sup> order low-pass Butterworth filter with a 6 Hz cut-off frequency (Morrow et al., 2011). The sternal notch motion data was used to characterize the trunk kinematics. The trunk kinematic metric derived from the sternum motion data in this analysis were computed with the wheel center as reference. Since our main variable of interest was the sagittal plane trunk excursion in X-direction (Figure 2.2), this analysis was based only on the X coordinate position data of the sternal notch.

The X coordinate position data of the sternal notch was extracted from each participant. Data beginning from the 6th propulsion cycle to the end of the trial was considered as steady state portion of a trial (Koontz et al., 2009, Jayaraman et al., 2014). The average value of the total sternal notch excursion (mm) in the X direction was computed. The absolute of the difference between the average value of the total sternal notch excursion (mm) and the initial reference position of the sternal notch was defined as $T_{Drift}$. Thus, $T_{Drift}$ is a metric that captures the net drift in the trunk position over the entire trial period with respect to the initial reference position of the trunk, along the sagittal plane X direction.

The average trunk flexion angle during steady state propulsion was also computed. The computing procedure for 'average trunk flexion angle' was adopted from previous literature (Rodgers et al., 2001).

**Kinetic spatial-temporal data extraction**

The kinetic data from the SMARTWheel were used to identify the start and end of each propulsion cycle. The start and end points of each cycle were identified as points where the moment applied to the hand rim (Mz) was greater or lower than 1 Nm, respectively, for at least 10 ms (Jayaraman et al., 2014). All mean spatial-temporal wheelchair propulsion variables were extracted from the SMARTWheel hand-rim kinetic data.

A custom developed MATLAB code was used for the data post-

processing.

## Statistical Analyses

All statistical data analyses were conducted using SPSS (version 21, IBM, Inc.). Descriptive statistics results are reported as mean(SD). For all statistical tests, the between group factor was shoulder pain status (i.e. pain vs. no pain). The significance levels for the statistical tests were set to P≤0.05.

## Independent variables

A series of two-tailed independent t-tests were conducted to verify if statistically significant differences existed in the independent variables between the groups. The independent variables included age, years of wheelchair propulsion experience, body weight, number of pushes and wheelchair axle position with respect to shoulder (XPOS and YPOS ). Two-tailed Mann–Whitney U tests were used to identify if statistically significant differences in VAS and WUSPI scores existed between the pain groups. The difference in gender composition between the pain/no pain groups was evaluated using a $X^2$ test.

## Dependent variables

The mean values of spatial-temporal variables, namely, contact angle, push time, peak resultant push force at hand-rim, speed, trunk flexion and

$T_{Drift}$ were treated as dependent variables. A series of two-way ANOVA's were conducted to verify if statistically significant between-group differences existed in the mean spatiotemporal variables as a function of shoulder pain.

## Results

### Demographics

No significant between-group differences in independent variables, namely, age, years of wheelchair propulsion experience, body weight, and wheelchair axle positions, were observed as a function of shoulder pain ($P$'s>0.05). The group with shoulder pain reported higher pain scores than the group without shoulder pain (VAS:[$U$=11, P<0.05]; WUSPI:[$U$=11, P<0.05]). Descriptive statistics of all the demographic variables are furnished in Table 4.1.

### Mean spatial-temporal propulsion variables

No significant differences were observed in mean spatiotemporal propulsion variables as a function of shoulder pain ($P$'s>0.05). The group-wise mean (SD) of spatiotemporal propulsion variables are reported in Table 4.2. A significant main effect of shoulder pain was observed for $T_{Drift}$; [$F(1,23)$=9.6, $P$<0.05,$\eta^2$=0.30]. The group with shoulder pain had larger $T_{Drift}$ (mm) than the group without shoulder pain (see Fig 4.3) [with shoulder pain: 46.5(25.1) mm; without shoulder pain: 22.3(10.5) mm].

**Discussion**

The relationship between shoulder pain and trunk kinematics during steady state manual wheelchair propulsion in a group of experienced adult MWCUs with and without shoulder pain was investigated. Overall in agreement with the postulated hypothesis the participants with shoulder pain had significantly larger $T_{Drift}$ (mm) (i.e. larger trunk movement in the sagittal X direction) than the group without shoulder pain. This observation highlights that trunk kinematic metrics can differentiate between mWCUs with and without shoulder pain.

A logical explanation for the pain group to exhibit such differences in trunk kinematics could be that individuals with shoulder pain adopted a movement trunk strategy that minimized shoulder discomfort during the demanding propulsion task. This view is consistent with recommendations from occupation ergonomics literature (Madeleine et al., 2008, Lomond et al., 2010, Lomond et al., 2011). Individuals with neck/shoulder pain performing rhythmic repetitive occupational tasks adopted a spatial strategy manipulating the movement of their trunk to minimize discomfort (Madeleine et al., 2008, Lomond et al., 2010, Lomond et al., 2011).

The results holds significance because, in general a propulsion style with the trunk flexed forward has been reported to expose the arm joints of mWCUs to risk of injury (Rodgers et al., 2001, Raina et al., 2012). Moreover,

from a pure human movement ergonomics perspective, when the trunk is flexed forward, the arm joints grab the hand-rim with greater internal rotation at the initiation of the start of the push phase. Application of load to arm joints with such orientation is a well known to increase the risk of injury (Bridger RS 2009). Based on these observations, it is maintained that analyzing trunk kinematics in the context of shoulder pain in mWCUs is beneficial.

Previous wheelchair propulsion research included only a few propulsion cycles (~10-20) (Raina et al., 2012) to study trunk kinematics while our analysis uses the metrics computed over the entire trial (~135 cycles). The $T_{Drift}$ metric in this work focuses on the net trunk excursion over the entire trial, rather than separating them as push/recovery portions. Another highlight that differentiates this work is that, the kinematic data was collected on individuals with shoulder pain. This is the first work documenting trunk kinematic differences between mWCUs with and without shoulder pain.

The mean values of the wheelchair propulsion variables obtained from our sample qualitatively compare with previous research (Collinger et al., 2008, Boninger et al., 2002). Similar to previous research there was no mean difference between group for the kinetic parameters (Collinger et al., 2008). The mean values for trunk flexion during push phase from our

sample matches with the values reported in wheelchair literature (Rodgers et al., 2001, Gagnon et al., 2015). Such benchmarking of the qualitative behavior of our data with previous research is essential to extend complementary inferences and findings.

Wheelchair configuration is known to influences the trunk kinematics during propulsion. No significant differences existed between groups for the wheelchair configuration variables (XPOS, YPOS). Based on this we maintain that the effect of wheelchair configuration differences between the pain groups were negligible.

Trunk control data for the different spinal injury level was not collected from the participants, which is a major limitation. However, the demographic spread in spinal injury levels between the groups with and without pain in our sample were very similar (Table 4.1). These exploratory results must be replicated with future research efforts using a sample with more homogenous injury demographics.

**Limitations**

Despite producing new results there are some limitations to this study. The study is cross sectional and cause/effect relationships cannot be inferred from these results. Trunk control data for the different spinal injury level was not collected from the participants. The pain scores were self-reported and

no radiographic/ultrasonic information regarding shoulder pain was collected. The sample size was small. Data was collected in a laboratory environment and it is not clear if the results are replicable in real life environment.

**Conclusions**

Trunk kinematics differences between individuals with and without shoulder pain was studied. The net drift in trunk X coordinate position from the initial references position was calculated over the entire trial duration. Our observation highlights that trunk kinematic metrics can differentiate between mWCUs with and without shoulder pain. Individuals with shoulder pain had larger net drift in trunk position than those without shoulder pain ($p<0.05$). Incorporating trunk kinematics based measures, may provide additional knowledge to understand the adaptive strategies employed by mWCUs with shoulder pain. To our knowledge this is the first work documenting trunk kinematic differences between mWCUs with and without shoulder pain.
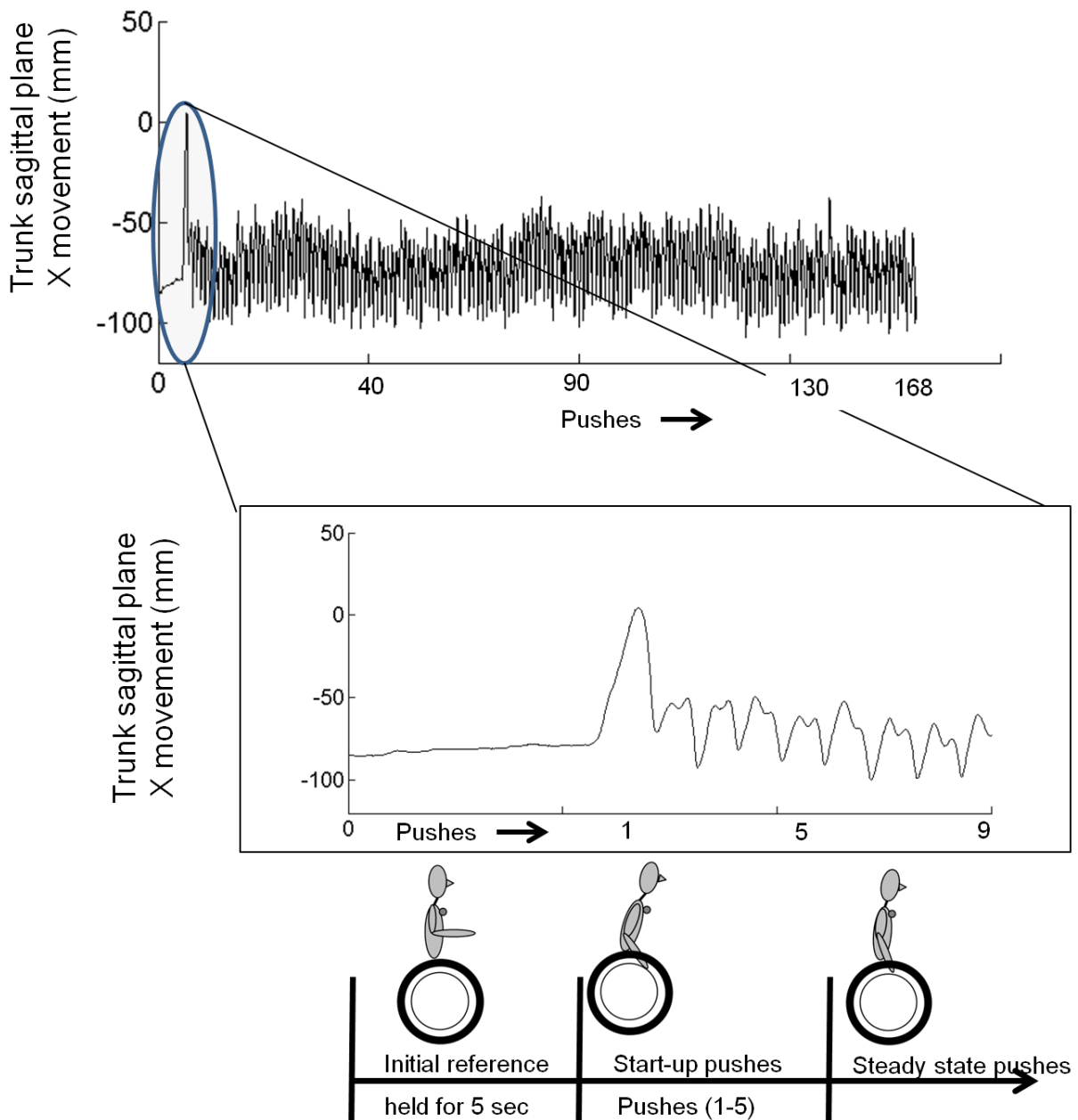
Figure 4.1 A sample trunk Sagittal plane motion (X kinematics) data. The initial reference position, the start-up push phases and the steady state portions are shown in the magnified view.
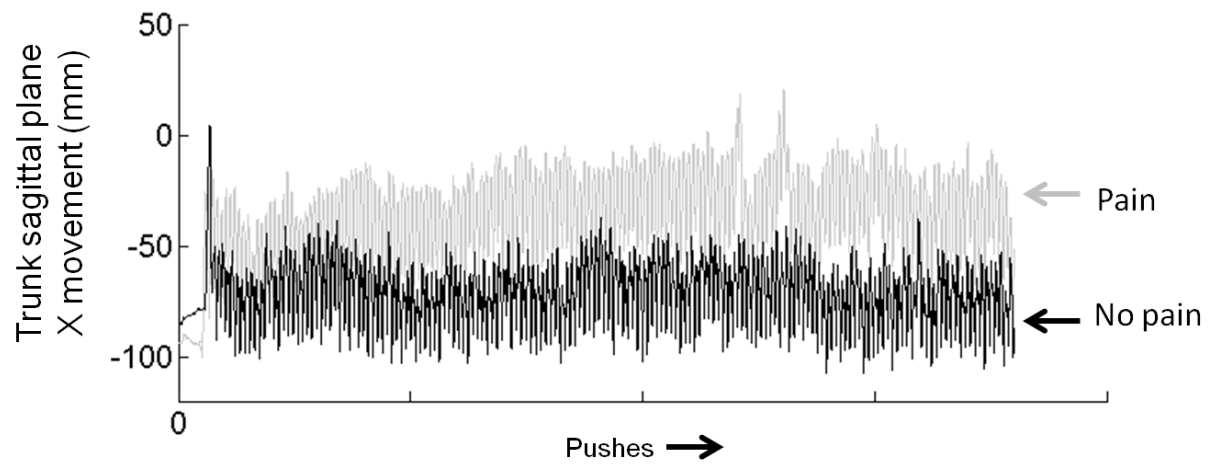
Figure 4.2 Sample trunk (X coordinate) kinematics for participants with and without shoulder pain. The data is for the X coordinate motion of the trunk for a 3 minute trial. It can be obsrerved that for th individual with pain, the trunk position drift is larger than the individual without pain.
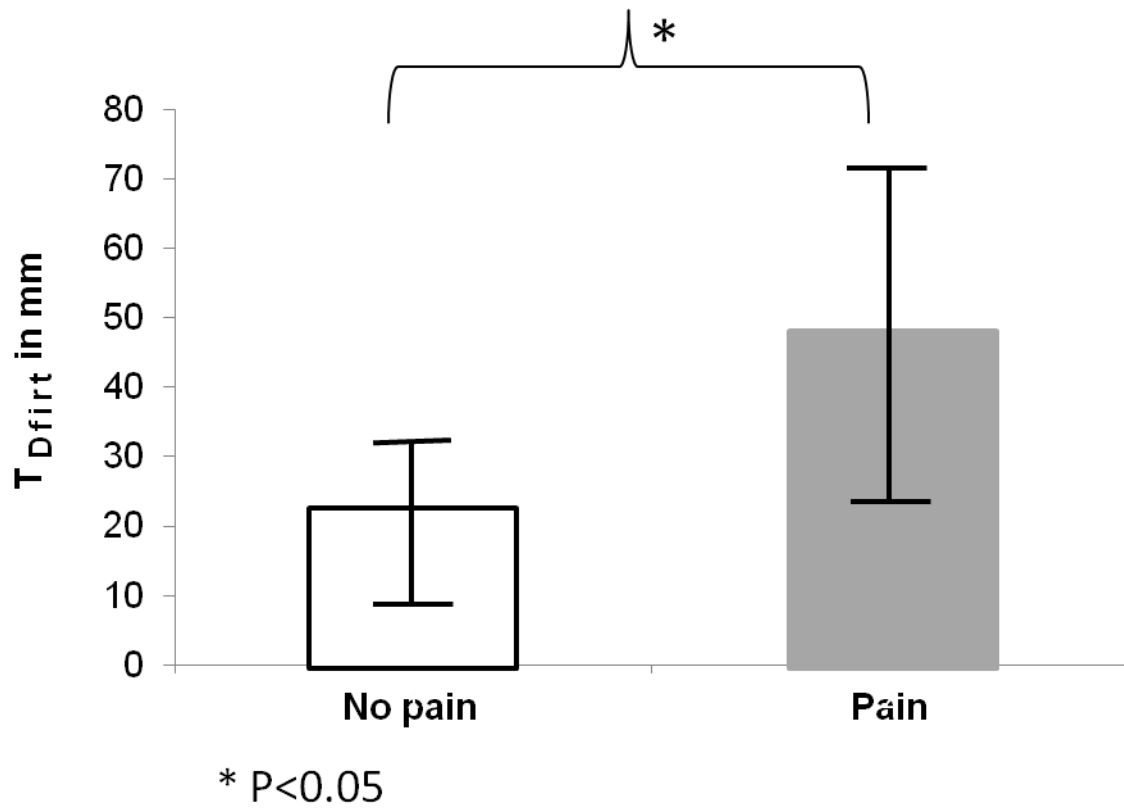
Figure 4.3 Group mean T $_{Drift}$ between groups with and without shoulder pain. The group with shoulder pain had higher net T $_{Drift}$

# List of tables

**Table 4.1** Demographic variables

| Demographic variables | No Pain Mean(SD) | Pain Mean(SD) |
|---|---|---|
| Age (Yrs)† | 20.1(2.0) | 28.3(12.5) |
| Body weight(Kg) | 55.5(15.68) | 75.63(24.71) |
| Wheelchair experience (yrs) | 13.7(4.3) | 15.6(11.0) |
| XPOS (mm) | 3.1(53.8) | 17.4(27.6) |
| YPOS (mm) | 660.1(60.8) | 695.8(57.4) |
| VAS† | 0.1(0.4)* | 4.5(2.5)* |
| WUSPI† | 2.7(4.9)* | 23.9(20.3)* |
| Camber (deg) | 3(1.5) | 2.0(1.9) |
| Gender (F / M) | 6 / 6 | 6 / 7 |
| Injury levels | SB (n=5); T12 complete (n=1); T6 (n=2); T9 incomplete (n=1); Spinal AVM T6-T9 (n=1); L4 (n=1); Arthrogryposis (n=1) | SB (n=4); T12 complete (n=1); T6 (n=2); T8 incomplete (n=2); Birth defect T11 to L2 (n=1); L3 incomplete (n=1);L1 (n=1); Sacral Agenisis (n=1) |

† Non parametric test

* P<0.05

**Table 4.2** Mean wheelchair propulsion variables

| Mean propulsion variables | No Pain Mean(SD) | Pain Mean(SD) |
|---|---|---|
| Contact angle (deg) | 98.4(19.7) | 98.2(19.5) |
| Peak resultant force at hand-rim | 55.1(16.01) | 69.3(23.28) |
| Mean push speed (m/s) | 1.13(0.05) | 1.11(0.04) |
| Trunk flexion angle (deg) | 3.3(0.6) | 3.3(0.8) |
| Number of pushes | 139.2(29.3) | 145(31.6) |

* P<0.05

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# Chapter 5

# Relationship between cycle-to-cycle structure in variability and shoulder pain in manual wheelchair users

**Chapter abstract:** The use of a manual wheelchair places considerable repetitive mechanical strain on the upper limbs leading to shoulder pain. While recent research indicates that the amount of variability in wheelchair propulsion variables is related to shoulder pain, there has been minimal enquiry in studying the relation between shoulder pain and the time dependent structure of variability in wheelchair propulsion variables. Data from 27 experienced adult manual wheelchair users with and without shoulder pain from varying disability demographics was analyzed. Participants propelled their own wheelchair fitted with SMARTWheels on a roller dynamometer at 1.1 m/s for 3 minutes. Sample entropy of cycle to cycle fluctuations in contact angle and the inter push time interval was compared between the groups with and without shoulder pain using non-parametric test. The main findings were: (1) variability observed in the fluctuation in contact angle during manual wheelchair propulsion is structured (Z=3.15,p<0.05); (2) individuals with shoulder pain exhibited higher sample entropy for contact angle during wheelchair propulsion compared to those without pain ($\chi^2(1) = 6.12$, $p = 0.013$); and (3) the SampEn measure correlated significantly with the amount of self-reported shoulder pain ($r_{s(WUSPI)}$ =0.41; $r_{s(VAS)}$=0.56;p<0.05). Overall results show that

studying the time dependent structure in variability provides novel

knowledge for tracking and monitoring shoulder pain in manual wheelchair

users.

**Introduction**

Motor variability provides unique information concerning the control and health of the neurophysiologic systems (Lipsitz et al., 2004, Sosnoff et al., 2006). Researchers have shown that both the amount and the time dependent dynamical structure of motor variability provides useful insights about the control and function of the neurophysiologic systems (Newell, 1993, Sosnoff et al., 2006). The amount and structure of motor variability are distinct in that they can be uniquely influenced by experimental or population factors (Sosnoff et al., 2006). Moreover, dynamical measures of motor variability have been found to be more sensitive to pathology than distributional statistics (Slifkin et al., 1999, Stergiou et al., 2011). It is maintained that a lack of variation results in insufficient time to adapt (i.e. heal) between loading occasions.

The importance of examining the structure of variability can be seen with a brief review of locomotion research. A detailed examination of locomotion reveals that there are subtle structured (i.e. non-random) variations in the timing between each step (i.e. inter-stride interval) (Hausdorff et al., 1997). The structure of stride intervals in ambulatory walking have been shown to be influenced by pathology, developmental age, speed, and coordination pattern (walking vs. running) (Jordan et al., 2008). Currently, it is maintained that the dynamic structure of walking and running result from the complex interaction of supraspinal inputs to the spinal motor

neurons (Newell, 1993, Sosnoff et al., 2006, Lipsitz et al., 2004).

A unique form of locomotion is manual wheelchair propulsion. Although, an estimated 2 million individuals in the United States propel manual wheelchairs as their primary form of locomotion (LaPlante et al., 2010) very little motor control research has focused on it.  The majority of investigations of wheelchair propulsion have examined mean performance variables and not explicitly focused on the variability characteristics of propulsion (Boninger et al., 2002, Collinger et al., 2008, Rankin et al., 2011, Richter et al., 2007, 2011, Koontz et al., 20009, 2012, de Groot et al., 2003,2004, Kwarciak et al., 2012 ).

Recent research indicates that variability in wheelchair propulsion mechanics is related to shoulder pain (Sosnoff et al., 2015). This association between motor variability and pain is consistent with observations from occupation biomechanics and human motor control literature (Srinivasan et al.,2012, Madeleine et al., 2009, Hamill et al., 2012, Stergiou et al., 2011).

Although promising, a limitation of research by Sosnoff et al., 2015, is that it has only focused on the amount of variability in wheelchair propulsion variables. This approach seemingly ignores the time-dependent structure inherent in motor output (Lipsitz et al., 1992).

Based on the tenets of the loss of complexity hypothesis of aging (Lipsitz et al., 2004) it has been theorized that the time-dependent structure of motor output is a marker of physiological complexity and provides novel information concerning the health of the musculoskeletal system (Sosnoff et al., 2006, Stregiou et al., 2011). Specifically, it has been proposed that musculoskeletal injury leads to motor fluctuations that are more structured. For instance, Tochigi et al., 2012 utilized sample entropy to analyze the structure of variability in gait in individuals with knee osteoarthritis with and without pain. Consistent with the loss of complexity hypothesis, it was reported that the pain group had significantly lower SampEn values when compared to those without pain. In a similar fashion, researchers have successfully employed other time dependent measures such as approximate entropy to study the gait pattern related to anterior cruciate ligament (ACL) injury (Georgoluis et al., 2006) and reported lower approximate entropy values in the knee with ACL deficiency.

Presently there is no information regarding the time dependent structure of variability in the context of shoulder pain in mWCUs. Consistent with the tenets of the *loss of complexity* hypothesis it is hypothesized that individuals with shoulder pain will demonstrate lower complexity compared to those without shoulder pain. Consequently the purpose of this investigation is to examine if the variable structure in wheelchair propulsion parameters is related to shoulder pain.

**Methods**

**Participants**

   Wheelchair propulsion data from 27 experienced adult mWCUs was analyzed for this study. This dataset is a subset of data collected for a larger study focusing on wheelchair propulsion and shoulder pain (Sosnoff et al., 2015). Inclusion criteria for this secondary analysis were (1) more than one year of manual wheelchair experience; (2) between 18-64 years of age; and (3) trials with a minimum 106 wheelchair propulsion data cycles. All procedures were approved by the local institutional review board at University of Illinois, Urbana-Champaign.

**Kinematic and kinetic data collection**

   The kinematic and kinetic data collection procedures have been detailed in Chapter 2. Since participants switched the propulsion pattern used for different speeds, studying the kinematics at one specific speed (i.e.1.1 m/s) was most appropriate. Consequently, only the data belonging to trials conducted with a speed of 1.1 m/s was used for this analysis.

**Data post-processing**

   SMARTWheel data were collected at a sampling rate of 100Hz and digitally filtered with an eighth-order, zero-phase, low-pass Butterworth filter with 20Hz cutoff frequency (Collinger et al., 2008). The start and end of a

propulsion cycle was defined when the push-rim moment (Mz) was above and below 1 Nm respectively (Jayaraman et al., 2014). To reduce the transient effects, data belonging to the first five propulsion cycles were not included for this analysis (Jayaraman et al., 2014). For consistency, the number of data cycles analyzed for each participant was maintained constant at 100 cycles (i.e. starting from the $6^{th}$ cycle to $105^{th}$ cycle of a SMARTWheel data, figure 5.1(a)). The contact angle and resultant forces at the hand-rim were extracted for each participant (Figure 5.1(b)). Following this, the intra-push time interval between peak resultant force were extracted (Figure 5.1(b)). Thus, this process yielded two time series from each participant wheelchair propulsion dataset namely, (1) a time series of cycle-to-cycle contact angle (Figure 5.2(a)) and (2) a time series of the cycle-to-cycle intra-push time interval between peak resultant force during push phase (Figure 5.2(b)). Sample entropy was then computed for these two time series (see below).   A custom developed MATLAB code was used to accomplish all data post-processing. Based on the VAS scores, for the shoulder pain group, the data belonging to the side with greatest shoulder pain level was analyzed (right (n=11) and left (n=2)), while the data belonging to the dominant hand (right (n=12) and left (n=2)) was analyzed for the group without shoulder pain.

**Structure in variability**

Complexity analysis quantifies the time dependent regularity and predictability of a time series. There are various nonlinear dynamics tools to measure the complexity of a physiological time series (Stergiou et al., 2004). The choice of a specific tool depends on the characteristics of data being analyzed. In this investigation, we utilized sample entropy (SampEn), a widely utilized approach (Yentes et al., 2013, Tochigi et al., 2012).

SampEn is a nonlinear metric used for measuring the complexity of a time series. The magnitude of the SampEn is an indication of the regularity or irregularity of a particular time series. SampEn values typically ranges from 0 to 4.   Values closer to 0 are consistent with greater regularity, such as a sine wave, while values nearing 4 represent greater irregularity such as pink noise (Richman et al., 2000). Higher values of SampEn indicate that the time series is more unpredictable (i.e. unstructured).

Details of the sample entropy algorithms and input parameters used are available elsewhere (Richman et al., 2004; Lake et al., 2002). A MATLAB codes obtained from http://www.physionet.org/physiotools/ was used for computing SampEn (Goldberger et al., 2000). The performance (SampEn magnitudes) of the software program was tested with synthetically generated signals (Chapter addendum material: Section A) to ascertain the validity of the code. The tests revealed that the SampEn values obtained

from using the software program provided by physionet.com is comparable with those reported in literature (Chapter addendum: Section A).

The parameters chosen for the SampEn analysis were (m=2, r=0.2) for the contact angle and (m=2,r=0.15) for the inter-push interval. The justifications for these parameter choices are provided in the Chapter addendum: Section B.

From each participant's post-processed data, the SampEn for the original time series belonging to the contact angle and the within individual inter-push time interval between peak resultant force was calculated. To statistically establish if the SampEn measure obtained from the original time series was indeed structured and not random, a simple surrogate analysis was carried out (Shelhamer, 2006).

Surrogate data sets were generated by randomly shuffling the original time series in order to determine whether the original time series was structured (Figure 5.3). Such random shuffling preserves the distributional statistics (i.e., mean, standard deviation, and higher moments) between the original time series and its corresponding surrogate time series pool as they contain the same elements; however, the sequential ordering is destroyed (Jordan et al., 2008). Each participant's original time series was randomly shuffled 100 times to produce a pool of corresponding surrogate data. This surrogate procedure resulted in a pool of hundred surrogate time series for

each original time series. The sample entropy values were computed for each of the 100 surrogate time series in the pool and then averaged to generate a mean sample entropy for the surrogate pool (Figure 5.3). Finally non-parametric pair-wise tests (p<0.05) comparing the SampEn values between each original time-series with the corresponding mean sample entropy obtained from its surrogate pool was conducted. If the sample entropy value of the original time series was significantly (statistically) smaller than its corresponding surrogate counterpart, then this indicated that the original data was not randomly derived and therefore its variability may be structured (Hausdroff et al., 1997, Newell et al., 1993, Dingwell et al., 2000).

The SampEn magnitudes can be influenced by non-stationarity of the time series (Yentes et al., 2013, Slifkin et al., 1999). To check for this influence, the SampEn for both the spatial and temporal variables were calculated for the differenced time series. No significant differences were found between the SampEn calculated from the original and the differenced times series (Chapter addendum: Section C). Thus the SampEn from the original time series were used for statistical analyses.

**Statistical Analysis**

IBM SPSS (version 21) was used for conducting all the statistical tests. The significance level for all the statistical tests were set at p≤0.05. All

descriptive values are reported as mean (SD) unless otherwise noted. The between group factor for all the statistical tests was set as shoulder pain (with pain = 1; No pain = 0).

Normality checks for data distribution using Shapiro-Wilk tests revealed that the age, self-reported WUSPI scores, mean inter push time interval, mean speed and the SampEn required    non-parametric statistical tests.

**Between group comparisons for independent variables**

Participant demographics information (age, body weight, and manual wheelchair propulsion experience in years) and self-reported scores (WUSPI & VAS) were treated as independent variables.

A series of two tailed Mann-Whitney U tests were conducted to check if demographic variables (age and self-reported shoulder pain (WUSPI & VAS)) were significantly different between groups. A series of two tailed independent t-tests were conducted to check if significant between group differences in existed in body weight and wheelchair propulsion experience. Separate Chi-squared tests were conducted to check if category variables (gender and propulsion pattern type) were significantly different between the groups with and without shoulder pain.

**Mean wheelchair propulsion variables**

A series of two tailed Mann-Whitney U tests were conducted to check if the mean inter push time interval and mean push speed were significantly different between groups with and without shoulder pain. A series of two tailed t-tests were performed to check if statistically significant group differences existed in other mean wheelchair propulsion variables (i.e. contact angle, peak resultant force at hand-rim).

**Pair-wise comparisons between SampEn original and surrogate data**

To test if the SampEn of the original time series were significant different from their surrogate counterpart, a series of pair-wise Wilcoxon signed rank test were conducted for the SampEn of contact angle and inter push time.

**Shoulder pain and SampEn of spatiotemporal variables**

To investigate the main effect of shoulder pain on SampEn, the mean SampEn obtained from the original time series for the contact angle and inter-push time interval were compared between the groups using Kruskal-Wallis ANOVA's.

Spearman's rank correlational analyses were conducted to investigate if the SampEn for the contact angle and time interval was correlated with the self-reported pain scores (WUSPI; VAS).

## Results

### Demographics

No significant group differences in demographics were observed, (*P*'s>0.05; Table 1). Per design the shoulder pain group reported higher pain than the no shoulder pain group (WUSPI: [$U=17,P<0.05$] ; VAS: [$U=14.5, p<0.05$] ;Table 5.1).

### Mean wheelchair propulsion variables

No significant differences were observed in mean spatial-temporal propulsion variables between groups (*P's*>0.05; Table 5.2).

### SampEn comparison between original and surrogated time series

Wilcoxon signed rank pair wise tests revealed that the SampEn obtained from the original time series were statistically smaller than the SampEn from their corresponding surrogates : inter push time interval :[$Z=2.59, p=0.009$]$_{(m=2, r=0.15)}$; contact angle:[$Z=3.15, p=0.002$]$_{(m=2, r=0.2)}$. This indicates that the variability structure found in contact angle and inter push interval variable in wheelchair propulsion are not random but rather structured.

**Shoulder pain and structure of propulsion variability**

The SampEn for contact angle was higher in the pain group than the no-pain group (Figure 5.4(a)). This was confirmed by Kruskal-Wallis test, $\chi^2(1) = 6.12$, $p = 0.013$. No significant main effect of shoulder pain group was observed for the SampEn of inter-push time interval (Figure 5.4(b); p>0.05).

A statistically significant moderate positive correlation was observed between the WUSPI and VAS scores and the SampEn of contact angle [rs(25)=0.41, p<0.05]WUSPI; [rs(25)= 0.56, p<0.05]VAS. There was no association between SampEn of inter push time interval and self reported pain scores (p>0.05).

**Discussion**

This investigation explored the relation between shoulder pain and the structure of motor variability in spatiotemporal variables of manual wheelchair propulsion. Three novel observations were made, (1) variability observed in the fluctuations in contract angle and inter push time interval during manual wheelchair propulsion is structured, (2) individuals with shoulder pain exhibited higher SampEn during wheelchair propulsion compared to those without pain; and (3) SampEn measures correlated significantly with the amount of self reported shoulder pain. Overall the observations suggest that SampEn analysis of wheelchair propulsion may

provide novel insights for monitoring the development and treatment of shoulder pain in mWCUs.

While the majority of wheelchair research focuses on mean kinematic and kinetics during propulsion (Boninger et al., 2002, Collinger et al., 2008, Rankin et al., 2011, Richter et al., 2007, 2011, Koontz et al., 20009, 2012, de Groot et al., 2003,2004, Kwarciak et al., 2012), growing evidence suggests that variability measures could be a sensitive marker of shoulder pain in mWCUs (Sosnoff et al., 2015). To date, the few wheelchair propulsion studies which incorporate measures of intra-individual variability have exclusively focused on linear distributional statistics (standard deviation and coefficient of variation) and not measures of time dependent structure.

The current investigation used SampEn , to quantify the structure in two propulsion parameters closely implicated with shoulder pain in mWCUs, namely, contact angle and inter-push time interval of peak resultant force, (Boninger et al., 2002, Collinger et al., 2008, Rankin et al., 2011, Richter et al., 2007, 2011, Koontz et al., 20009, 2012, de Groot et al., 2003, Kwarciak et al., 2012). Surrogate analyses revealed that the time dependent variable structure observed in both measures was not random. This observation is consistent with reports from occupational ergonomics and motor control research that time dependent structure in variability of physiologic output may offer meaningful insights concerning health and function (Madeleine et al., 2008, Madeleine et al., 2009, Vaillancourt et al.,2002, Vieluf et al., 2015).

It is important to note that differences in wheelchair users with and without shoulder pain in SampEn were only observed in fluctuations in contact angle. This is distinct from examinations of the magnitude of variability which revealed that individuals with shoulder pain had greater amounts of variability in spatial and temporal variability as indexed by coefficient of variation (Rice et al., 2014). The divergent results between time-dependent and magnitude metrics are consistent with the view that they are at least partially independent (Sosnoff et al., 2004, Sosnoff et al., 2006).

The observation that SampEn was higher for the shoulder pain group is in agreement with occupation ergonomics research (Madeleine et al., 2009). For instance, individuals with shoulder/neck pain performing repetitive upper limb tasks exhibited greater complexity in movement than healthy controls (Madeleine et al., 2008, Madeleine et al., 2009, Lomond et al., 2010, Lomond et al., 2011). Such time dependent structure in the movement has been implicated as compensatory strategies adopted by the neuromuscular system to mitigate the discomfort arising from musculoskeletal pain (Madeleine et al., 2009, Srinivasan et al., 2012). Along these lines, the higher SampEn magnitude observed for the contact angle at the hand-rim in mWCUs with shoulder pain could be a manifestation of their compensatory strategies to minimize shoulder discomfort when performing the repetitive propulsion task.

Consistent with previous research where higher discomfort (pain) levels were associated with higher spatial complexity (Madeleine et al., 2009), SampEn of contact angle was positively correlated with the self-reported pain scores. This observation suggests that, in addition to being able to differentiate between mWCUs with and without shoulder pain, SampEn may also be a potential biomarker to track shoulder pain progression in mWCUs. This exciting possibility warrants further investigation.

It is important to note that the current observations are counter to the original hypothesis. The original hypothesis that individuals with shoulder pain would have lower complexity in their wheelchair propulsion was based on the tenets of the *loss of complexity* hypothesis. The loss of complexity hypothesis maintains that with senescence and pathology there is a decline in physiological complexity resulting from reductions in control process and their interaction (Goldberger et al., 1992). A criticism of this theoretical framework is that it does not take into the intrinsic dynamics of the motor task (Vaillancourt et al., 2002; Sosnoff et al., 2006). Indeed previous research demonstrates that observed differences in complexity between pathological groups can be bi-directional and their directionality is influenced task constraints studied (Vaillancourt et al.,2002, Sosnoff et al., 2006, Vieluf et al., 2015).

Finally, while SampEn is one among the many available non-linear dynamics metrics, numerous researchers have used other techniques like de-trended fluctuation analysis (DFA) and Lyapunov exponent (Ly Ep) to study the time dependent structure in human gait (Cavanaugh et al., 2006,Hausdroff et al., 1997, Stergiou et al., 2011, Sethi et al., 2013). All these studies reported that physiologic time series complexity is sensitive to age, pathology and functional state in humans. It is important to note that the current observations concerning SampEn do not necessarily indicate existence of any long range correlations. Nevertheless, outcome from our study is a first step towards showing that pursuing further research in this direction could be beneficial and improve our knowledge to provide better health/diagnosis/intervention to prevent shoulder pain in mWCUs.

**Limitations**

Despite being novel this study is not devoid of limitations. The study is cross-sectional in nature and hence any cause/effect inference cannot be draw. The choice of metrics utilized was hampered by the short time series length which prevented other complementing non-linear analyses such as de-trended fluctuation analysis. In general, adopting a combination of non-linear methods is recommended as best practice (Stergiou 2004). The data were collected in a laboratory setting with roller dynamometer setup, so it is not clear if these results would occur in real life propulsion environment. It could be possible that complexity measures of spatiotemporal propulsion

variables other than ones considered in this analyses are more sensitive to

shoulder pain in mWCUs. This requires further analysis. The injury

demographics of our sample was diverse. Consequently, it is possible that

differences in complexity between pain groups were due to different

disability demographics. Though significant, these demographic and

environmental limitations are relatively common to wheelchair propulsion

research. Finally, despite these limitations, this investigation provides novel

contributions that are important and compliments the research findings from

previous research (Sosnoff et al., 2015, Madeleine et al., 2009, Srinivasan et

al., 2012).


**Conclusions**

This cross sectional study provided novel information regarding the

relationship between the time-dependent structure of wheelchair propulsion

and shoulder pain in mWCUs. Examining the structure of wheelchair

propulsion using sample entropy as an index of complexity, this investigation

produced three novel observations, namely, (1) variability observed in the

fluctuation in contract angle during manual wheelchair propulsion is

structured, (2) individuals with shoulder pain exhibited greater complexity as

indexed by SampEn during wheelchair propulsion compared to those without

pain; and (3) SampEn measure correlated significantly with the amount of

self-reported shoulder pain. The higher sample entropy in shoulder pain

group may be a compensatory mechanism adopted to minimize discomfort

to shoulder while performing the propulsion task. Overall, we conclude that incorporating non-linear dynamics based measures in wheelchair propulsion analyses may provide new knowledge and can be an important tool for better health, diagnosis, and therapeutic interventions to prevent shoulder pain pathology in mWCUs.
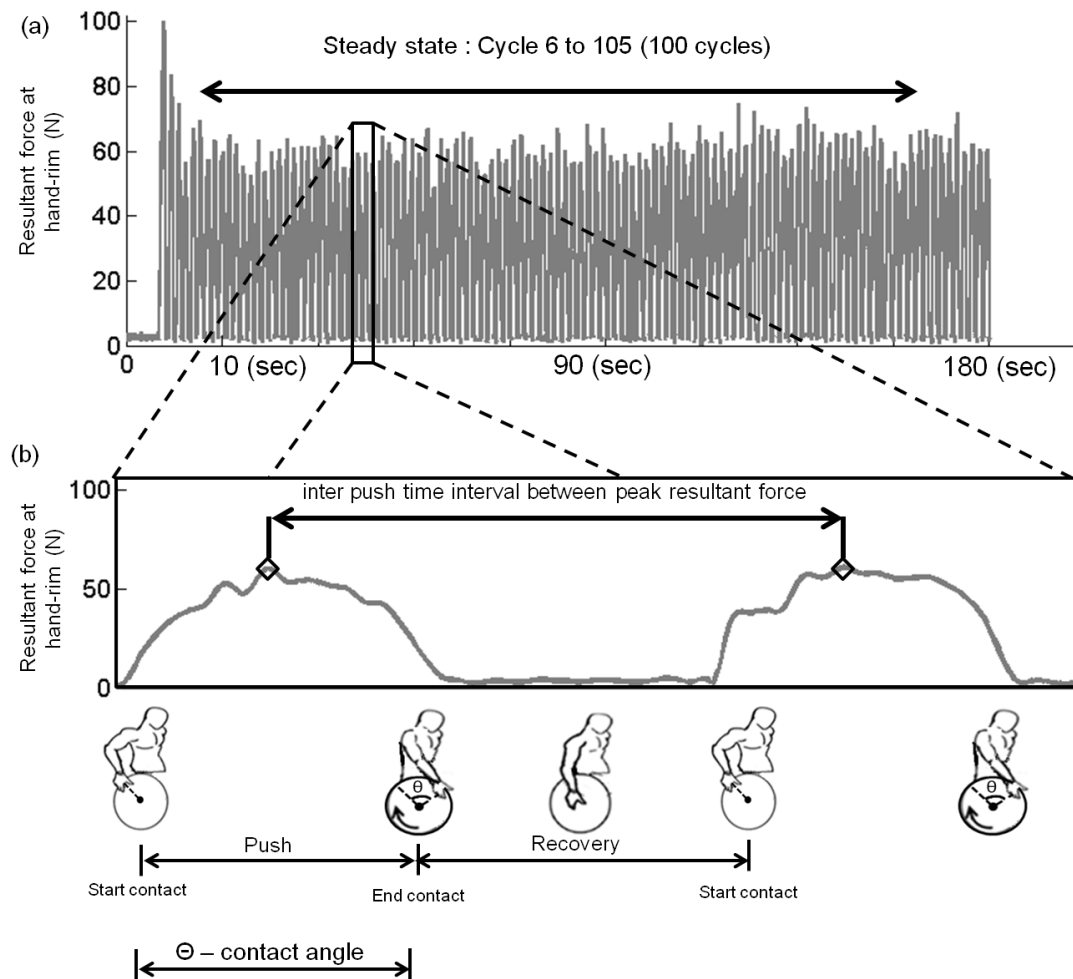
**List of figures**



Figure 5.1. Data processing and variable extraction for SampEn calculation. (a)   resultant force time series at hand-rim showing the steady state portion of the time series used for all calculations; (b) a magnified view of two sample resultant force cycles to show the details of the variables extracted. The contact angle is defined as the angular measure between the start and end of contact. The time interval between peak to peak resultant forces was extracted from cycle-to-cycle during steady state to extract the time series for inter push time interval. The cycle-to-cycle contact angle magnitude during the steady state propulsion was extracted to create the time series capturing fluctuations in contact angle.
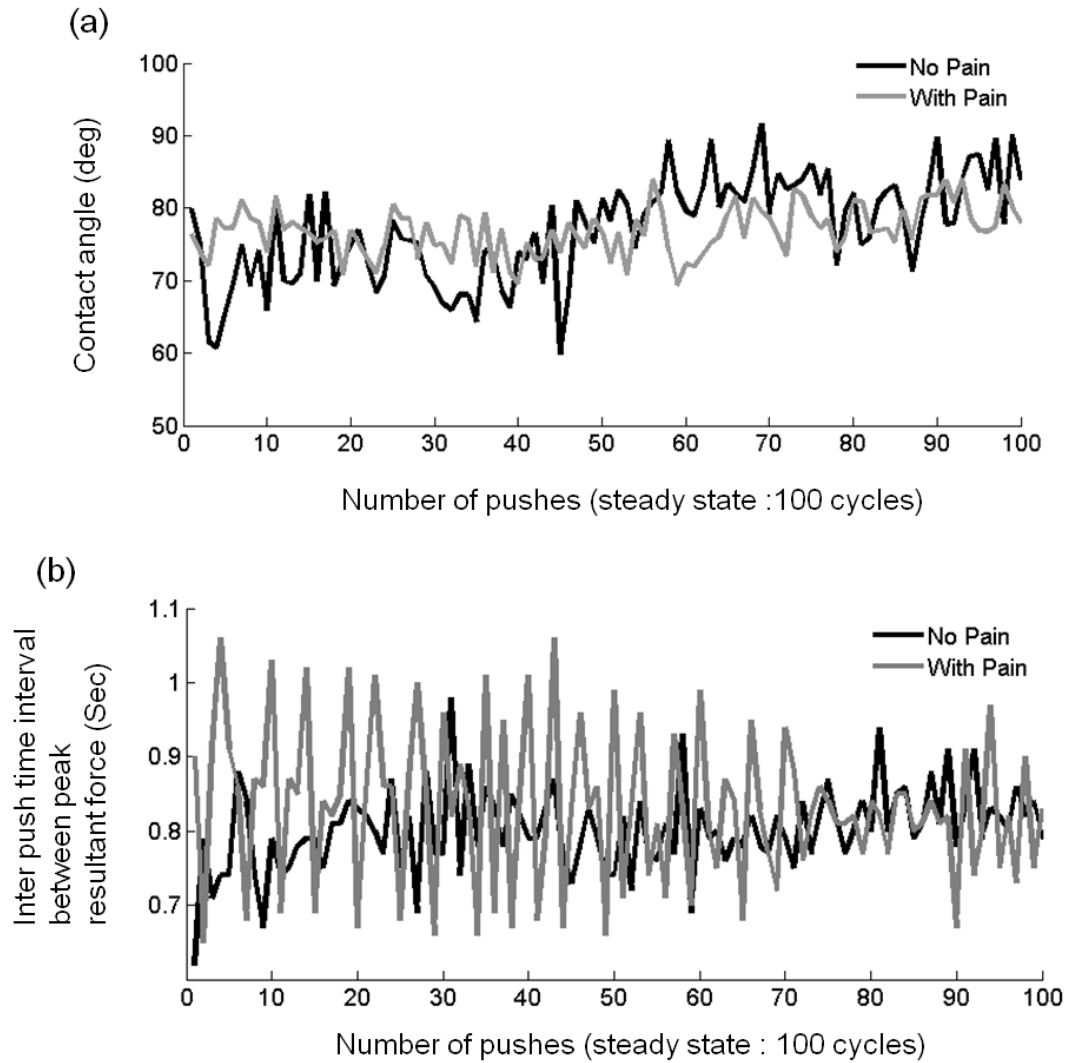
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Figure 5.2. Sample contact angle and inter push time interval time series extracted for SampEn calculation. (a) the cycle-to-cycle contact angle magnitude at hand-rim for 100 pushes; (b) the cycle-to-cycle inter push time interval between peak resultant force at hand-rim for 100 pushes.
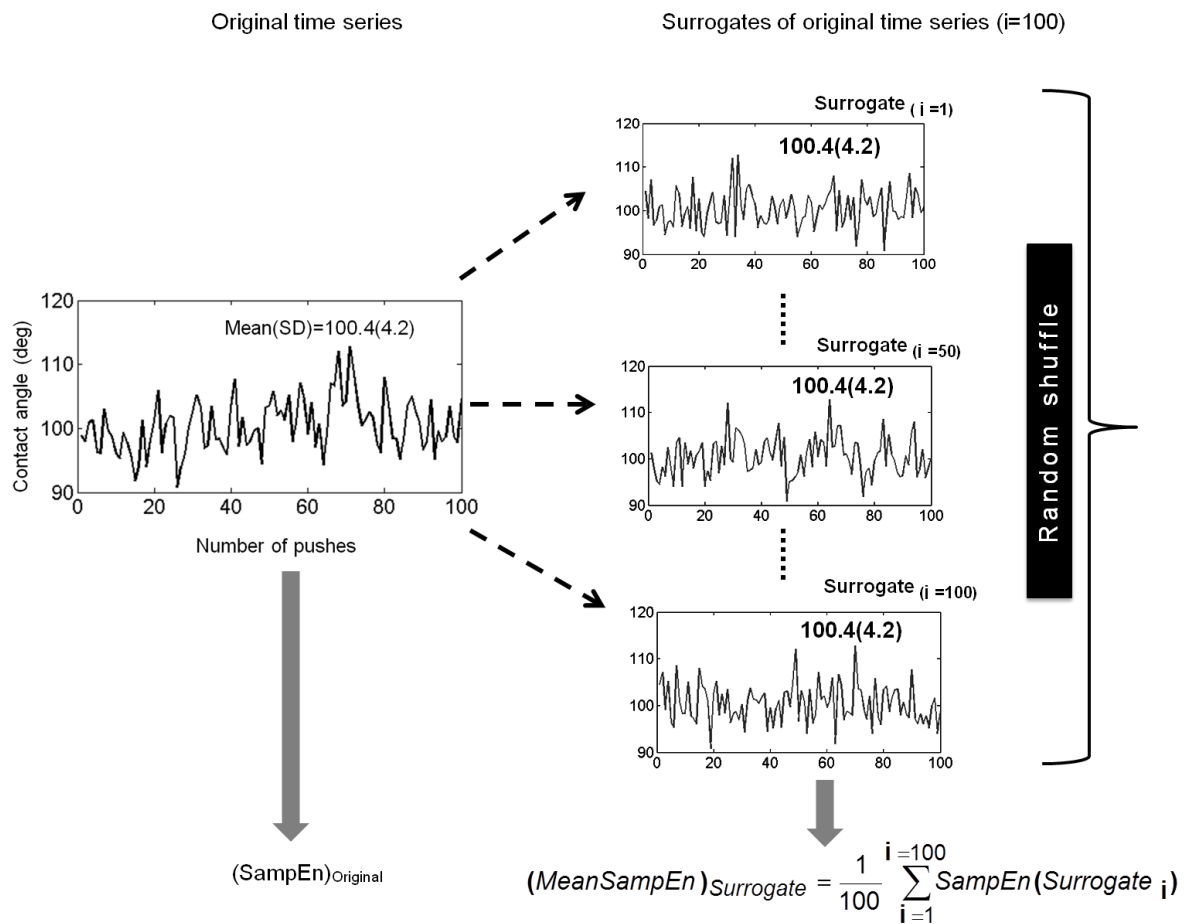
Figure 5.3. The surrogate analysis process.   The mean(SD) of the original and the surrogated time series pool are same. The surrogation procedure used was a simple random shuffling of data. A pool of 100 randomly shuffled surrogates was created from the original time series. If $(SampEn)_{original}$ is statistically lesser ($p<0.05$) than $(MeanSampEn)_{Surrogate}$, then the time dependent structure observed in the original time series is not random and may have some meaningful insight.

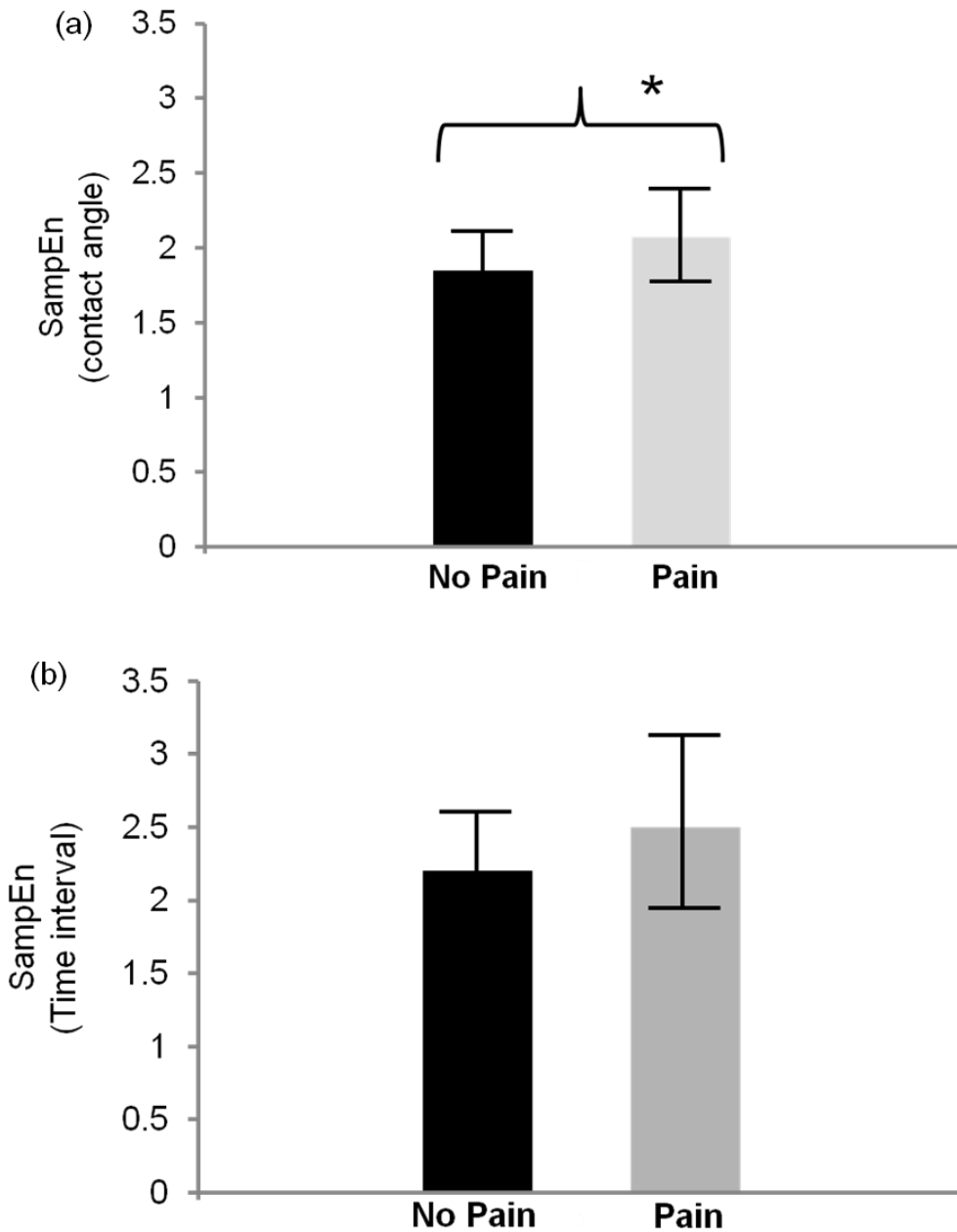Figure 5.4 Groupwise box plot comparison for SampEn. (a) the SampEn for contact angle was significantly greater (p<0.05) for the group with shoulder pain; (b) the SampEn for inter push time interval at hand-rim between groups with and without shoulder pain was not significantly different (p>0.05) between the groups with shoulder pain.

# List of tables

**Table 5.1** Demographic information

| Demographic variable | Pain (n=13) Mean(SD) | No Pain (n=14) Mean(SD) |
|---|---|---|
| Age (Yrs) | 28.23(12.52) | 21.21(4.92) |
| Body weight (LB) | 164.50(56.39) | 131.17(36.96) |
| Experience using wheelchair (Yrs) | 15.84(11.25) | 13.64(5.15) |
| Gender (F/M) | 6/7 | 6/8 |
| Injury demographics | Spina Bifida(n=5) T6-T12 (n=5) L1-L4 (n=1) Amputee- double (n=1) Sacral agenisis (n=1) | Spina Bifida (n=4) T6-T12 (n=6) L1-L4 (n=1) Amputee-single (n=1) Arthrogryposis (n=1) C7 (n=1) |
| WUSPI* | 22.84(21.27) | 3.28(5.02) |

**Table 5.2** Mean propulsion variables at hand-rim

| Mean propulsion parameters at hand-rim | Pain (n=13) Mean(SD) | No Pain (n=14) Mean(SD) |
|---|---|---|
| Peak resultant force (N) | 69.82(23.84) | 60.23(19.72) |
| Contact angle (deg) | 100.52(20.25) | 97.68(17.64) |
| Mean speed (m/s) | 1.1(0.04) | 1.1(0.06) |
| time interval between peak resultant force (Sec) | 1.15(0.22) | 1.20(0.22) |

## Chapter addendum: Selection of sample entropy parameters

### A. Checking the validity of the SampEn algorithm from physionet.com

To check the validity of the SampEn code from physionet.com, its output was benchmarked with Sample entropy values reported for standard theoretical time series in the literature (Richmann et al., 2000, Lake et al., 2002, Yentes et al., 2013). Five time series with 100 points were created, namely, (1) a periodic sinusoid; (2) a sinusoid with additive white noise; (3) a random white noise signal; (4) a periodic logistic map, and (5) chaotic logistic map. All time series were generated using custom code written in MATLAB. The details of the schemes used for generating these theoretical time series are shown in figure 5.5. The SampEn values obtained using the physionet.com code for a range of r values for each of these theoretical signals are show in figure 5.6. The SampEn magnitudes for the theoretical signals are consistent with those reported in literature (Richmann et al., 2000, Lake et al., 2002, Yentes et al., 2013). Similar to previously reported literature, the SampEn values dropped for increasing r values for a fixed m=2 (Yentes et al., 2013). Based on these observations, the SampEn code from physinet.com was deemed suitable to be used for our analyses.
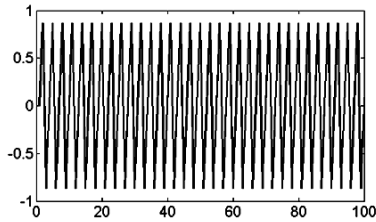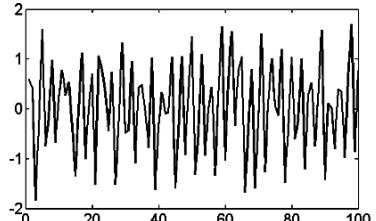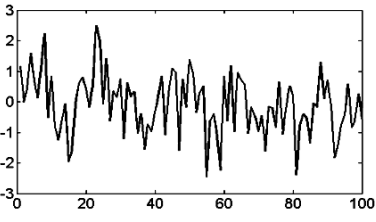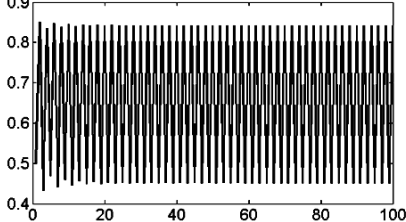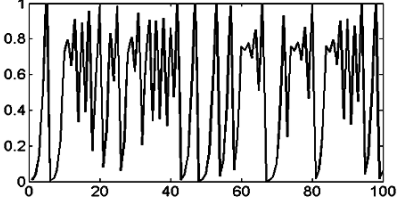
| Signal type | Scheme used for generation | Parameters | Generated signals |
|---|---|---|---|
| Periodic Sinusoid | $Y = A*Sin(\alpha t)$ | A = Amplitude; $\alpha$=circular frequency $(2*\pi*f)$; t=time sample length. |  |
| Sinusoid+ white_noise | $S = Y +$ $(N_A*$ white_noise$)$ | $Y = A*Sin(\alpha t)$; $(N_A*$ white_noise$)$ = additive white noise. Where, $N_A$= White noise amplitude; white_noise term generated using (randn) command in MATLAB. (randn-normally distributed pseudo random white noise ) |  |
| Random white noise | S=rand | uniformly distributed random white noise was generated using (rand) command in MATLAB. |  |
| Periodic logistic map | $x_{n+1}=a*(x_n)*(1-x_n)$ | a=3.4 and initial condition $x_n$=0.5. xn is a real number between (0-1) |  |
| Chaotic logistic map | $x_{n+1}=a*(x_n)*(1-x_n)$ | a=4 and initial condition $x_n$=0.99. xn is a real number between (0-1) |  |

Figure 5.5 Schemes used to generate the synthetic signals for SampEn calculations. The X axes in all plots denote the number of time sample points (100 points). The number of time series points was chosen to match with the number of points analyzed in the data. Y axes denote the amplitude for each signal.

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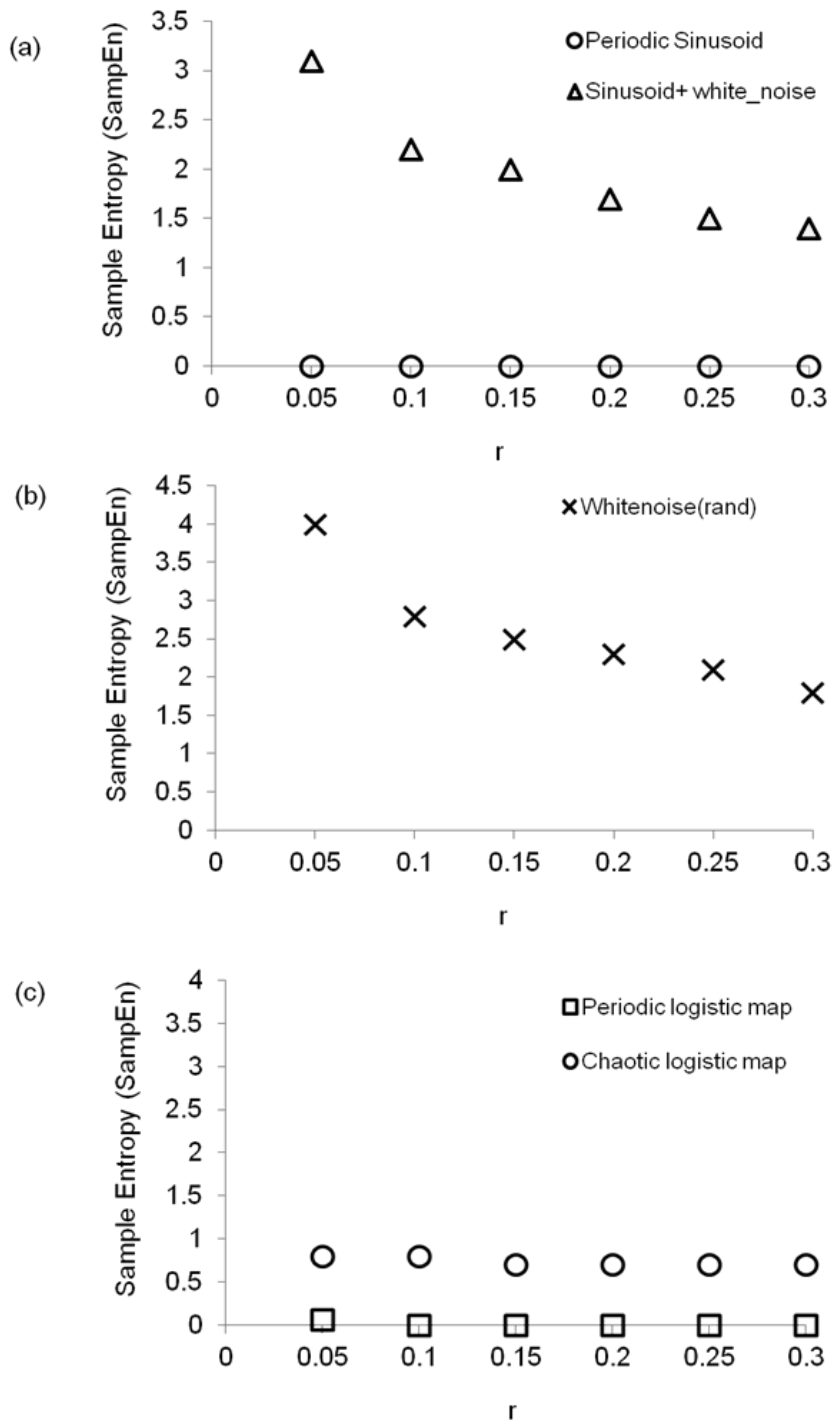
Figure 5.6 SampEn values obtained for the theoretical time series using the code from physionet.com. For all the calculations a m value of 2 and r values ranging from 0.1 to 0.3 at steps of 0.05 were used. (a) shows the SampEn for a regular sinusoid and a sinusoid with additive white noise for a range of r values; (b) SampEn for a random white noise signal; (c) shows the SampEn for a regular logistic map and a chaotic logistic map for a range of r values. Values and trends are consistent with previous literature (Yentes et al., 2013).

## B. Sample entropy (SampEn) parameter selection

### B1. Justification for selection of the embedding dimension (m)

For all SampEn calculations the embedding dimension (m) used was $m=2$. We offer two rationales for choosing m=2, namely, (1) since our focus was to capture the temporal evolution of cycle to cycle variability, a value of m=2 was chosen (m=2 signifies that the dynamics between the adjacent pairs of events are studied during the template similarity matching process) (Yentes et al., 2013) and (2) from a numerical perspective for reliable SampEn results, the recommended thumb rule relation between the time series length and embedding dimension (m) is usually $10^m – 20^m$ (Richmann et al., 2000, Lake et al., 2002). Since the length of the time series we studied had 100 points, based on the above recommendations, a value of m=2 (i.e. 100 time series points= $10^2$) and range of r values, [$r$=0.15, 0.2, 0.25 and 0.3] where m is the embedding dimension and r is the tolerance (Richman et al., 2004; Lake et al., 2002, Yentes et al., 2013) were most appropriate for our dataset.

The ranges for r values for m=2, were decided based on previous research on human movement, which recommended r values greater than 0.1 when m=2 (Yentes et al., 2013). The appropriate r was chosen based on the relative consistency of the SampEn values observed in the dataset (Yentes et al., 2013).

## B2. Justification for selecting the tolerance (r) values

Based on recommendations from literature, a range of r values were tested before picking the final r value. The ranges chosen were [$r$=0.1,0.15, 0.2, 0.25 and 0.3]. These ranges were picked based on recommendations for human movement data (Yentes et al., 2013). Based on the SampEn trends from the range of r value, the appropriate tolerances (r) for the spatial and temporal variables were picked.

In order to pick the correct r value, the relative consistency of the SampEn magnitudes between the pain and no pain groups were checked. A particular r is termed as '*relative inconsistent'* if there occurs a switch in SampEn magnitude direction between the pain/no pain groups when compared to the trends obtained from the for the preceding/proceeding r value (Pincus et al., 1994, Yentes et al., 2013). It was observed that the relative magnitude of the group mean SampEn for the inter push time interval switched direction between the pain/no pain groups at a parameter choice of r=0.25 (Figure 5.7). It can be observed from figure.B1, that at r=0.2, the pain group demonstrated higher SampEn magnitude than the group without pain. At r=0.25 this trend switched direction with the no pain group having higher SampEn magnitude (Figure 5.7). Due to this relative inconsistency, the r values around 0.25 were deemed as not ideal choices (i.e. r values 0.2,0.25 and 0.3 are not ideal) (Yentes et al., 2013). Based on this observation an r value of 0.15 was chosen for the inter push time

interval).

The SampEn trends for contact angle variable was very consistent between group during all the ranges of r value studied (Figure 5.8). Following this, an r=0.2 was chosen for the spatial variable in-order to be consistent with human movement literature (Yentes et al., 2013) thus enabling benchmarkability. Consistent with previously reported literature, the SampEn magnitudes for both, the contact angle and the inter push time interval decreased with increasing r values (Yentes et al., 2013).

Figure 5.7 Selecting r value for the inter push time interval. The SampEn values between the pain/ no pain groups for different r values. At r=0.25 a relative inconsistency in trends observed when compared to the SampEn magnitude trends at r=0.2.
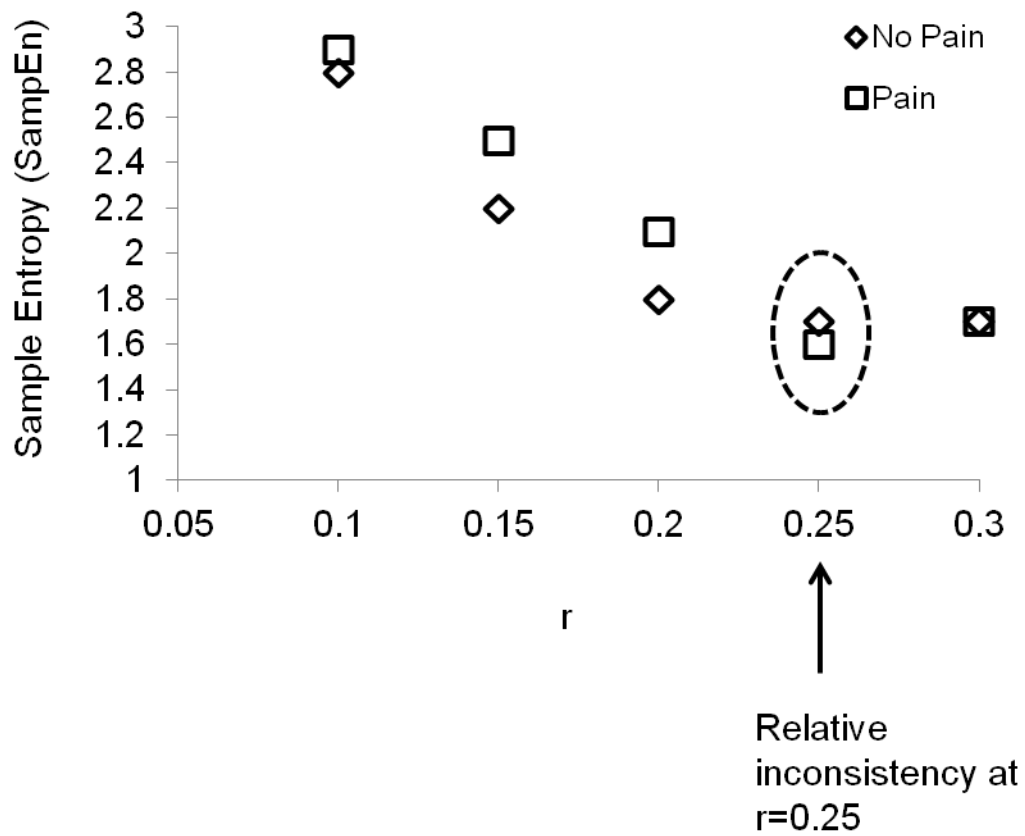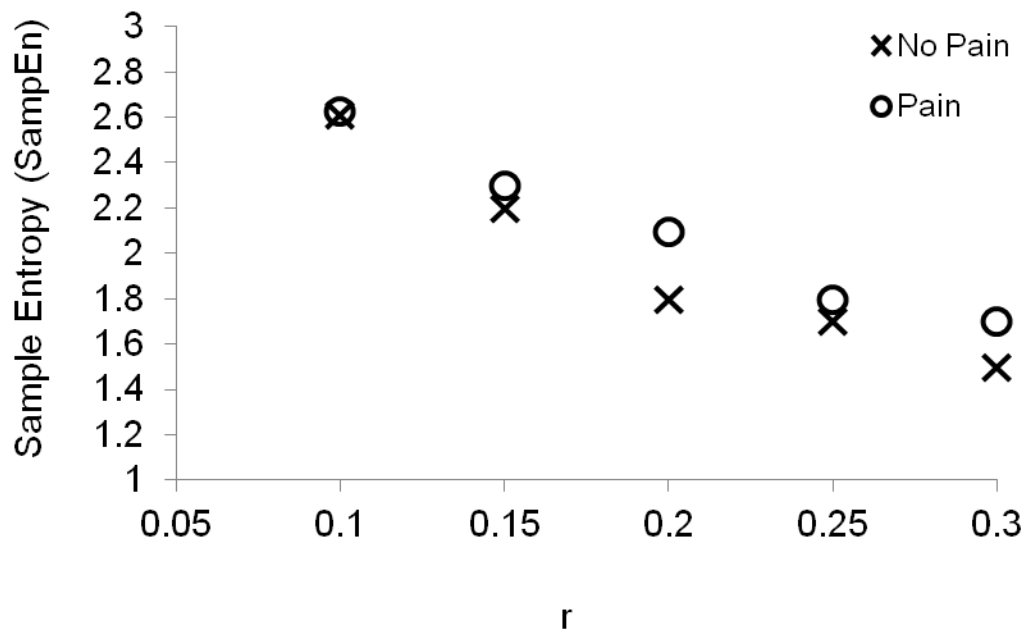
Figure 5.8 Selecting r value for the contact angle. The SampEn values between the pain/ no pain groups for different r values. At r=0.25 a relative inconsistency in trends observed when compared to the SampEn magnitude trends at r=0.2.

## C. <u>Check for Non-stationary time series effect to SampEn</u>

Non-stationary time series data can lead to spurious SampEn value, leading to wrong inferences. To avoid this, the SampEn between the original and a differenced version of the original time series were computed for both the spatial and temporal variables. Wilcoxon signed rank pair wise tests revealed that the there were no statistically significant difference in SampEn values obtained from the original and the differenced time series ($p>0.05$; group mean(SD) shown in Table 5.3).

**Table 5.3** Group mean (SD) values of SampEn between the original and the differenced time series

| Variables for which SampEn were computed | Between group factor | Group mean (SD) of SampEn for original time series | Group mean (SD) of SampEn for differenced time series | m | r |
|---|---|---|---|---|---|
| | | | | | |
| Spatial (Contact angle) | No Pain | 1.8(0.23) | 1.8(0.27) | 2 | 0.2 |
| | Pain | 2.1(0.24) | 2.0(0.13) | 2 | 0.2 |
| | | | | | |
| Temporal (Time interval) | No Pain | 2.2(0.41) | 2.2(0.40) | 2 | 0.15 |
| | Pain | 2.5(0.60) | 2.5(0.60) | 2 | 0.15 |

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# Chapter 6

# Prototype and validation of custom wearable technology for manual wheelchair users

## Introduction

The reseach studies reported in Chapters 3,4 and 5 all have a common drawback, they are cross section in nature. Further, the data was collected in a laboratory environment. To overcome these limitations a frameowrk to collect wheelchair propulsion data from the user in his/her activity of daily living is proposed. User data collected outside the laboratory is needed to understand the day-to-day propulsion practices that lead to shoulder pain. At present, the widely used equipment for wheelchair propulsion data collection is an instrumented wheel, called, SMARTWheel which is an expensive equipment designed for laboratory based data collection.

Further, research studies to date show that training manual wheelchair users on proper pushing technique and providing continuous feedback on their technique minimizes the risk of shoulder injury. However such technology laden training is not widely accessible to MWCUs and clinicians. This inadequate state of wheelchair propulsion training is due to two factors: (1) the high cost of equipment (~ US $ 40,000 for force sensing instrumented wheels), and (2) the limited number of available professionals with the capacity to provide such training. Moreover, manual wheelchair users face various barriers (e.g. transportation, cost, etc) that reduce their

ability to visit specialized clinics to receive propulsion training. This lack of training exposes them to higher injury risk, and health care costs (i.e. treatment after onset of injury). Consequently, in this concluding chapter of the dissertation the preliminary results from a prototype wearable device, custom developed for manual wheelchair users are reported.

This wearable technology provides individuals access and self-monitor the day-to-day wheelchair propulsion activity through their mobile devices (via wireless integration of wearable sensor data). The goal of this mobile technology is to increase user's awareness on the repetitive usage of their arm and wheelchair propulsion metrics that relate to injury. Integration of such technology and novel research information into rehabilitation and home-based technologies may pave the way for new interventions for tracking, treating, and/or preventing shoulder related overuse pathologies in the manual wheelchair population.

This concluding chapter reports the preliminary validation of this custom developed wearable device. Wheelchair propulsion data obtained from the device will be benchmarked with data from the currently available technologies for tracking manual wheelchair propulsion (SMARTWheel and motion capture). Finally a framework is proposed for incorporating the research finding into the custom developed wearable technology for home-based rehabilitation training purposes.

**The custom developed device**∗

Embedded force and acceleration sensors are placed on a glove. The custom developed device is capable of accurately measuring and securely transmitting real-time wheelchair propulsion data via Bluetooth to any mobile device (Figure 6.1). The sensor/hardware modules are easily attached and detached from most pairs of gloves. The hardware encloses a chargeable lithium ion battery. The data collected can be used to provide a bio-feedback to the user on shoulder activity. For instance, if the arm kinematics during movement is deviating from the usual smooth pattern, the data collected from the device can be used to provide a visual or auditory bio feedback to user to increase their awareness about the situation. The benchmarking of the signal from the custom developed device with SMARTWheel and motion capture will be reported.

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*The hardware was designed and developed by Mr.Adam Burns, MS in Civil and Environmental Engineering. I was instrumental in device selection procurement and device testing once the hardware was ready.

## Preliminary validations

### Force sensor validation

In order to validte the device, the custom device was used on a wheelchair fitted with the SMARTWheel. The participant, pushed the wheelchair wearing the cusotme device on a glove. This enabled us to collect both SMARTWheel data and the data from the sensor simultaneously to benchmark if the force values as recorded from the custom device sensor compare well with that from the SMARTWheels (figure 6.1). The data were recorded at different speeds (figure 6.3 ; 0.5 m/s , 1.1 m/s and 1.3 m/s).

### Acceleration sensor validation

To validate the acceleration sensor, the wrist acceleration signatures from the custom device while pushing a wheelchair was compared with wrist acceleration signature data for wheelchair propulsion collected using a 10 camera motion capture system.

### Validation results

Our preliminary observation shows that the data obtained using the custom device were comparable with th SMARTWheel and the motion capture device. The peak resultant force at hand-rim compared between the custom device and SMARTWheel were within ~5% magnitude error and the push counts were accurate between both devices (Figure 6.2; Figure 6.3).

In a similar fashion, a comparison of the wrist acceleration signatures for the four propulsion pattern types obtained using the custom device with that a 10 camera motion capture system is shown in figures 6.4 - 6.7. It is to be noted that the plots in these figures are from different individuals, but the features for the wrist acceleration compared well between the two systems.

**Proposed framework**

The proposed framework is to use this custom developed dvice with cloud integration for continuous monitoring/tracking and training of wheelchair users (Figure 6.8). The goal is to collect propulsion data from the user on a day-to-day basis and provide useful feedback information leading to adopting best practices of wheelchair propulsion that could minimize injury risk.

The information about the three biomarkers identifed in Chapters 3,4 and 5 can be provided as a feedback to keep users aware of their safe propulsion practice. One such sample interface is shown in Figure 6.9. A safe zone of propulsion can be established for each individual and a biofeedback (beep/buzz/email) can be triggered whenever an activity happens to fall above the threshold. For instance, from Figure 6.9, an individuals can be easily observe the alert that around noon and around 5 PM, there were

propulsion activities regarding his/her wheelchair propulsion that could have posed a risk to his/her arms. The user, once made aware of this can review what the activities during these time that could have led to such risk and seek help to make necessary adaptations to reduce risk. As one possible scenario, drawing from Figure 6.9, taking a specific path/route while propelling a wheelchair to go for lunch or evening stroll, may be a reason. The goal is to create this awareness, so that users can self-monitor and reduce situations that could lead to injuries.

**List of figures**



**Figure 6.1**  Custom hardware being tested.  F- Force sensor. A-Acceleration sensor.

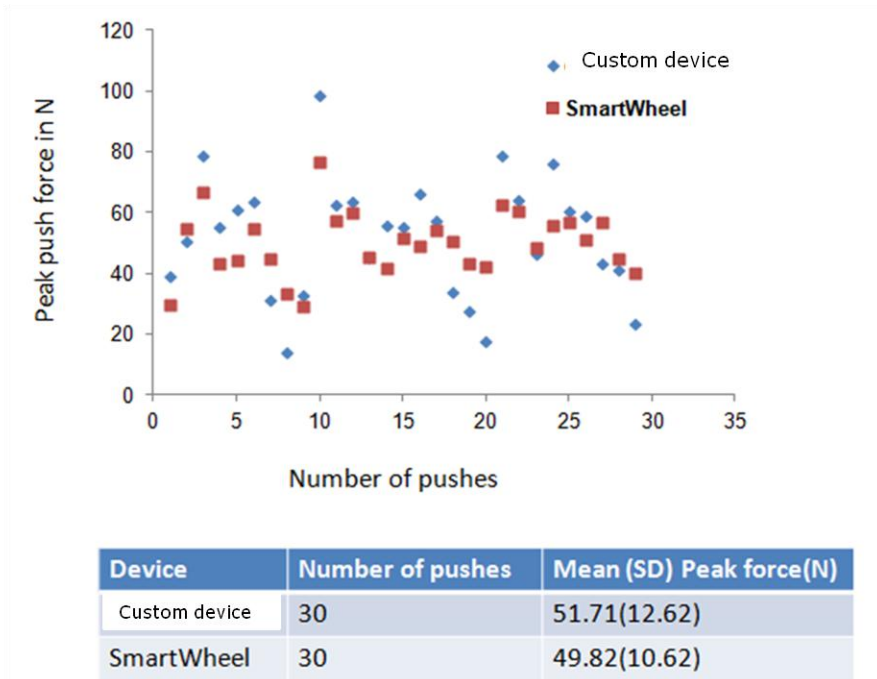| Device | Number of pushes | Mean (SD) Peak force(N) |
|---|---|---|
| Custom device | 30 | 51.71(12.62) |
| SmartWheel | 30 | 49.82(10.62) |

Figure 6.2 Pilot data validation comparing the peak resultant force at palm between the custom device and the SMARTWheel.

**Push speed : 0.5 m/s ; Push duration 1 minute**

| Device | Number of pushes as counted by devices | Mean (SD) Peak force(N) |
|---|---|---|
| Custom device | 16 | 37.71(10.13) |
| SmartWheel | 16 | 35.13(8.86) |

**Push speed : 1.1 m/s ; Push duration 1 minute**

| Device | Number of pushes as counted by devices | Mean (SD) Peak force(N) |
|---|---|---|
| Custom device | 30 | 53.29(9.67) |
| SmartWheel | 30 | 50.94(6.32) |

**Push speed : 1.5 m/s ; Push duration 1 minute**

| Device | Number of pushes as counted by devices | Mean (SD) Peak force(N) |
|---|---|---|
| Custom device | 41 | 67.29(13.90) |
| SmartWheel | 41 | 64.47(11.89) |

Figure 6.3 Comparing peak force at wrist between the SMARTWheel System and prototyped custom developed wearable technology.
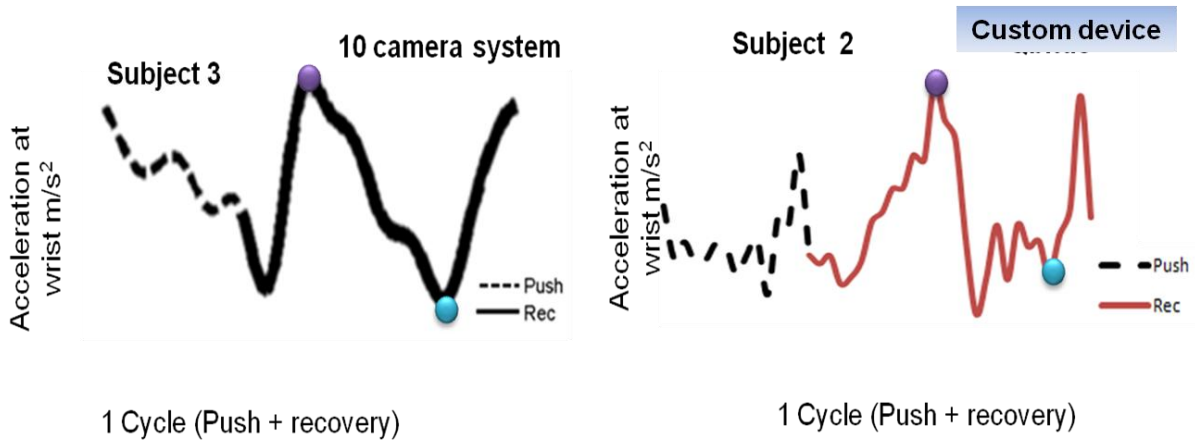
Figure 6.4 Wrist acceleration data for a semi-circular propulsion pattern. Acceleration data from custom device (right) identifies all salient features as that from the 10 camera motion capture system (left- Cortex). Sample data shown in this figure are from different participants.
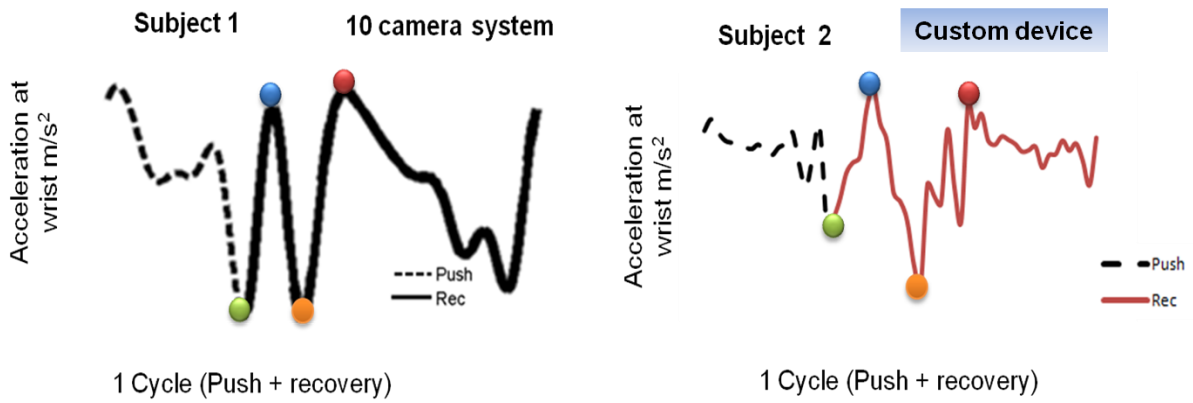
Figure 6.5 Wrist acceleration data for a double-loop propulsion pattern. Acceleration data from custom device (right) identifies all salient features as that from the 10 camera motion capture system (left- Cortex). Sample data shown in this figure are from different participants.

Figure 6.6 Wrist acceleration data for a single-loop propulsion pattern. Acceleration data from custom device (right) identifies all salient features as that from the 10 camera motion capture system (left- Cortex). Sample data shown in this figure are from different participants.

Figure 6.7 Wrist acceleration data for an arc propulsion pattern. Acceleration data from custom device (right) identifies all salient features as that from the 10 camera motion capture system (left- Cortex). Sample data shown in this figure are from different participants.
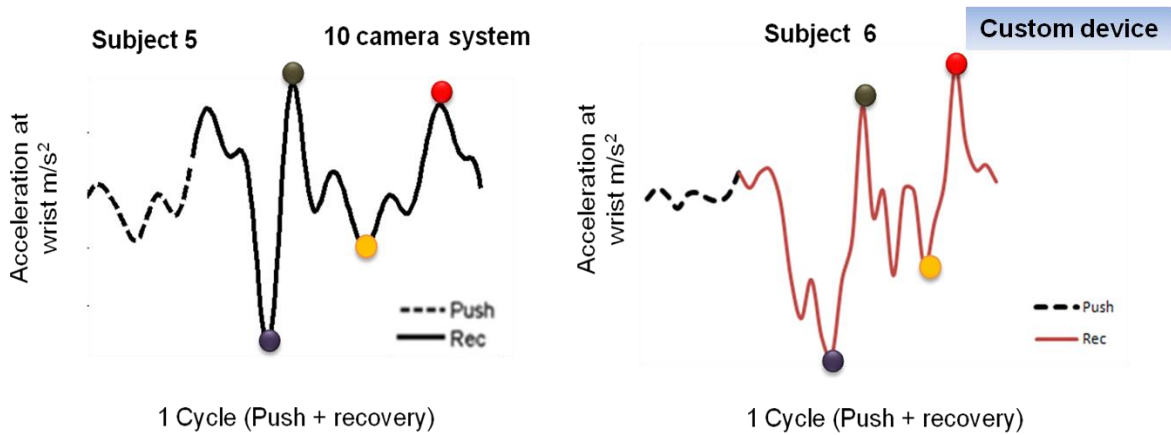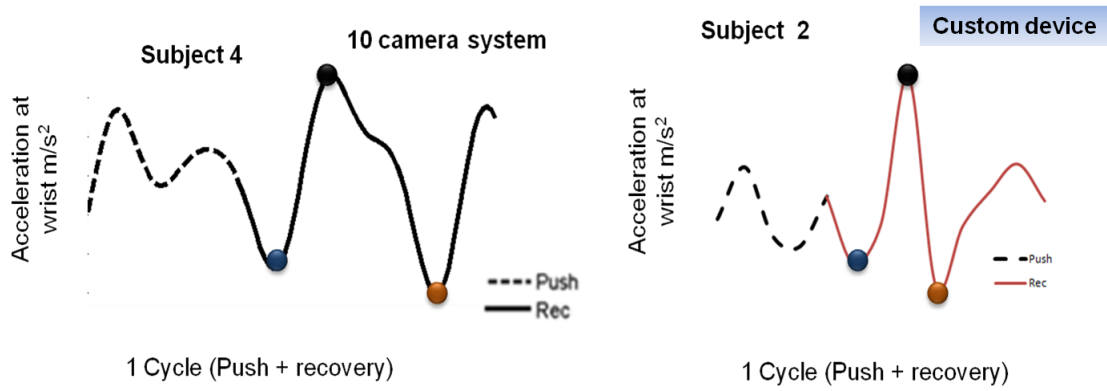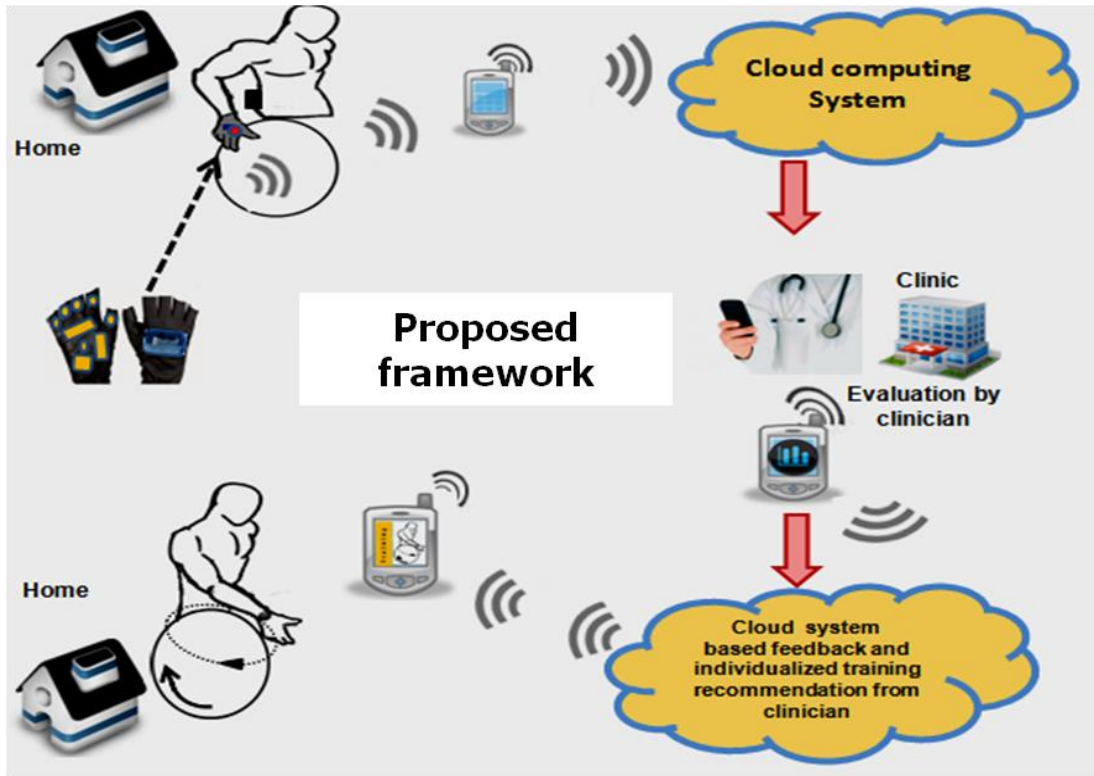
Figure 6.8 The proposed framework. A cloud integrated system to minimize injury risk in wheelchair users.
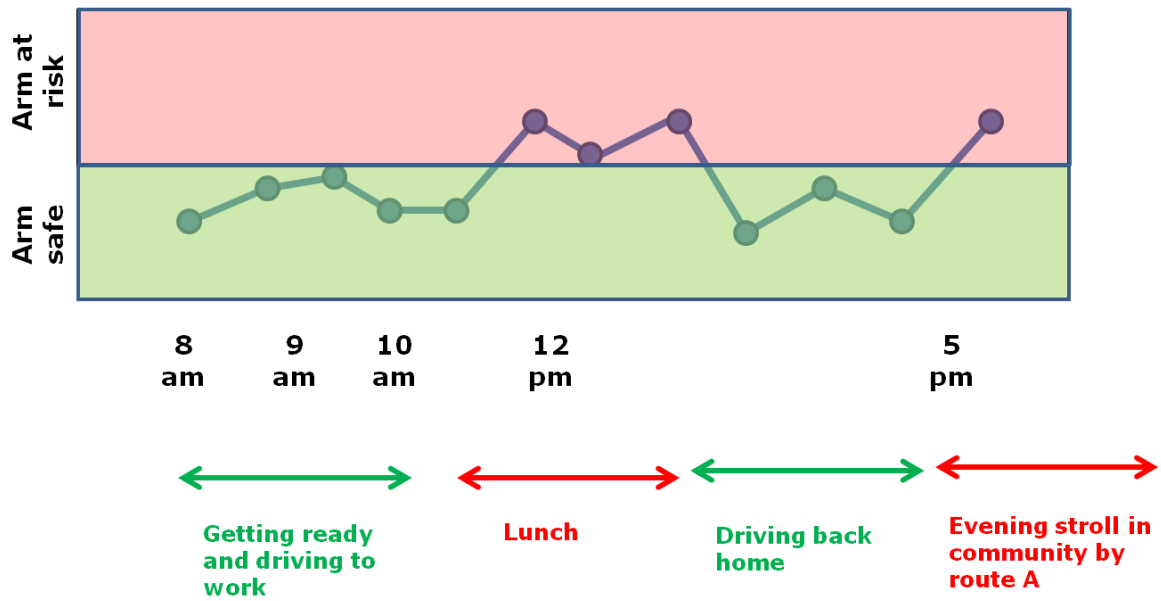
Figure 6.9 A sample interface to track arm usage and increase awareness to avoid situation that can cause risk of injury.

## Chapter 7

## Overall conclusions and future directions

Shoulder pain occurring in manual wheelchair users arising from the mechanical strain of repeated pushing is a multi-faceted problem. To identify biomarkers of shoulder pain in manual wheelchair users, to date research has taken two main approaches, (1) a pure biomechanical approach and (2) improving the wheelchair design (ergonomics). Although valuable in establishing clinical guidelines on best practices of wheelchair propulsion, the previous research have had very limited success in extracting biomarkers of shoulder pain from propulsion data. Consequently, this dissertation started with the main aim to identify biomarkers of shoulder pain in manual wheelchair users using a multi-disciplinary approach.

The dissertation implemented theories and approaches from occupation ergonomics, human motor control and non-linear dynamics to identify three such biomarkers. Chapter 3 of this dissertation examined the jerk during recovery phase of wheelchair propulsion to investigate the differences in kinematics between manual wheelchair users with and without shoulder pain. Chapter 4 of this dissertation explored the trunk kinematics during wheelchair propulsion between groups with and without shoulder pain. Chapter 5 investigated the *time dependent structure in variability of wheelchair propulsion variables* between groups with and without shoulder

pain. Finally Chapter 6, prototyped and benchmarked the data quality from a custom developed wearable device, which has capabilities to implement the findings from Chapters3,4 and 5 in real-time. Further, Chapter 6 also proposed a framework to implement this prototyped wearable device for continuous propulsion data monitoring via a mobile device and home based rehabilitation training.

**Major findings and implications**

**Shoulder pain and Jerk in wheelchair propulsion**

Jerk, the third derivative of position has been widely employed in clinical rehabilitation and human motor control research to quantify movement smoothness and evaluate the performance of upper limb tasks (Hogan et al., 1987, Flash., 1990, Chang et al., 2005, Caimmi et al., 2008). Occupational ergonomics research has revealed distinct differences in arm jerk between movements in individuals with and without shoulder pain (Cote et al., 2005). However, there has been minimal inquiry on understanding wheelchair propulsion kinematics from a human movement ergonomics perspective. Consequently, this investigation employed an ergonomic metric, jerk, to characterize the recovery phase kinematics of two recommended manual wheelchair propulsion patterns: semi-circular and the double loop. Further it examined if jerk is related to shoulder pain in manual wheelchair

users. Two hypotheses were postulated, namely, (H1) that individuals using a SC recovery pattern will experience lower jerk magnitudes at their wrists than individuals using a DLOP recovery pattern. (H2) that individuals with shoulder pain will minimize peak jerk magnitude at their upper arm joints during the recovery phase kinematics in an effort to avoid pain. H1 rested on the logical rationale that the arm's movement trajectory during a SC pattern is simpler than a DLOP pattern. H2 was based on the observation that the neuromuscular system avoids large acceleration changes to avoid pain (Berret et al., 2008).

Overall observations were, (1) the recovery phase kinematics of individuals using a SC recovery pattern placed lower jerk magnitudes than those using a DLOP and (2) mWCUs with shoulder pain had lower peak jerk magnitude during the recovery phase of wheelchair propulsion.

The logical reason for the DLOP recovery pattern to incur greater $J_c$ was attributed to the joints kinematics undergoing sharp directional turns, leading to frequent switching between acceleration and deceleration during the recovery trajectory. In contrast, when executing a SC pattern the arm underwent less directional change leading to relatively lower jerk. With context to shoulder pain and jerk, it was maintained that individuals with shoulder pain adopt a smoother arm motion pattern to reduce momentary discomfort at the shoulder during wheelchair propulsion.

In conclusion the results from the Chapter 3 led to the following generalized conclusions:

(1)     adopting jerk based quantification of wheelchair propulsion kinematics is worthwhile and yields insightful inferences;

(2)     investigating the recovery phase kinematics in the context of shoulder pain is as important as studying the push phase, and

(3)     in the future it may be beneficial to incorporate jerk based metric into rehabilitation practice.

**Shoulder pain and trunk kinematics in wheelchair propulsion**

Trunk kinematics during wheelchair propulsion is known to influence propulsion biomechanics and also bears implication for shoulder injury. The main aim of this investigation was to study the trunk kinematic differences between manual wheelchair users with and without shoulder pain. Research guidelines from maintain that propelling a manual wheelchair with a 'trunk flexed / anterior-tilt' position exposes mWCUs to secondary injury risk (Rodgers et al., 2000, Rodgers et al., 2001, Rankin et al., 2011, Gagnon et al., 2015, Gagnon et al., 2009, Sanderson DJ et al., 1985, Vanlandewijck YC et al., 1989, Chow et al., 2009, Yang et al., 2006, Rice et al., 2004). However, presently there is very limited information regarding trunk kinematics differences between groups propelling a manual wheelchair with and without shoulder pain.

Consequently, this cross-sectional study analyzed the trunk kinematics during manual wheelchair propulsion in a group of experienced adult MWCUs with and without shoulder pain. We hypothesized that the trunk kinematics of individuals propelling a manual wheelchair with shoulder pain will significantly differ from the trunk kinematics of the group without shoulder pain. The net drift in the trunk position (during stead state propulsion (~135 cycles)) from the initial reference position (rest) along the sagittal plane (X direction) was computed and the group means statistically tested to validated the postulated hypothesis.

Our investigation revealed that individuals with shoulder pain had larger net deviation trunk position in sagittal plane (trunk movement in sagittal plane) from the initial reference position than those without shoulder pain. This observed phenomenon was potentially attributed to be a compensatory mechanism to minimize discomfort to shoulder during propulsion. This attribution is consistent with recommendations from occupation ergonomics literature (Madeleine et al., 2008, Lomond et al., 2010, Lomond et al., 2011). Individuals with neck/shoulder pain performing rhythmic repetitive occupational tasks adopted a spatial strategy minimizing the movement of their trunk to minimize discomfort (Madeleine et al., 2008, Lomond et al., 2010, Lomond et al., 2011). Further, it was discussed that from a pure human movement ergonomics perspective, when the trunk was flexed forward, the arm joints grab the hand-rim with greater internal

rotation at the initiation of the start of the proceeding push phase. It is well known that application of load to arm joints with at such orientation contributes to increased risk of causing injury to joints (Bridger RS 2009).

Based on these rationale's and observed results overall it was concluded that studying trunk kinematics is related to shoulder pain and incorporating trunk kinematics based measures, may provide additional knowledge to understand the adaptive strategies employed by mWCUs with shoulder pain.

In conclusion the results from the Chapter 4 led to the following generalized conclusions:

(1)     trunk kinematics can be used to differentiate between manual wheelchair users with and without shoulder pain;

(2)     investigating the trunk kinematics over the entire propulsion cycle ( push + recovery) captures the adaptive strategy as opposed to only studying the push phase, and

(3)     including data from the entire trials (i.e. in our case the trial has ~135 cycles on average) for analysis may have facilitated the capture of adaptive dynamics, better than merely using a portion or a small fraction of data from the full trial ( i.e. for example only, 10 to 20 cycles   from a trial rather than the full trial data).

**Shoulder pain and time dependent structure in propulsion variables**

Based on the tenets of the *loss of complexity* hypothesis of aging (Lipsitz et al., 2004) the time-dependent structure of motor output has been associated with physiological complexity and shown to provide novel information concerning the health of the musculoskeletal system (Hausdroff et al., 1997, Sosnoff et al., 2006, Stregiou et al., 2011). Specifically, it has been proposed that musculoskeletal injury leads to motor fluctuations that are more structured (i.e. more regular – loss of complexity). The amount of variability is usually quantified using the distribution statistics (standard deviation, coefficient of variation). However, dynamical measures of motor variability have been found to be more sensitive to pathology than measures involving distributional statistics (Slifkin et al., 1999, Stergiou et al., 2011, Stergiou et al., 2004).

The size and time dependent structure of motor variability have been shown to provide unique information concerning the control and health of the neurophysiologic system (Lipsitz et al., 2004, Sosnoff et al., 2006, Hausdroff et al., 1997, Newell et al., 1993, Madeleine et al., 2009). Recent research indicates that the size of variability in wheelchair propulsion mechanics is related to shoulder pain (Sosnoff et al., 2015). This association between size of motor variability and pain is consistent with observations from occupation biomechanics and human motor control literature (Srinivasan et al.,2012, Madeleine et al., 2009, Hamill et al., 2012 Stergiou

et al., 2011).

Presently there is no information relating shoulder pain and variable structure in wheelchair propulsion. Research evidence from occupation ergonomics shows that the structure of variability is sensitive discomfort level arising from repetitive strain injuries (Madeleine et al., 2009). However to date, the structure in variability of wheelchair propulsion variables have not been investigated. Consequently chapter 5 investigates the *loss of complexity* hypothesis. It was hypothesized that manual wheelchair users with shoulder pain will demonstrate lower complexity compared to those without shoulder pain. Consequently the purpose of this investigation is to examine if shoulder pain and the variable structure in wheelchair propulsion are related.

In this investigation, sample entropy (SampEn), a widely utilized approach to quantify the structure of variability was used (Yentes et al., 2013, Tochigi et al., 2012). The SampEn for the cycle-to-cycle fluctuations in contact angle and inter push time interval was computed. As a first step a simple surrogate analyses was performed to establish that the structure found in these variables were not random and had some meaningful structure. Following this the group mean values of SampEn were compared between the groups with and without shoulder pain to validate the acceptance or rejection of the postulated hypotheses.

Our results revealed that the SampEn for contact angle was able to differentiate between groups with and without shoulder pain. This in agreement with occupation ergonomics research (Madeleine et al., 2009). The higher SampEn magnitude observed for the contact angle at the hand-rim in mWCUs with shoulder pain could be a manifestation of their compensatory strategies to minimize shoulder discomfort when performing the repetitive propulsion task. Consistent with previous research where higher discomfort (pain) levels were associated with higher spatial complexity (Madeleine et al., 2009), SampEn of contact angle was positively correlated with the self-reported pain scores.

The SampEn for time interval was not able to discriminate between groups with and without shoulder pain. A possible reason for this could be the disruption in temporal structure of the time interval due to the visual feedback on propulsion speed (Hausdroff et al., 1996).

Overall in conclusion, this investigation evinced three novel observations:

(1) variability observed in the fluctuation in contract angle and time interval during manual wheelchair propulsion is structured;

(2) individuals with shoulder pain exhibited higher SampEn magnitude during wheelchair propulsion compared to those without pain, and

(3) SampEn measure correlated significantly with the amount of self-

reported shoulder pain.

This observation is consistent with reports from occupational ergonomics and motor control research that variability of motor output may offer meaningful insights concerning health and function (Madeleine et al., 2008, Madeleine et al., 2009, Vaillancourt et al.,2002, Sosnoff et al., 2015, Vieluf et al., 2015, Srinivasan et al., 2015). Overall in conclusion, this investigation evinced that incorporating non-linear dynamics based measures in wheelchair propulsion analyses may provide new knowledge and can be an important tool for better health, diagnosis, and therapeutic interventions to prevent shoulder pain pathology in mWCUs.

**Limitations**

Since all the studies were from the same data set, the major limitations are common for all the three studies (Chapters 3, 4 and 5). A major limitation of this dissertation is the cross sectional nature of the data. The data were collected in a laboratory setting with roller dynamometer setup, so it is not clear if these results would occur in real life propulsion environment. The injury demographics of our sample was diverse. Consequently, it is possible that the between group differences observed for the biomarkers were due to different disability demographics. Another limitation is that, the pain scores were self-reported and no radiographic or ultrasonic information on pain were collected. Though significant, these

demographic and environmental limitations are relatively common to wheelchair propulsion research. Finally, despite these limitations, this investigation provides novel contributions that are important and compliments the research findings from previous research (Sosnoff et al., 2015, Madeleine at al., 2009, Srinivasan et al., 2012, Madeleine et al., 2009). All these findings should be repeated in larger samples to check repeatability of result trends.

**Future directions**

In this section, first, some future research directions/improvements for each of the studies (Chapters 3, 4, and 5) will be proposed. Following this, in conclusion, a future research framework to address the overall limitation identified is being proposed.

**Shoulder pain and jerk during wheelchair propulsion.**

- **Improvement to method**: One of the methodological limitation faced in the jerk analysis was the constraint to group participants by pattern type. To overcome this limitation a dimensionless version of jerk metric is proposed. Such dimensionless approach will enable comparison collapsing across recovery pattern types.

- **Addressing open research question in wheelchair propulsion literature**: *A main open question that still lies unanswered in wheelchair propulsion literature is, how does the human motor system*

*choose a particular type of recovery pattern type given the environment, physical state and task constraints?*

One way to seek answer to this open question is to verify if individuals choose propulsion pattern based on cost criteria which is encompasses a jerk minimization criteria. For this a pilot study with able bodied individuals is proposed. Individuals without any prior experience with manual wheelchair can be trained to use all the four propulsion pattern and on reaching certain level of propulsion skill, can be asked to propel with various task constraints (ramp, different floor types, different speed etc) and the self chosen pattern kinematics for these conditions can be analyzed to see if individuals minimizes jerk to choose the pattern.

**Shoulder pain and jerk during wheelchair propulsion.**

- **<u>Improvement to method</u>**: As a methodological improvement, a study of the acceleration profile of the trunk is proposed. Research evidence from occupational ergonomics, show that the trunk acceleration metrics and adaptive responses to neck/shoulder pain are related (Madeleine et al., 2008). On the same lines, a jerk based analysis of trunk kinematics is also warranted.

- **Addressing open research question in wheelchair propulsion literature**: *One unexplored research question in wheelchair propulsion literature is, how does constraining or un-constraining the trunk affect wheelchair propulsion biomechanics/performance?*

  To answer this, a study with able bodied novice individuals is proposed. The participants can be trained to propel wheelchair in one specific/different pattern. Once they reach a certain skill level, the participants can propel the wheelchair on a treadmill with different task constraints (speed and different ramp), (1) trunk unconstrained, (2) trunk partially constrained to backrest and (3) trunk fully constrained to backrest. Results from such study will provide novel knowledge to understand what strategy the neuromuscular system adapts given the task constraint.

**Shoulder pain and structure in variability during wheelchair propulsion.**

- **Improvement to method**: One main limitation of the methodology for the SampEn investigation was the small data record length. Our work is the first to apply a non-linear dynamics approach to wheelchair propulsion. There is no wheelchair propulsion data specific information on guidelines to choose the sample entropy parameters (embedding dimension (m), tolerance (r)). A proposed improvement to the chapter 5 in terms of methodology will be to collect wheelchair

propulsion data with for longer duration (6 to 8 minutes) to be able to study the influence of m and r on wheelchair propulsion data.  Longer data length will also enable complimenting the SampEn results with other non-linear dynamic methods like Lyapunov exponent (LyE) and de-trended fluctuation analysis (DFA). In general, adopting a combination of non-linear methods is recommended as best practice (Stergiou 2004). A second methodological improvement will be to remove the visual feedback for speed to study the time dependent structure in the time interval. In the current procedure in Chapter 5, the visual feedback may have disrupted the temporal structure for the time interval variable.

- **<u>Addressing open research question in wheelchair propulsion literature</u>**: *It is an open question to study the relationship between SampEn and propulsion variables at the shoulder. Presently, the Chapter 5 only evaluates, the SampEn for propulsion variables at hand-rim. It is not clear if the same trend in SampEn will exist at the site of pain (shoulder). This is an open area for future enquiry.*

**A wearable technology for longitudinal data collection**

As was stated in the limitation section, one of the main limitations with wheelchair propulsion research in general is the very limited access to longitudinal data and propulsion data outside lab environment. The currently

available instrument wheels (SMARTWheel and OptiPush) are very expensive/ bulky and not viable options for individualized home based longitudinal data collection. To overcome this limitation the dissertation proposes a 3 to 5 year longitudinal study tracking novice manual wheelchair users longitudinally using an improved version of the wearable technology prototyped in chapter 6. The wearable technology can also be used to continuously track the identified biomarkers to validate if these variables can track/aid in preventing the progression of shoulder pain n manual wheelchair users.

# Chapter 8

## Intellectual contributions to wheelchair propulsion research

To study shoulder pain in manual wheelchair users, the traditional approach for the past two decades focused only on the mean wheelchair propulsion parameters (biomechanics approach) and improvements to wheelchair build (wheelchair ergonomics). Given that the association between shoulder pain and manual wheelchair propulsion is multi-faceted in nature, a multi-disciplinary approach to analyze and address this problem is more appropriate. Such multi-disciplinary approaches can provide better understanding of the pathology and may lead to new knowledge for better monitoring/tracking/ and prevention of shoulder injury in manual wheelchair users.

This dissertation integrated approaches and theories from human motor control, occupation ergonomics and non-linear synamics to investigate shoulder pain in manual wheelchair users. The dissertation research was successful in identifying biomarkers that relate to shoulder pain using the multi-disciplinary approach proposed. Futher, the dissertation also prototype tested a custom developed wearable technology that has the potentital to translate the research findings from the dissertation for the benefit of manual wheelchair users.

This dissertation made the following intellectual contributions to wheelchair propulsion research:

- *identified three novel metrics* that relate shoulder pain and manual wheelchair propulsion;

- *opened new avenues for future research* in wheelchair propulsion research;

- proposed methods using a multi-disciplinary approach that has *potential to addreess open questions* in wheelchair propulsion literature;

- established that *investiagating the recovery phase during wheelchair propulsion* yields novel and useful inference to understand adaptive stragtegies during wheelchair propulsion;

- validated the prototype of an affordable custom developed *wearable device* with the potential to translate the research findings to acual wheelchair users.