Evaluation of Risk to the Lumbar Spine and Shoulders During Simulated Wheelchair Pushing

Thesis

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Abstract

Health care workers and at-home caregivers have a high prevalence of occupational back and shoulder injuries related to attendant-propelled, or manual, wheelchair pushing. Wheelchair design affects posture, comfort, and biomechanical load for attendants. Thus, the objective of this study was to determine how simulated manual wheelchair pushing influences biomechanical loading to the lumbar spine and shoulders.

62 subjects performed simulated wheelchair pushing and turning in a laboratory. An overhead rig allowed for the setup of a simulated standard manual wheelchair. Linear or rotational resistance provided by an overhead braking system increased incrementally during each pushing or turning trial such that subjects ended each trial with a maximum voluntary exertion (MVE). A dynamic, electromyography-assisted biomechanical model was used to estimate spinal loads; moments at the shoulder joint, measures of external hand force, and net torque were also assessed. Finally, multiple linear regression techniques were employed to develop biomechanically-based wheelchair pushing guidelines relating resultant hand force or net torque to spinal load.

The highest compressive spinal loads were found at the L3/L4 Inferior endplate, while peak lateral and anterior/posterior (A/P) shear loads were found at the L5/S1 Superior and L5/S1 Inferior endplates, respectively. Compressive and A/P shear spinal loads were significantly higher for male subjects ($p<0.01$) and were increased for wheelchair turning compared to straight wheelchair pushing ($p<0.001$). Peak shoulder moments were 62.7±18.0 Nm for males and 35.2±8.7 Nm for females. Maximum applied handle force during straight wheelchair pushing was 295±72 N for males and 206±53 N for females, while maximum torque applied for turns was 70±20 Nm and 51±19 Nm. For straight wheelchair push exertions, subjects applied hand force at a mean angle of 39.8±15.6 degrees relative to horizontal. Results from multiple linear regression calculated biomechanically-determined maximum acceptable resultant hand forces during wheelchair pushing. Biomechanically-determined values were 17-18% lower than the closest psychophysically determined comparisons.

Attendant-propelled wheelchair pushing poses biomechanical risk to the lumbar spine in compression and A/P shear and the shoulders, particularly for males. Current psychophysically-determined maximum acceptable push forces are not protective enough of this biomechanical risk; rather, the biomechanically-based wheelchair pushing guidelines presented in this study should instead be implemented. Additionally, there is opportunity for improved manual wheelchair design, particularly in regard to wheelchair handle height; higher handle heights might result in a more horizontally applied hand force, which would decrease rolling resistance in manual wheelchair pushing. Finally, the results of the study should be considered with its limitations. The study population was young and inexperienced in patient handling, and subjects were not guaranteed to give a true MVE. Spinal loads were also interpreted relative to compression and shear thresholds that are neither sex nor age dependent, and shoulder moments were compared to maximum strength capabilities generated under controlled experimental conditions.

Dedicated to my family and friends

Acknowledgments

I wish to thank my advisor, Dr. William Marras, for inviting me to learn and contribute to a lab environment in which I have already learned so much and am excited to continue to learn in the future. Thank you for challenging me in ways I had never before been challenged, yet always being there to offer your help when it was needed. Thank you, too, for teaching me to begin to view research in terms of the big picture.

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Pollard, J.P., Porter, W., Mayton, A., Xu, X., and Weston, E.B. The effect of vibration exposure during haul truck operation on grip strength, touch sensation, and balance. Int. J. Ind. Ergonomics, 2017. **57**: p. 23-31.

Fields of Study

Major Field: Industrial and Systems Engineering

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Chapter 1: Introduction

1.1 Work-Related Musculoskeletal Disorders

1.1.1 Prevalence of Low Back and Shoulder Musculoskeletal Disorders

Musculoskeletal disorders (MSDs) present a major socioeconomic problem in modern society, particularly in modern occupational environments. Among all MSDs, back problems are particularly disabling. With an 80% prevalence within the general population (Manchikanti et al. 2009), low back pain (LBP) is the most crippling condition affecting man worldwide (Hoy et al. 2014).

It is well recognized that occupational exposures contribute to risk of developing MSDs in the low back and shoulder (NRC/IOM 2001). Work-related MSDs required a median of 12 days to recuperate before returning to work in 2015 (BLS 2016). Moreover, it is common for individuals to report pain in more than one site.

1.1.2 Cost of Treatment

Nationally, the direct cost of LBP, workers' compensation, and time lost from work due to LBP totals over $50 billion annually. Some estimates including indirect cost place this statistic well over $100 billion (Davis et al. 2012). Likewise, though less costly, the direct costs of treating shoulder pain are still estimated at $7 billion annually (Meislin et al. 2005).

1.2 Patient Care

1.2.1 Musculoskeletal Disorders in Healthcare

The risk of low back and shoulder MSDs among health care professionals including nurses and nurses' assistants is particularly high. LBP prevalence among nurses has been noted to reach as high as 82% after just five years of employment (Videman et al. 2005). Moreover, nursing assistants were among the occupations with the highest number of MSD cases resulting in days away from work in 2015, second only to laborers and heavy/ tractor-trailer truck drivers (BLS 2016).

Patient care activities are thought to be a frequent factor contributing to MSDs in health care workers. Manually lifting and transferring patients has been shown to result in excessive compressive loads to the spine (Garg et al. 1991a; Garg et al. 1991b; Winkelmolen et al. 1994; Marras et al. 1999), but interventions such as floor-based and ceiling-based mechanical lifts have been developed to help mitigate this biomechanical risk (Brophy et al. 2001; Evanoff et al. 2003; Engst et al. 2005). It has also been recognized that pushing and pulling patients, as performed in sling placement, pushing mechanical lifts, and wheelchair pushing, also poses biomechanical risk (van der Woude et al. 1995; Marras et al. 2009; Nagavarapu et al. 2016). However, interventions to help mitigate musculoskeletal risk from pushing and pulling patients are ill-defined.

1.2.2 Musculoskeletal Disorders in Home Caregivers

Another population at risk of musculoskeletal injury from patient handling tasks includes unpaid at-home caretakers. At-home caretakers are often untrained and regularly lift and transfer their dependent in addition to pushing their dependents in wheelchairs indoors and outdoors across a variety of terrains (Suzuki et al. 2015). Most often, these attendants are elderly themselves (Kaye et al. 2000).

Roberts et al. (2012) gathered questionnaire data about at-home wheelchair users and determined the average wheelchair user to be 57 years old. Of wheelchair users, 56% were overweight or obese, and 61% of these users were totally dependent. Back problems predominated among caregivers, with 72% reporting low back pain. Shoulder pain was also an issue, with 48% of caregivers reporting pain in this region. 35% of at-home caregivers pushed their dependent in a wheelchair at least four times per day.

1.3 Research Voids Surrounding Wheelchair Pushing

Attendant-propelled, or manual, wheelchair pushing is a patient handling task common to both of the aforementioned populations and is thus of particular interest. Roberts et al. (2012) showed that caretakers often push loads in excess of 100 kg during manual wheelchair handling. The design of the wheelchair being pushed has been shown to affect the posture, comfort, and biomechanical loads placed onto the attendant (Abel and Frank 1991), yet there is not significant evidence justifying the push handle design on standard manual wheelchairs.

Only one study has estimated biomechanical risk of manual wheelchair pushing in terms of shoulder moments and spinal loads (van der Woude et al. 1995). This particular study employed inverse dynamics techniques to estimate moments at the shoulder joint and a two-dimensional linked-segment model to estimate spinal loads for eight females that performed manual wheelchair pushing tasks on level ground and up an incline. Results showed low compressive and shear spinal loads at L5/S1 (1052 and 93 N, respectively) relative to accepted damage thresholds for spinal loading (NIOSH 1981; Gallagher and Marras 2012) when pushing a wheelchair with a total weight of 105 kg. However, the results of this study should be interpreted relative to two major limitations. In particular, this study used a 75 kg (165 pound) patient for their exertions. Patients are expected to weigh more today considering that more than one-third of U.S. adults are considered to be obese (Odgen et al. 2015). Additionally, this study lacks exploration of the role of muscle coactivation along the lumbar spine, especially considering that wheelchair pushing is a dynamic patient handling task. In wheelchair pushing, trunk flexors in particular must activate to compensate for an extension moment created by the reaction force at the hands (van der Woude et al 1995).

1.4 Objective

Given the high prevalence of low back and shoulder injuries in healthcare and at-home care and voids in understanding of the biomechanics of wheelchair pushing, the objective of this study was to determine how the current standard manual wheelchair design influences biomechanical loading to the low back and shoulders.

Chapter 2: Methods

2.1 Approach

A laboratory study was conducted in an attempt to understand biomechanical measures of spinal load, shoulder moment, and hand force during simulated wheelchair pushing and turning. Hand forces were calculated via load cell, while a biologically-driven, electromyography (EMG)-assisted spine model was implemented to evaluate joint moments and three-dimensional spinal loads (compression, A/P shear, and lateral shear).

2.2 Subjects

Sixty-two subjects (31 male, 31 female) inexperienced in patient handling were tested in this study. Subjects were recruited from University and local community populations. The ages of the subjects ranged from 20 to 54 years, with additional anthropometric details provided in Table 1. All subjects were asymptomatic for LBP in the past three years and had no prior history of back surgery. Additionally, all of the subjects provided informed consent, and the study was approved by the University's institutional review board.

Table 1 - Anthropometric data of subjects (mean ± standard deviation)

	Overall (62)	**Male (31)**	**Female (31)**
Age (years)	26.1 ± 6.3	25.6 ± 4.7	26.6 ± 7.6
Height (cm)	173.9 ± 9.7	179.8 ± 8.3	167.9 ± 7.1
Weight (kg)	73.0 ± 14.7	79.1 ± 13.1	66.9 ± 13.8

2.3 Experimental Design

A repeated measures design was implemented. The two experimental conditions, including a straight wheelchair push and a turning wheelchair push, were counter-balanced such that half of the subjects performed the straight push first and half of the subjects performed the turning push first. Subjects performed three repetitions of each trial type back-to-back. An overhead braking system (Figure 1) incrementally added linear or rotational resistance as each trial progressed such that each trial ended with a maximum voluntary exertion (MVE).

2.3.1 Independent Variables

Independent variables included individual factors of sex, age, height, and weight, push direction including straight or turn, and individual factor * push direction interactions. During the secondary analysis (as described in Secondary Analysis and Multiple Linear Regression), resultant hand force and net torque were also treated as predictor variables for the straight push and turning push exertions, respectively.

2.3.2 Dependent Variables

Dependent measures consisted of three-dimensional hand forces recorded at the hands, joint moments calculated at each shoulder, and spinal loads in compression, A/P shear, and lateral shear at each spinal level extending from T12/L1 to L5/S1 as predicted by an EMG-assisted biomechanical model.

2.4 Apparatus and EMG-Assisted Biomechanical Model

2.4.1 Overhead Rig and Braking System

A custom-built overhead rail system and rig were used in this study (Columbus McKinnon, Amherst, NY, USA); this rail system allowed for two-dimensional linear translation and rotation. Additionally, a magnetic particle braking system was implemented within the rail system (Placid Industries, Inc., Lake Placid, NY, USA). Resistance provided by the brakes was controlled via voltage output from custom-built Matlab software. The overhead rig and braking system may be observed in detail in Figure 1.

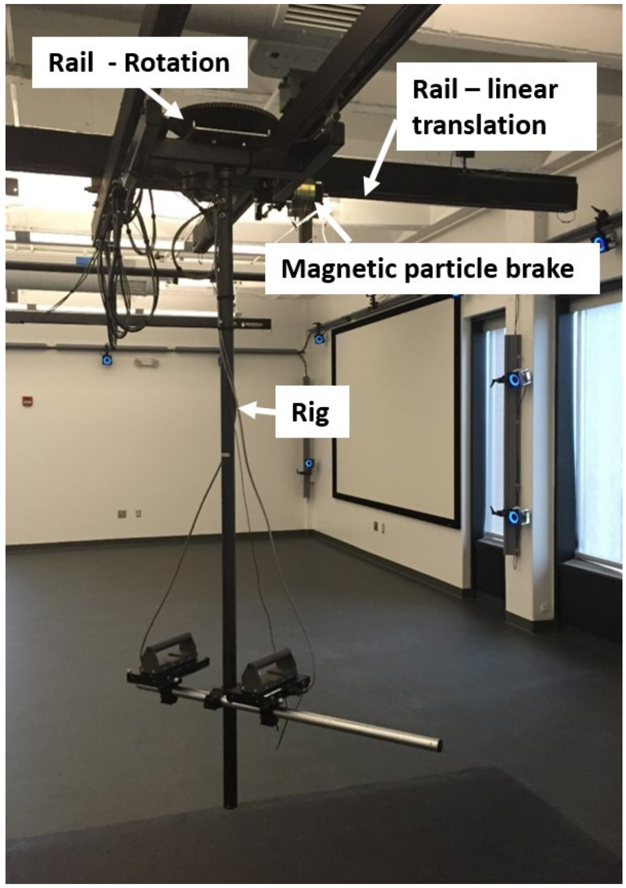

Figure 1 – Overhead braking system and magnetic particle brakes

2.4.2 Dynamic Model Inputs

Kinematic data were captured via a 36 infrared camera OptiTrack motion capture system (NaturalPoint, Corvallis, OR, USA) at a sampling frequency of 100 Hz; these data were low-pass filtered using a fourth-order Butterworth filter with a cutoff frequency of 10 Hz. Kinetic data were captured via a force plate (FP6090-15, Bertec, Worthington, OH, USA) and two custom-built hand transducers outfitted with handles to measure three-dimensional bilateral application of hand force (HT0825, Bertec, Worthington, OH, USA). All three load cells had six degrees of freedom, measuring tri-axial force and moment. Kinetic data were collected at a sampling frequency of 1000 Hz. Finally, EMG data were collected using bipolar surface electrodes placed bilaterally onto the erector spinae (ES), latissimus dorsi (LD), rectus abdominis (RA), external oblique (EO), and internal oblique (IO) muscles with an inter-electrode distance of 3 cm. The data were sampled at 1000 Hz, notch filtered at 60Hz and its aliases, and band-pass filtered at 30-450 Hz; then, the data were rectified, smoothed using a moving average filter, normalized, and finally, low-pass filtered using a second-order Butterworth filter and time constant of 100 ms.

2.4.3 Biomechanical Model

The EMG-assisted biomechanical model implemented in this study was developed in the Biodynamics Laboratory at The Ohio State University and is backed by over twenty-five years of peer reviewed research (Marras and Sommerich 1991a; Marras and Sommerich 1991b; Granata and Marras 1993; Granata and Marras 1995; Knapik and Marras 2009; Dufour et al. 2013; Hwang et al. 2016). This dynamic model combines kinematic, kinetic, and EMG data to estimate muscle force in the ten power-producing muscles of the trunk, moments at each spinal level from T12/L1 to L5/S1, and spinal loads. Unlike other spine models that are static or use a single equivalent trunk muscle, the biomechanical model used in this investigation accounts for neuromuscular activation patterns and muscle coactivity in addition to individual and spine geometries including: segment length, width, depth, and mass *and* vertebrae and IVD height, depth, and width. Finally, the model takes into account the following features influencing muscle force production: cross sectional area, origins and insertions, length-strength relationship, force-velocity relationship, and curved muscle geometry.

2.5 Procedure

Subjects were briefed on the study design and agreed to informed consent as per University institutional review board guidelines. Anthropometric measures were collected, and surface electrodes were placed on the aforementioned power-producing muscles of the trunk via standard placement procedures (Mirka and Marras 1993). Forty-one motion capture markers were also placed onto the head, torso, and upper and lower extremities. Finally, subjects performed exertions on a force plate according to a "no-max" calibration procedure as described by Dufour et al. (2013).

The hand transducers were set to a standard wheelchair handle configuration including: handle height of 0.9144 m (36 in.), handle width of 0.4572 m (18 in.), and backward-facing handles, as shown in Figure 2 (ADA 1990). Before each exertion, the braking system was zeroed relative to the global position of the rig in the motion capture space such that each trial began at the same (zero) level of resistance. Subjects performed each pushing exertion at a normal pace, and the overhead braking system incrementally increased the linear or rotational resistance based on changes from the initial global position during each trial. Subjects continued with the push or turn until they could no longer translate or rotate the rig and were told to exert statically at some maximum hand force or torque for an additional 1-2 seconds before relaxing.

Prior studies have shown that moments on the spine are low during pushing tasks due to the external moment created via the trunk counteracting the external moment contribution from reaction forces at the hands (van der Woude et al. 1995). In order to increase moment exposure onto the spine, subjects were instructed to perform each exertion in an upright posture (minimal trunk flexion). Maximizing external moments onto the spine represent a “worst-case” scenario in terms of biomechanical load.

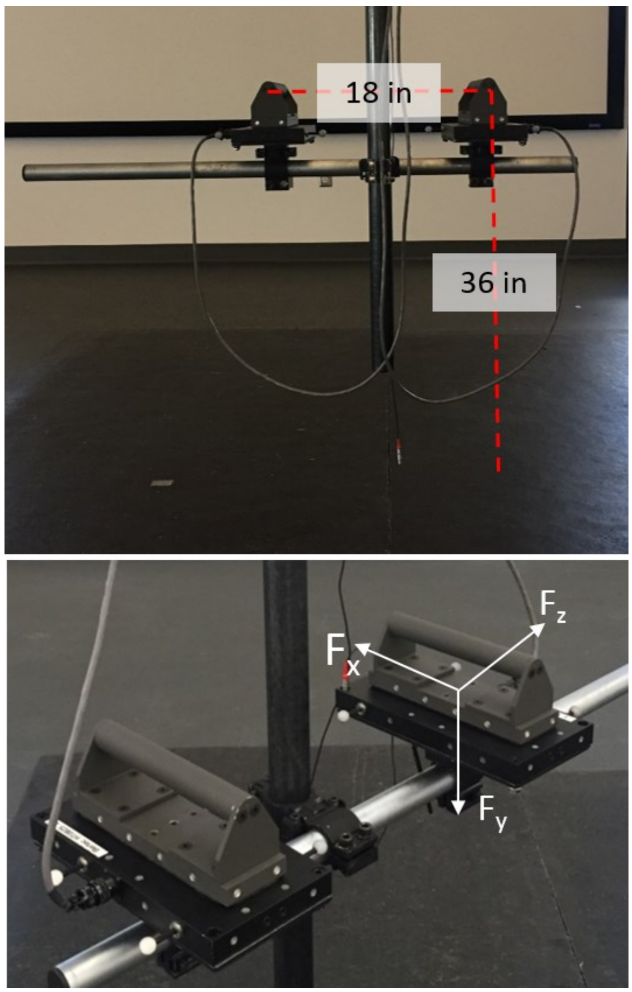

Figure 2 – Experimental setup of hand transducers

2.6 Analysis

2.6.1 Model Fit and Spinal Loads

Peak spinal loads were calculated at each spinal level in compression, A/P shear, and lateral shear for each trial. The endplate level with the highest peak loads served as the basis for subsequent biomechanical analysis. Spinal loads were interpreted relative to damage thresholds for spinal loading, those being 3400 N of compression (NIOSH 1981) and 700 N of shear (A/P or lateral) (Gallagher and Marras 2012). Spinal loads exceeding these thresholds are hypothesized to cause endplate micro-fractures that have been shown to lead to the development of scar tissue that disrupts nutrition to the intervertebral disc (Gallagher et al 2006).

2.6.2 Shoulder Moment

The peak magnitude of right and left resultant shoulder moments from each trial were assessed via Adams software (MSC Software, Santa Ana, CA, USA) in the same model simulations that estimated spinal load for each trial. Upper extremity segment mass and external hand forces applied at the hand transducers were the major effectors of shoulder moment in the simulation. Peak moments were compared to strength capabilities published within the literature.

2.6.3 Resultant Hand Force

For straight wheelchair pushes, hand forces produced by the right and left hands were summed, and hand force was represented as a single three-dimensional vector. The mean angle of force application relative to horizontal (Figure 3) was also calculated for each trial.

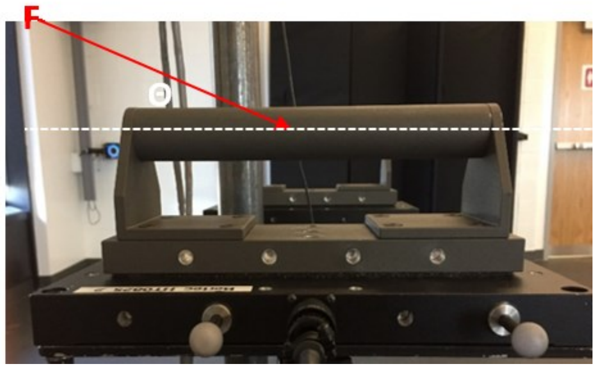

Figure 3 – Angle of force application relative to horizontal

2.6.4 Torque

Net torque was calculated for turning wheelchair pushes rather than resultant hand force. Measured moment arms were 0.2286 m (9 in) in the global Y direction and 0.08255 m (3.25 in) in the global X direction. Net torque was calculated via addition of torque moment contributors from both hands according to Equation 1, with force components and moment arms detailed via Figure 4.

$$Torque = F_{1X} * r_Z + F_{1Z} * r_X - F_{2X} * r_Z + F_{2Z} * r_X \quad (1)$$

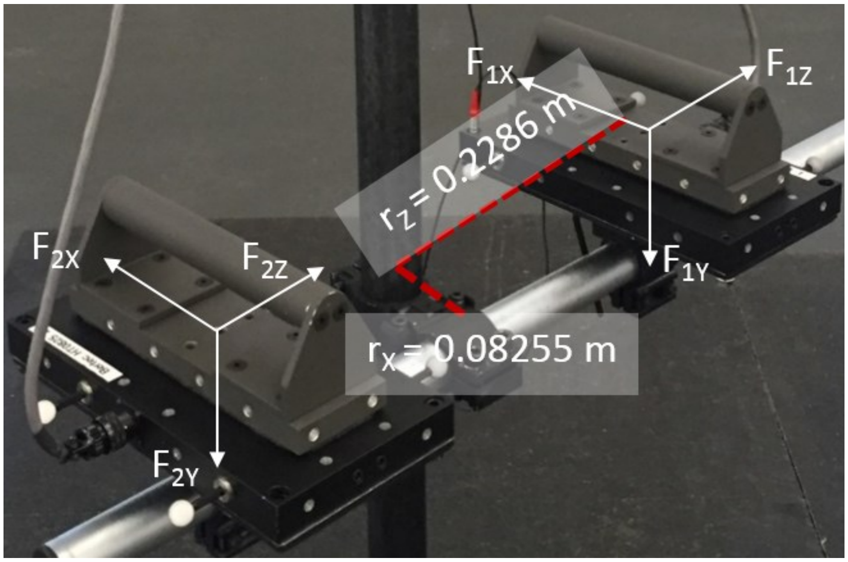

Figure 4 – Vectors used for calculation of net torque in wheelchair turning

2.6.5 *Statistical Analysis*

Post-processed spinal load, shoulder moment, resultant hand force, and hand torque data were analyzed using JMP 11.0 (SAS Institute Inc., Cary, NC, USA) with a one-way repeated measures ANOVA for independent measures of push direction and individual factors. Data were interpreted relative to a significance level $\alpha = 0.05$. The effects of the independent measures on spinal load were also interpreted relative to biological significance in which only differences between means for spinal loading of 100 N or more were assumed to reach significance.

2.6.6 *Secondary Analysis and Multiple Linear Regression*

In order to created biomechanically-based thresholds reliant on either hand force or net turning torque, it was important to relate resultant hand force (straight wheelchair pushes) and net turning torque (turning wheelchair pushes) to dependent measures of spinal load. Thus, in this phase of the study, four separate multiple linear regression models were implemented. Each multiple linear regression model used individual factors (age, sex, height, weight) and whichever variable was appropriate between resultant hand force or net torque as potential predictor variables. Four independent samples from each trial were obtained, measuring the hand force or torque corresponding to: 25%, 50%, 75%, and 100% of the range between the minimum and peak spinal loads for the trial, as shown in Figure 5. All models were fit via multiple linear regression with a stepwise backward elimination method. Outliers were excluded, and model fit was assessed via Adjusted R^2.

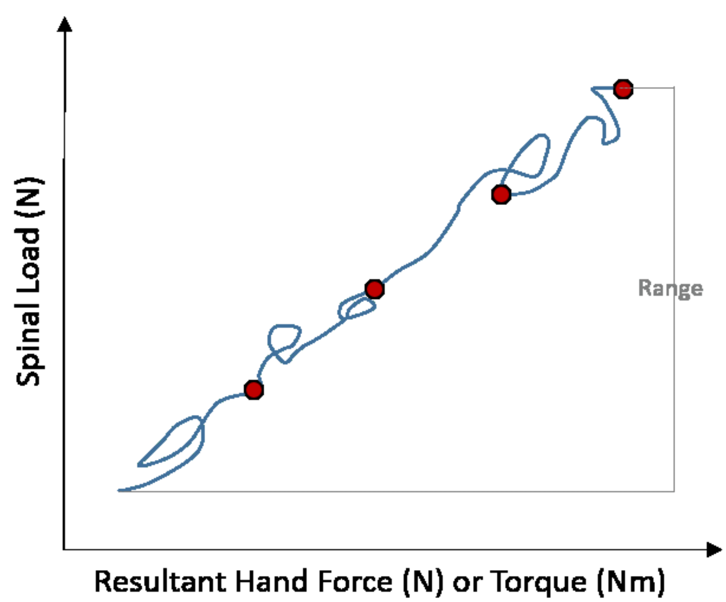

Figure 5 – Multiple linear regression sampling technique for spinal load data.

Data from all trials were included in the in multiple linear regression analysis

2.6.7 Biomechanically-Determined Pushing Thresholds

Multiple linear regression models were also used to determine maximum acceptable hand forces or torques corresponding to action limits for spinal loading (3400 N compression and 700 N A/P shear). A point estimate of the hand force or torque correlating with the action limit for spinal loading was determined; at this level of hand force or torque, 50 percent of the population is assumed to encounter biomechanical risk. Hand force or torque values causing other percentages of the population (90%, 75%, 25%, 10%) to cross the action limit were determined using the standard normal distribution of each model's residuals. Ultimately, the thresholds that were more protective between those determined for spinal compression and those determined for A/P shear were reported.

The values for straight wheelchair pushing were also compared to psychophysically determined thresholds from Snook and Ciriello (1991). The closest comparison from the psychophysically determined threshold tables was a 2.1 m (initial) push for males at a frequency of one push every five minutes and a handle height of 0.94 m. It is important to note that Snook and Ciriello (1991) reported horizontal hand force only whereas this study reports resultant hand force; additionally, Snook and Ciriello (1991) investigated horizontal handles, whereas this study investigated backward facing handles as seen on manual push wheelchairs.

Chapter 3: Results

3.1 General Results

Statistically and biologically significant differences for the dependent measures investigated are shown in Table 2. In general, the direction of the exertion (straight wheelchair push vs. turning wheelchair push) was shown to significantly influence dependent measures. Although various individual factors were investigated (age, sex, mass, stature), sex was the only individual factor that consistently influenced dependent measures. Additionally, there were no instances in which a significant push direction * individual factor interaction was observed.

Table 2 - Statistically and biologically significant results

Dependent Measure	Push Direction	Sex	Age	Stature	Mass	Interactions
Spinal Load						
Compression	***	**				
A/P Shear	***	***			X	
Lateral Shear	XXX	XXX		XX	XXX	
Shoulder Moment		***				
Hand Force						
Resultant (Straight)	N/A	***				
Net Torque (Turns)	N/A	*				

(*$p < 0.05$, **$p < 0.01$, ***$p<0.001$, ^{X} results lacking biological significance)

3.2 Spinal Loads

3.2.1 Compression

The highest values for spinal compression were found at the L3/L4 Inferior endplate level. Figure 6 shows the mean and standard deviations for peak L3/L4 Inferior compression for subjects grouped by sex and push direction. Compressive spinal load was increased for males relative to females ($p<0.01$) and increased by 13.9% for wheelchair turning trials relative to straight wheelchair pushes ($p<0.001$). Males crossed the 3400 N damage threshold for spinal compression in 34.4% of the straight wheelchair pushes and 45.6% of the wheelchair turning trials; females rarely crossed this same threshold (1.0% and 4.3% of trials for straight pushes and turns, respectively).

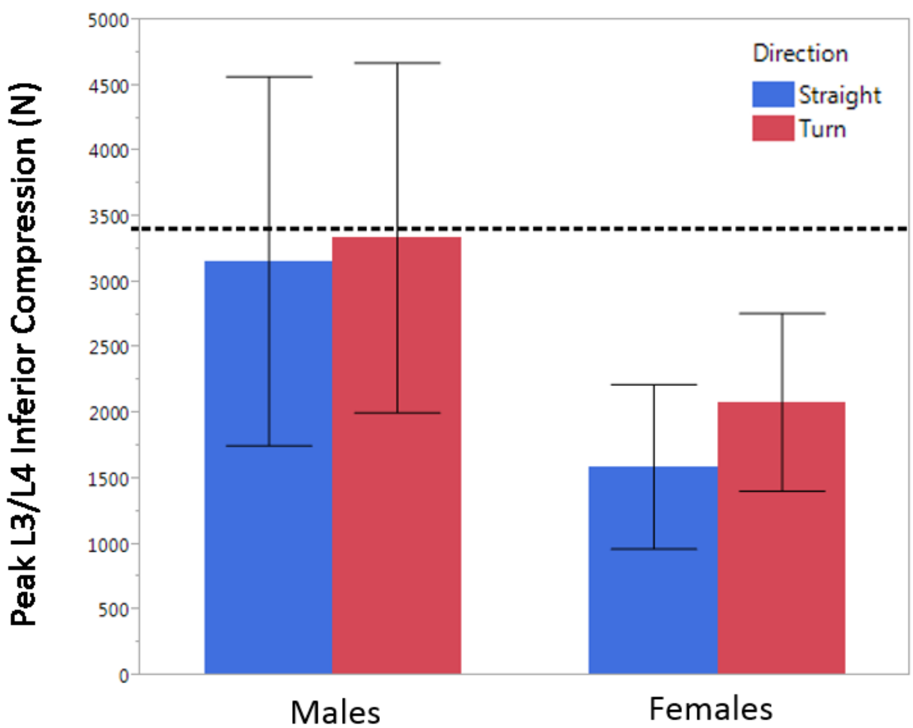

Figure 6 – Peak compressive spinal load by sex and push direction

3.2.2 A/P Shear

The highest peak A/P shear values were noted at the L5/S1 Inferior endplate level. Consistent with compressive spinal load results, males had significantly higher peak A/P shear loads than females ($p<0.001$), and turning wheelchair pushes had 39.1% higher peak A/P shear loads than straight wheelchair pushing ($p<0.001$); these results are detailed further in Figure 7. In addition, peak A/P shear spinal load increased with increased body mass ($p=0.039$), though the effect was not biologically significant.

Males crossed the 700 N damage threshold for A/P shear spinal load often, in 28% of straight wheelchair pushing trials and in 63% of wheelchair turns. Of particular note, the mean peak L5/S1 Inferior A/P shear value for males during wheelchair turning trials (789 N) exceeded the damage threshold of 700 N for A/P shear spinal loading; moreover, under these conditions, 27.2% of the turning trials also crossed the upper damage threshold of 1000 N A/P shear load. Females did not ever cross the damage threshold for A/P shear in straight wheelchair pushing but did cross this threshold in 11.8% of the turning exertions.

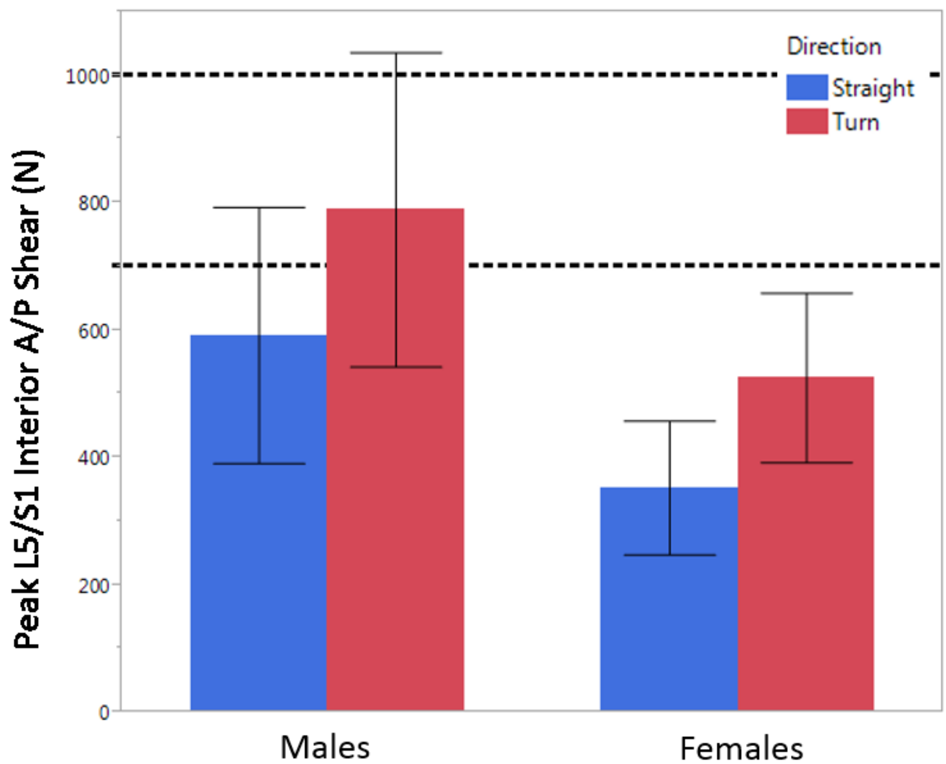

Figure 7 – Peak A/P shear spinal load by sex and push direction

3.2.3 Lateral Shear

Peak lateral shear values were noted at the L5/S1 Superior endplate for straight wheelchair pushes and wheelchair turns. All lateral shear values were determined to be below the accepted damage threshold value for shear, and none of the statistically significant results reached biological significance. Thus, spinal loading in terms of lateral shear was determined to be low risk and not investigated any further.

3.3 Shoulder Moments

Peak shoulder moment magnitudes are shown in Table 3. Shoulder moments were increased for males in both the left and right shoulder ($p<0.0001$). There was no statistically significant effect of push direction on right shoulder moment magnitude, though the magnitude of the left shoulder moment was reduced for wheelchair turning relative to straight wheelchair pushing ($p<0.0001$).

As shown in Table 4, peak shoulder moments often approached or exceeded shoulder moments within the distribution of shoulder moments generated from maximal isometric push exertions recorded in the literature (Chow and Dickerson 2016). Net shoulder moments for males crossed mean (50^{th} percentile) reported isometric strength (74.9 Nm net shoulder moment) in 31.3% of straight wheelchair pushes and 28.2% of turning wheelchair exertions. On the other hand, females rarely surpassed reported mean (50^{th} percentile) isometric strength (61.2 Nm net shoulder moment), crossing this strength value just 1.1% of all trials.

Table 3 - Peak shoulder moment magnitudes for straight pushing and turning

Sex	**Exertion**	**Left Shoulder Moment (Nm)**	**Right Shoulder Moment (Nm)**
Male	Straight Push	61.1 ± 18.3	62.0 ± 18.4
	Turning Push	48.5 ± 12.9	62.7 ± 18.0
Female	Straight Push	33.6 ± 7.7	35.2 ± 8.7
	Turning Push	27.7 ± 6.8	34.3 ± 10.1

Table 4 - Maximum Acceptable Shoulder Moment from Isometric Exertion

Sex	Percentile of Strength Distribution	Resultant Shoulder Moment (Nm) from Chow and Dickerson (2016)	% of Population in Present Investigation Exceeding Value
Male	10	40.3	90.8%
	25	56.7	62.2%
	50	74.9	29.7%
	75	93.1	8.1%
	90	109.5	0.5%
Female	10	40.6	22.6%
	25	50.3	8.6%
	50	61.2	1.1%
	75	72.1	0%
	90	81.8	0%

3.4 Maximum Hand Force and Torque

3.4.1 Resultant Hand Force in Straight Wheelchair Push

Maximum resultant hand forces produced by males and females during straight wheelchair pushing are shown in Figure 8. Males produced increased resultant hand force relative to female subjects ($p=0.0004$). The direction of force application for straight wheelchair pushing was 39.8±15.6 degrees.

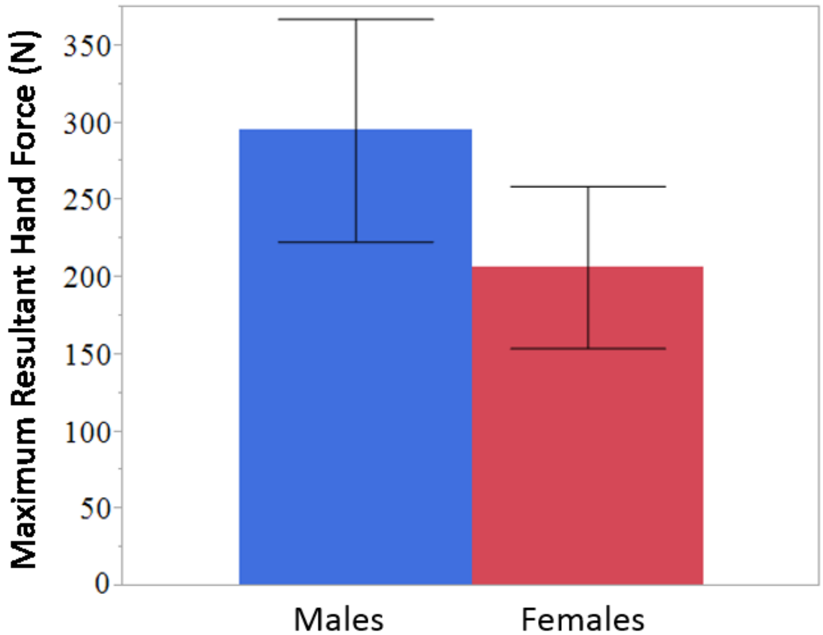

Figure 8 – Max resultant hand forces produced in straight wheelchair pushing

3.4.2 Net Torque in Wheelchair Turning

Maximum net (axial) turning torque produced during wheelchair turning trials are shown in Figure 9. Similar to straight pushing, males produced increased torque relative to females ($p = 0.0135$), but no other individual factors reached significance. Mean direction of force application in the left hand was -27.9.8±17.9 degrees and -18.0±19.4 degrees in the right hand, denoting that unlike in straight wheelchair pushing, subjects pulled upward on the handles while turning, and more so in the left hand than the right hand.

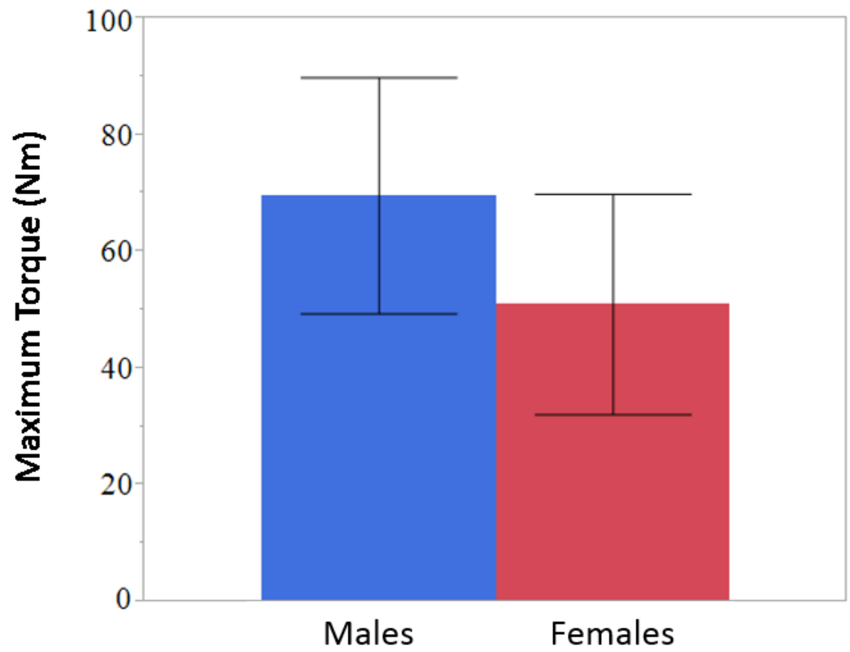

Figure 9 – Max torque produced during wheelchair turning

3.5 Secondary Analysis and Multiple Linear Regression

Four multiple linear regression analyses were run to establish a relationship between resultant hand force (straight wheelchair push) or net torque (wheelchair turn) and dependent measures of compressive and A/P shear spinal loads. Subtractive model selection methods denoted that the best multiple linear regression models should contain resultant hand force/ net torque and sex in the model (binomial: 1 if female, 0 if male). Model parameters and fit are evaluated in Table 5. Adjusted R^2 values for the models ranged between 0.825 and 0.874, denoting good fit.

3.6 Biomechanically-Determined Pushing Thresholds

Biomechanically-based maximum acceptable resultant hand force or torque thresholds were determined based on model outputs for male subjects because males crossed damage thresholds for spinal loading at lower hand force and net torque values than females. The Hand Force vs. L3/L4 Inf. Compression model was used to determine maximum acceptable resultant hand force for straight wheelchair pushing, while the Torque vs. L5/S1 Inf. A/P Shear model was used to determine maximum acceptable torque for wheelchair turning. Maximum acceptable forces and torques protecting various percentages of the population are shown in Table 6 and Table 7, respectively. The biomechanically-based hand force limits from this study were 17-18% lower than the closest psychophysically-determined comparison from Snook and Ciriello (1991).

Table 5 - Parameters and fit of multiple linear regression analyses

Direction	**Predictor Variable vs. Dependent Measure**	**Parameter Estimates** $\widehat{\boldsymbol{Spinal\ Load}} = \boldsymbol{b_0} + \boldsymbol{b_1}(\boldsymbol{Hand\ Force}) + \boldsymbol{b_2}(\boldsymbol{Female})$	$\mathbf{R^2_{Adj}}$	**Threshold Cross Estimate**
Straight	Hand Force vs. L3/L4 Inf. Compression	$\widehat{Comp} = 434 + 8.15(HF) - 285(Female)$	0.825	**363 N** @3400 N Compression
	Hand Force vs. L5/S1 Inf. A/P Shear	$\widehat{Shear} = 163 + 1.48(HF) - 39.6(Female)$	0.845	**364 N** @700 N A/P Shear
Turn	Torque vs. L3/L4 Inf. Compression	$\widehat{Comp} = 534 + 37.4(Torque) - 244(Female)$	0.856	**76.6 Nm** @ 3400 N Compression
	Torque vs. L5/S1 Inf. A/P Shear	$\widehat{Shear} = 281 + 6.31(Torque) - 60.9(Female)$	0.874	**66.3 Nm** @700 N A/P Shear

Table 6 - Biomechanically determined maximum acceptable hand force

(straight wheelchair push)

Percent of Population Protected	Biomechanically Determined Resultant Hand Force Limit (N)	Psychophysically Determined Horizontal Hand Force Limit (N)	Percent Decrease
90	226	275	17.8%
75	291	353	17.6%
50	363	441	17.7%
25	435	530	17.9%
10	501	608	17.6%

Table 7 - Biomechanically determined maximum acceptable torque

(wheelchair turning)

Percent of Population Protected	Biomechanically Determined Torque Limit (Nm)
90	36.2
75	50.4
50	66.2
25	82.0
10	96.2

Chapter 4: Discussion

4.1 Biomechanical Loads

Often in pushing and pulling studies, the initial and sustained maximum force required to accelerate an object from rest is used to determine the limits of acceptable forces and load weights (Jung et al. 2005). Thus, for comparison sake, subjects were also asked to exert up to their MVE during each trial in this study. Biomechanical measures of spinal load, shoulder moment, and hand force were influenced mainly by wheelchair push direction and sex.

4.1.1 Spinal Loads

Whereas results from van der Woude et al. (1995) indicate that spinal loads resulting from pushing a patient of moderate weight (75 kg) are not particularly risky as compared to common damage thresholds (NIOSH 1981; Gallagher and Marras 2012), the results of this investigation show that both straight wheelchair pushing and wheelchair turning pose significant biomechanical risk in terms of spinal loading. Biomechanical models should account for coactivity of the power-producing trunk muscles, particularly activation of flexor muscles as is common during pushing exertions (Knapik and Marras 2009). Thus, differences in spinal loading between studies can be attributed to this investigation employing a 3D EMG-assisted biomechanical and the prior investigation employing a 2D linked-segment model with a single muscle equivalent (van der Woude et al. 1995).

In the present investigation, males crossed damage thresholds for spinal loading in 34% of the straight wheelchair push trials and 63% of the wheelchair turning trials recorded. Spinal loads were significantly lower for female subjects, but this effect could be due to gender differences in both mass and MVE strength capacity. Contrary to a prior investigation employing the same EMG-assisted model that showed that most problematic spinal loads during pushing are associated with A/P shear forces (Knapik and Marras 2009), results of this present investigation indicate that straight wheelchair pushing and turning exertions are associated with problematic spinal loads in terms of *both* compression and A/P shear. Additionally, whereas Knapik and Marras (2009) saw high A/P shear spinal loads at higher levels of the lumbar spine (L3 and above), the highest shear loads were seen much lower in this study (L5/S1). These spinal load differences could largely be due to modeling improvements that now include muscle lines of actions wrapped around the trunk as opposed to straight-line vectors; wrapping lines of action for trunk muscles better approximate the complex and often asymmetric lumbar motions that frequently occur in occupational environments (Hwang et al. 2016).

It appears that this study is the first to examine biomechanical measures during wheelchair turning, whereas most literature has instead chosen to investigate pushing manual wheelchairs up and down inclines (van der Woude et al. 1995; Horiuchi et al. 2014; Suzuki et al. 2015). Wheelchair turning trials saw 13.9% higher compression and 39.1% higher A/P shear loads than straight wheelchair pushing. Moreover, the mean of peak L5/S1 Inferior A/P shear loads in turning trials for males (789 N) crossed over the

700 N damage threshold for spinal loading as established by Gallagher and Marras (2012). Increased spinal loads due to wheelchair turning are speculated to result from a variety of sources. In terms of compression, the upward direction of force application at the hands during wheelchair turning actually made the exertion approximate more of a pull, which has been noted in prior push and pull studies to result in increased compressive spinal load (Hoozemans et al. 2004). In terms of A/P shear, it is expected that the axial torque required to turn the overhead rig system required increased trunk flexor activation, particularly in the external oblique muscles. The horizontal line of action of the external oblique muscles relative to the geometry of L5/S1 increases A/P shear load with increased activation (Knapik and Marras 2009).

4.1.2 Shoulder Moments and Associated Strength Capabilities

Unlike measurements of spinal loading, the shoulder joint does not have force or damage thresholds that can be applied to directly estimate biomechanical risk. In part, this is due to the fact that the shoulder encounters large ranges of motion and has significantly different strength capability based on hand position (Koski and McGill 1994). It is known, however, that the likelihood of sustaining musculoskeletal injury to the shoulder increases as forces or moments approach or exceed an individual's strength capability (Chaffin 1975; Kahn and Monod 1989). As a result, biomechanical risk to the shoulder can be derived via comparison of shoulder moments recorded in this study and related measures of isometric strength capabilities in the literature.

In a recent paper, Chow and Dickerson (2016) reported strength capabilities during standing maximum isometric exertions while also recording corresponding shoulder moments; the results from this investigation were the most comparable strength values to the experimental setup of this study that could be found. As was shown by Table 4, shoulder moments recorded in this study for males approached and often crossed shoulder moment values presented by Chow and Dickerson (2016), indicating potential biomechanical risk. On the other hand, peak shoulder moments for females in this investigation only crossed the low percentile values within the shoulder moment distribution reported by Chow and Dickerson (2016). It should be noted, however, that the maximum resultant hand forces produced by both males and females within this study were significantly less than the maximum resultant hand forces produced by subjects during standing, isometric exertion; this reduction in resultant hand force likely accounts for the reduction in shoulder moments observed, particularly for female subjects.

As peak shoulder moments recorded in this study were likely aligned to the subjects' MVE, it is also important to examine shoulder moments from other studies pushing manual wheelchairs at non-MVE with varied patient weights. The literature on this topic is scarce, but shoulder moments recorded in this investigation approximate shoulder moments recorded by Suzuki et al. (2015), who examined shoulder moment while pushing a manual wheelchair loaded with 95 kg. Pushing wheelchairs exceeding this weight, as is likely, should be expected to pose significant risk to the shoulders.

Finally, though peak shoulder moments did not vary widely between straight wheelchair pushing and wheelchair turning (at least in the right shoulder), peak shoulder moments for females in this study were noted to be approximately 57% of peak shoulder moments recorded in male subjects. This result is expected to be attributable to strength capabilities based on sex. Prior studies have shown female subjects to have between 50%

and 70% of the strength capabilities of males (Koski and McGill 1994; Lin et al. 2012; Chow and Dickerson 2016). In this investigation, females produced 70% of the maximum resultant hand force and 73.2% of the net torque of males.

4.1.3 Maximum Hand Force and Torque

No other studies have examined maximum hand force production for wheelchair pushing, nor have any studies examined maximum net torque for wheelchair turning. As a result, it is difficult to *directly* compare the magnitude of the peak resultant hand forces or net torques seen in this study with maximum strength values from prior literature. It is apparent, however, that maximum strength capabilities are reduced by under wheelchair pushing conditions by 29% or more when compared to the closest strength comparisons available within the literature (Chow and Dickerson 2016).

4.2 Biomechanically Determined Pushing Thresholds

Current pushing and pulling recommendations were determined psychophysically, relying on the assumption that subjective perception of an individual's maximum acceptable external forces corresponds to biomechanical tolerance (Snook 1978; Snook and Ciriello 1991). It is well known, however, that individuals are unable to sense biomechanical loading on the spine due to the lack of nociceptors in the IVD (Adams et al. 1996). Le et al. (2012) discovered no association between spinal load and psychophysically-determined maximum acceptable forces. Therefore, biomechanically-determined guidelines for pushing and pulling would provide value.

Multiple linear regressions run for resultant hand force vs. spinal load or net torque vs. spinal load allowed for establishment of biomechanically-based guidelines under wheelchair pushing and wheelchair turning conditions, respectively. It is apparent that current psychophysically-determined maximum acceptable push forces are not protective enough of biomechanical risk, as the biomechanically-determined maximum acceptable hand force values were 17-18% lower than the closest psychophysically-determined values reported by Snook and Ciriello (1991). It is important to note, too, that Snook and Ciriello (1991) reported horizontal hand force only whereas this study reports resultant hand forces and net torques. Additionally, Snook and Ciriello (1991) investigated horizontal handles, whereas this study investigated backward facing handles as seen on manual push wheelchairs. The biomechanically-based wheelchair pushing guidelines presented in this biomechanical study should be implemented moving forward.

As was the case for shoulder moment, few studies have related external hand forces required to push manual wheelchairs with patient weight. Thus, it is difficult to relate maximum resultant hand force or net torque estimates obtained from MVE to the expected hand forces that might be required to actually push patients of various weights. Suzuki et al. (2015) and van der Woude et al. (1995) did examine external hand forces required to push "dummy" patients weighing 80 kg and 75 kg, respectively. Accounting for the weight of the manual wheelchairs used, the total weight being pushed in both studies was approximately 185 pounds and required approximately 150 N of hand force to push; note, however, that for the Suzuki et al. (2015) horizontal hand force was measured, whereas the van der Woude et al. (1995) study measured resultant hand force. Nonetheless, the external hand forces required to push dummy patients in both prior studies would, by the biomechanically-determined guidelines presented in this study, be protective in terms of spinal load. It is important to remember, however, that many patients in the United States today are expected to weigh much more than 185 pounds; in fact, anthropometric data for the United States population collected between 2007 and 2010 estimates that 50% of adult males and approximately 25% of adult females would weigh over this amount (Fryar, Gu, and Ogden 2012). The assumed caregivers, namely health care workers and at-home caregivers, are also expected to have decreased tissue tolerance behavior with age or cumulative trauma from repetitive loading on the job (Marras 2008).

4.3 Limitations

It is vital to consider the results of this work in context with the limitations encountered. First off, the study was run under laboratory conditions. Dimensions of standard wheelchairs were imitated, but the task performed nonetheless was only a simulated wheelchair pushing and turning task. Although subjects were instructed to remain upright throughout each trial to increase moment exposure to the spine, the posture assumed by each subject during each trial was only deemed as acceptable or unacceptable via subjective assessment by the researcher responsible for data collection. It is possible that trials with extensive torso flexion (and subsequently reduced moment exposure) were included incidentally within the final data set.

During each trial, subjects were instructed to exert until reaching their MVE. It is impossible to determine, however, if true MVEs were actually obtained. Similar to collection of maximum voluntary contractions (MVC) in EMG, MVEs collected from this study were sensitive to sincerity of effort, fatigue, posture, or any pain encountered by subjects from exertion (Warwick et al. 1980; Chow and Dickerson 2016).

Participants were recruited mainly from a young college aged population (mean age 26.1 years). The small standard deviation for the population investigated in this study (6.3 years) could have caused significant age effects to be missed. Moreover, this population is not perfectly representative of populations expected in health care work (nursing or nursing aide), nor is it representative of at-home caregivers. However, it may represent the age of health care workers that are first starting at work. Finally, this same young population was also inexperienced with patient handling tasks. It is expected that a population experienced with patient handling would display slightly different neuromuscular activation patterns than the population tested within this study (Marras et al. 2006).

Spinal loads were interpreted relative to compression and shear thresholds that are neither sex nor age dependent, as spinal loading thresholds that are sex or age dependent are otherwise nonexistent. Only one prior regression analysis performed by Jager, Luttmann, and Laurig (1991) predicted the effects of individual factors such sex or age on the strength of the lumbar spine, and this research study reported decreased compressive strength for females and with age. As such, it is particularly possible that biomechanical risk was higher for females than was reported in this present investigation due to limitations of comparing spinal loading to just one threshold value.

Moreover, there remain no accepted damage thresholds for net shoulder moments within the scientific literature that can be applied to realistic pushing tasks. The only values reported within the literature are maximum strength capabilities under controlled experimental conditions. This investigation used shoulder moments resulting from standing maximum isometric push exertions to determine relative biomechanical risk placed onto the shoulder. It should be noted, however, that the experimental conditions were not directly matched; the values derived from Chow and Dickerson (2016) recorded shoulder moments during pushing exertions at a higher handle height of 100 cm (39 in) and with the use of horizontally facing handles. Strength capabilities have also been shown to be decreased for backward facing handles (Seo et al. 2010) and lower handle heights (Kumar 1995), as encountered during wheelchair pushing. Koski and McGill (1994) noted reduced maximal strength in flexion and extension during dynamic shoulder torque generation rather than static torque generation. The dynamic nature of wheelchair pushing and turning tasks performed in this study should be expected to decrease maximal strength capabilities even further.

4.4 Design Implications and Future Work

While current manual wheelchair design was determined to pose biomechanical risk to the low back and shoulders, future work could focus on mitigating biomechanical risk via improved wheelchair design. The most vital design change relates to push handle height. Abel and Frank (1991) performed manual wheelchair pushing at varied handle heights and found that preferred handle height is about 60% of stature, which corresponds to around 0.9 m (35 in) – 1.1 m (43 in); current manual wheelchair push handles are set at 0.9144 m (36 inches), at the low end of the preferred range. Direction of force application could also be more optimal at higher handle heights. As shown in this investigation, subjects are likely to lean on handles while pushing if the handles are set to a low height.

In the case of wheelchair pushing, downward force application increases the rolling resistance (i.e., frictional force between the wheels and floor), thereby even further increasing the horizontal force required to push the patient. Higher handle heights could allow for more horizontally applied force at the hands to maximize force application (Abel and Frank 1991; de Looze et al. 2000; Knapik and Marras 2009).

Other potential design changes could include: wider handles to allow for increased torque for turning, horizontal rather than backward facing push handles to allow for more neutral wrist and elbow postures and increased strength (van der Woude et al. 1995; Seo et al. 2010), and increased wheel diameter to reduce rolling resistance (Al-Eiwawi et al. 1999). Electrically powered wheelchairs may also be an option and could be tied to the biomechanically-based maximum force guidelines produced in this study; in particular, a load cell on the wheelchair handles could potentially sense when caregivers are approaching the recommended force limits and provide power to keep hand forces (and therefore spinal loads) under prescribed thresholds. Said design would be particularly helpful for at-home caregivers that often push their dependents across high friction or uneven terrain such as carpet (Suzuki et al. 2015).

The present investigation recorded external shoulder moments that were largely dependent on upper extremity segment mass and external hand forces applied at the hand transducers without consideration of the internal forces and moments created by surrounding musculature. Combining this modeling limitation with the fact that the maximum isometric strength values presented by Chow and Dickerson (2016) were not directly matched to the experimental setup seen in this investigation, it was difficult to create realistic biomechanically-based thresholds for the shoulder joint as was performed for the low back. Future work should attempt to relate net shoulder moment with external hand force or net torque and subsequently develop thresholds based on more comparable strength data. This way, one compiled set of biomechanically-based thresholds that are protective of both the low back *and* shoulders could be determined for pushing exertions.

Finally, future work should investigate if and/or how the hand force vs. spinal load relationship established in this study is affected by wheelchair pushing on an inclined surface such as a ramp. A natural “ramp” was established in this study, but only in terms of the overhead brake resistance and hand force required to keep the overhead rig moving. Walking up or down a physical ramp could change factors such as the direction of applied hand force that were missed via this particular study’s methodology.

Chapter 5: Conclusion

Collectively, the results of this study suggest that manual wheelchair pushing tasks (including straight wheelchair pushing and wheelchair turning) pose biomechanical risk to the lumbar spine and shoulder. Compressive and A/P shear spinal loads often exceeded damage thresholds, while shoulder moments approached or exceeded isometric strength capabilities reported elsewhere in the literature. Manual wheelchair design should be reconsidered, particularly in regard to push handle height.

Finally, results of this study showed that current psychophysically- determined maximum acceptable push forces are not protective enough of biomechanical risk. The biomechanically-based push thresholds proposed in this investigation are more objective and physiologically relevant than the current subjective, psychophysically-determined thresholds commonly in use and should be implemented in subsequent evaluations manual wheelchair pushing.

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