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ABSTRACT

The objective of this study was to determine how simulated manual wheelchair pushing influences biomechanical loading to the lumbar spine and shoulders. Sixty-two subjects performed simulated wheelchair pushing and turning in a laboratory. An electromyography-assisted biomechanical model was used to estimate spinal loads. Moments at the shoulder joint, external hand forces and net turning torque were also assessed. Multiple linear regression techniques were employed to develop biomechanically based wheelchair pushing guidelines relating resultant hand force or net torque to spinal load. Male subjects experienced significantly greater spinal loading ($p < 0.01$), and spine loads were also increased for wheelchair turning compared to straight wheelchair pushing ($p < 0.001$). Biomechanically determined maximum acceptable resultant hand forces were 17–18% lower than psychophysically determined limits. We conclude that manual wheelchair pushing and turning can pose biomechanical risk to the lumbar spine and shoulders. Psychophysically determined maximum acceptable push forces do not appear to be protective enough of this biomechanical risk.

Practitioner Summary: This laboratory study investigated biomechanical risk to the low back and shoulders during simulated wheelchair pushing. Manual wheelchair pushing posed biomechanical risk to the lumbar spine (in compression and A/P shear) and to the shoulders. Biomechanically determined wheelchair pushing thresholds are presented and are more protective than the closest psychophysically determined equivalents.

1. Introduction

The risk of low back and shoulder musculoskeletal disorders among health care professionals including nurses and nurses' assistants is particularly high (Videman et al. 2005; BLS 2016). While manually lifting and transferring patients has been shown to result in excessive compressive loads to the spine (Garg et al. 1991a, 1991b; Winkelmolen, Landeweerd, and Drost 1994; Marras et al. 1999), interventions such as floor-based and ceiling-based mechanical lifts have been developed to help mitigate this biomechanical risk (Brophy, Achimore, and Moore-Dawson 2001; Evanoff et al. 2003; Engst et al. 2005). It has been recognised, however, that pushing and pulling patients also poses biomechanical risk. Patient handling interventions designed to mitigate musculoskeletal risk by replacing lifting with pushing or pulling may still subject caregivers to spine forces past that which would be considered safe (Marras, Knapik, and Ferguson 2009; Nagavarapu, Lavender, and Marras 2016).

Attendant-propelled, or manual, wheelchair pushing is a patient handling task 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 no significant evidence justifying the push handle design on standard manual wheelchairs. Only one study has estimated biomechanical risk related to the low back and shoulder for manual wheelchair pushing (van der Woude et al. 1995). However, this study lacked exploration of the role of muscle co-activation in the power-producing torso muscles, as is expected for dynamic push exertions (Knapik and Marras 2009).

Additionally, current accepted 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 unlikely to sense biomechanical loading on the spine due to the sparseness of nociceptors in the intervertebral disc (IVD)(Adams, McNally, and Dolan 1996). Le et al. (2012) found no association between spinal load and psychophysically determined maximum acceptable forces. Therefore, biomechanically determined guidelines for patient handling tasks related to pushing and pulling would be valuable.

Given the high prevalence of low back and shoulder injuries and voids in understanding of the biomechanics of wheelchair pushing, the primary objective of this study was to determine how the current standard manual wheelchair design influences biomechanical loading to the low back and shoulders. Since prior literature supports the theory that there is little association between spinal load and psychophysically determined maximum acceptable forces (Jorgensen et al. 1999; Davis, Jorgensen, and Marras 2000; Le et al. 2012), the secondary objective of this study was to develop biomechanically based wheelchair pushing guidelines that can be implemented in the field. It was hypothesised that subjects will encounter high biomechanical loads to the lumbar spine and shoulders during wheelchair pushing and that biomechanically based pushing guidelines will be more protective of biomechanical risk than comparable psychophysically derived guidelines.

2. Methods

2.1. Approach

A laboratory study was conducted in an attempt to understand biomechanical spine loads, shoulder moments and hand forces during simulated wheelchair pushing and turning. Throughout the investigation, the linear resistance provided by an overhead braking system linearly increased to document how the dependent measures change throughout a range of exposures. Hand forces were measured via load cell, while a biologically driven, electromyography (EMG)-assisted spine model was implemented to evaluate joint moments and three-dimensional spinal loads in compression, anterior/posterior (A/P) shear and lateral shear. The EMG-assisted biomechanical model implemented in this study uses inputs of subject-specific anthropometry, MRI-derived muscle locations and sizes (Jorgensen et al. 2001; Marras et al. 2001), full body kinematics, kinetics, muscle activity for 10 power-producing muscles of the torso and tissue material properties as model inputs to predict dynamic model outputs of muscle forces and tissue loads. It has been described extensively in the literature and is validated by over 35 years of peer-reviewed research (Marras and Sommerich 1991a, 1991b; Granata and Marras 1993, 1995; Knapik and Marras

2009; Dufour, Marras, and Knapik 2013; Hwang et al. 2016a, 2016b).

2.2. Subjects

Sixty-two subjects (31 male, 31 female) inexperienced in patient handling were recruited from university and local community populations for this study. This sample size was found to be sufficient to detect a moderate effect size in variables of interest with a power of 0.85 and significance level ($α$) = 0.05. 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 and other musculoskeletal injury such as shoulder pain in the past three years. Subjects provided informed consent, and the study was approved by the university's institutional review board.

2.3. Experimental design

A repeated measures design was implemented. Two experimental conditions, including a straight wheelchair push and a turning wheelchair push, were tested. The order in which the subjects experienced the two experimental conditions were counter-balanced, and 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

The independent variable tested was wheelchair push direction (straight or turn). Resultant hand force and net torque were also treated as predictor variables in multiple linear regression for the straight push and turning push trials, respectively.

2.3.2. Dependent variables

Dependent measures consisted of the magnitude and direction of three-dimensional forces recorded at the hands, joint moments calculated at each shoulder and spinal loads (compression, A/P shear and lateral shear) at the superior and inferior endplates of each spinal level extending from T12/L1 to L5/S1.

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

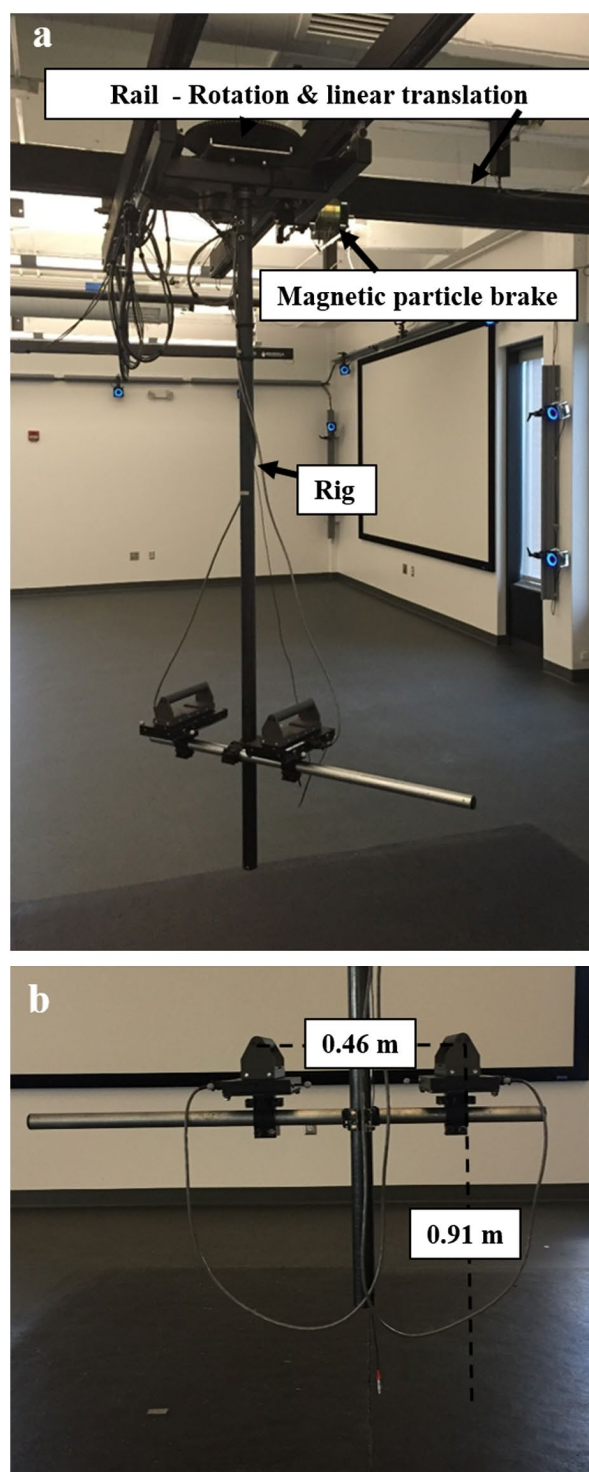|	Overall (62)	Male (31)	Female (31)
Age (years)	26.1 ± 6.3	25.6 ± 4.7	26.6 ± 7.6
Height (cm)	173.9 ± 9.7	179.8 ± 8.3	167.9 ± 7.1
Weight (kg)	73.0 ± 14.7	79.1 ± 13.1	66.9 ± 13.8

Figure 1. Experimental set-up. (a) Overhead braking system and magnetic particle brakes; (b) Hand transducers.

2.4. Apparatus

As shown in Figure 1(a), 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. 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.

Kinematic data were captured via a Prime 41 OptiTrack motion capture system (NaturalPoint, Corvallis, OR, USA) at a sampling frequency of 120 Hz; these data were low-pass filtered using a fourth-order Butterworth filter with a cut-off 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). Kinetic data were collected at a sampling frequency of 1000 Hz. Finally, EMG data were collected with a Motion Lab Systems MA300-XIV (Motion Lab Systems, Baton Rouge, LA, USA) 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 60 Hz and its aliases and band-pass filtered at 30–450 Hz; then, the data were rectified, smoothed using a moving average filter and normalised. Kinematic, kinetic and EMG data were synchronised using a data acquisition board (USB-6225, National Instruments, Austin, TX, USA) and custom laboratory software.

2.5. Procedure

Subjects were briefed on the study design and agreed to informed consent. 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 placed onto the head, torso and upper and lower extremities consistent with the standard placement locations specific to OptiTrack's motion capture software. Finally, subjects performed a series of concentric and eccentric lumbar motions in the sagittal and lateral planes while holding a 9.07 kg weight, consistent with a 'no-max' calibration procedure described by Dufour, Marras, and Knapik (2013); this calibration procedure eliminated the need to collect maximum voluntary contractions (MVCs) for EMG normalisation.

The hand transducers were set to a standard wheelchair handle configuration including: handle height of 0.916 m, handle width of 0.46 m and backward-facing handles, as shown in Figure 1(b) (ADA 1990). The handles were centred on the rig for all exertions. 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 without 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. Before any experimental conditions were collected, subjects were given the opportunity to practice the few exertions (submaximal) so as to become comfortable with the overhead rail and braking system. In straight pushing, subjects walked while pushing the rig an approximate distance of 3.2 m, though this distance was dependent on subject strength capability. In turning, subjects rotated the rig counter-clockwise in place, leading (pushing) with the right hand and trailing (pulling) with the left hand. During collection of the experimental conditions, subjects continued with the push or turn until they could no longer translate or rotate the rig and were told to exert statically at their MVE for 1–2 s before relaxing. Subjects were given 1–2-min rest between each trial and additional rest if they felt it necessary.

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 the 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). The posture assumed by each subject during each trial was deemed as acceptable or unacceptable via careful assessment by the researcher responsible for data collection, and unacceptable repetitions were repeated. Maximising external moments onto the spine represented a 'worst-case' scenario in terms of biomechanical load.

2.6. Analysis

2.6.1. 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).

2.6.2. Shoulder moment

The peak magnitude of right and left resultant external 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. External moment calculations were driven by external hand forces and upper extremity segment mass and inertia. Peak moments were compared to strength capabilities published within the literature (Chow and Dickerson 2016).

2.6.3. Hand force and torque

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. Net turning torque was reported for wheelchair turning rather than resultant hand force; it was calculated via summation of each of the torque-generating hand forces multiplied by their respective moment arms as measured from the rig. The mean angle of force application relative to horizontal was also calculated for each trial as the arctangent of the vertical and horizontal components of the hand forces produced by subjects.

2.6.4. Statistical analysis

Post-processed spinal load, shoulder moment, resultant hand force and hand torque data were analysed using JMP 11.0 (SAS Institute Inc., Cary, NC, USA) with a one-way repeated measures ANOVA for the independent measure of push direction, with individual factors of sex, height and weight included as potential covariates. All 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 (operationally defined based upon the resolution of the EMG-driven spine model employed) in which only differences between means for spinal loading of 100 N or more were assumed to reach significance.

Resultant hand force (straight wheelchair pushes) or net turning torque (turning wheelchair pushes) were also related to dependent measures of spinal load. Multiple linear regression models were implemented using 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. 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.

2.6.5. Biomechanically determined pushing thresholds

A point estimate of the hand force or torque correlating with the action limits for spinal loading (3400N compression or 700N A/P shear) was determined from each multiple linear regression model output; at this level of hand force or torque, 50% of our subjects were assumed to exceed spinal load action limits, that being the point at which biomechanical risk may first be encountered. (Figure 2(a)). Hand force or torque values causing other
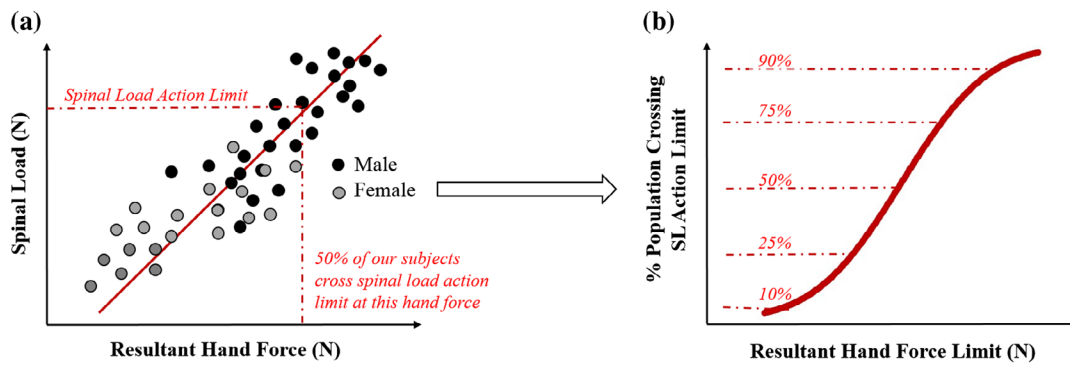
Figure 2. Biomechanically determined pushing threshold calculation. (a) Independent samples were collected for each trial and fit via multiple linear regression. A point estimate of the hand force or torque corresponding to the action limit for either spinal compression or shear was determined from the regression model. At this level of hand force or torque, 50% of our population was assumed to cross the action limit for spinal loading, denoting that they might begin to encounter biomechanical risk. (b) The standard normal distribution of strength data then estimated hand force or torque limits that were protective to various other percentages of the population.

Table 2. Statistically and biologically significant results.

Dependent measure	Push direction	Sex	Stature	Mass	Interactions
Spinal load					
Compression	***	**			
A/P shear	***	***		*	
Lateral shear	***	***	**	***	
Shoulder moment		***			
Hand force					
Resultant (straight)	N/A	***			
Net torque (turns)	N/A	*			

※Shaded results lack biological significance.
*$p < 0.05$; **$p < 0.01$; ***$p < 0.001$.

percentages of our population (90, 75, 25 and 10%) to cross the action limits were determined using the standard normal distribution of hand force strength data (Figure 2(b)). 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 values was a 2.1-m (initial) push for males at a frequency of one push every 5 min and a handle height of 0.94 m.

No studies have related external hand forces required to push manual wheelchairs with varied patient weights. Thus, as a follow-up to the determination of biomechanically determined pushing and turning limits, a pilot was conducted in which hand forces and torques required to push or turn a dummy patient (weight 86 kg) in a wheelchair on a flat tile surface were recorded using a hand dynamometer. The wheelchair (weight 23.1 kg, 50.9 lb) and dummy patient were incrementally loaded with weight up to a total patient weight of 227 kg (500 lb). Hand forces and torques required to initiate motion of the chair from a standstill were recorded, and a simple

linear regression analysis was employed to correlate patient weight to hand force or torque.

3. 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 versus turning wheelchair push) was shown to significantly influence dependent measures. Although various individual factors were investigated (sex, mass, stature), sex was the only individual factor that consistently influenced dependent measures. There were no instances in which a significant push direction * individual factor interaction was observed.

3.1. Spinal loads

Peak values for spinal compression were found at the L3/L4 Inferior endplate. These values ranged between 648 and 7252 N across all subjects, with a mean peak compressive force of 2544 ± 1295 (SD) N across all trials. Likewise, peak values for A/P shear values were noted at the L5/S1 Inferior endplate level. These values ranged between 194 and
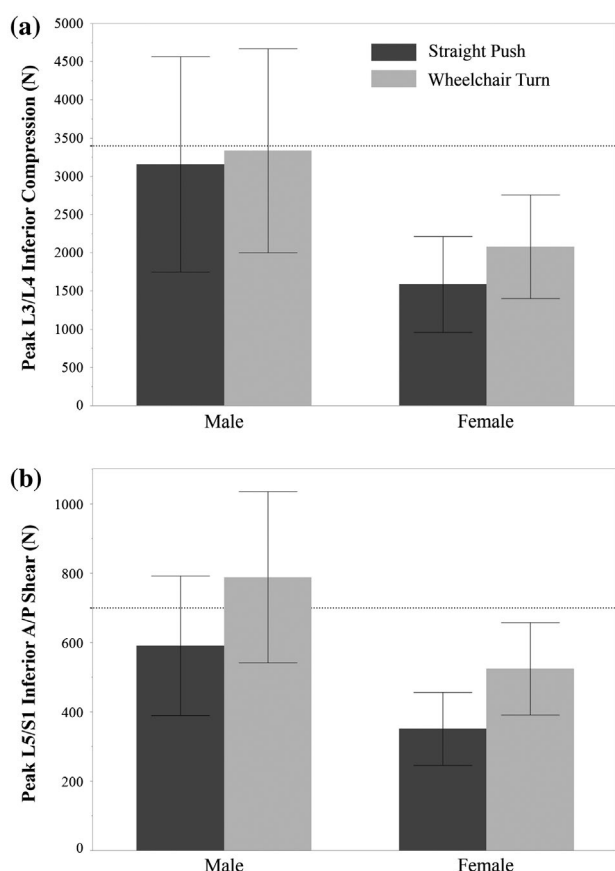
Figure 3. Peak (a) compressive spinal load and (b) A/P shear spinal load by sex and push direction. Error bars denote standard deviation.

Table 3. Peak shoulder moment magnitudes for straight pushing and turning (mean ± standard deviation).

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

1273 N, with a mean peak A/P shear force of 564 ± 238 N. Finally, peak lateral shear values were noted at the L5/S1 Superior endplate. Lateral shear values ranged between 44 and 513 N. Because all lateral shear values were determined to be below the accepted damage threshold value for shear spinal loading, lateral shear was determined to be low risk and was not investigated any further.

Figure 3(a) 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 by 13.9% for wheelchair turning trials relative to straight wheelchair pushes ($p < 0.001$). Compressive spinal load was also increased for males relative to females ($p < 0.01$). While the mean values for

peak spinal compression did not exceed 3400 N, many of the individual exertions did exceed this level. This was true particularly within the group of male subjects, in which peak spinal compression surpassed the 3400 N damage threshold in 34.4% of the straight wheelchair pushes and 45.6% of the wheelchair turning trials.

Consistent with compressive spinal load results, wheelchair turning had 39.1% higher peak A/P shear loads than straight wheelchair pushing ($p < 0.001$), and males had significantly higher peak A/P shear loads than females ($p < 0.001$) (Figure 3(b)). In addition, peak A/P shear spinal load increased with increased body mass ($p = 0.039$), though the effect was not biologically significant. Of particular note, the mean peak L5/S1 Inferior A/P shear value for males during wheelchair turning trials (790 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 1000N A/P shear load set forth by Gallagher and Marras (2012).

Males crossed the 700N damage threshold for A/P shear spinal load often, in 28% of straight wheelchair pushing trials and in 63% of wheelchair turns. 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.

3.2. Shoulder moments

Mean peak shoulder moments ranged between 48.5 and 62.7 Nm for males and 27.7 and 35.2 Nm for females (Table 3). 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 ($p < 0.0001$). Likewise, peak net moments in the right shoulder were higher than peak net moments in the left shoulder for turning ($p < 0.0001$) but not straight wheelchair pushing ($p = 0.58$). Shoulder moments were increased for males compared to females in both shoulders for both experimental conditions of straight wheelchair pushing and wheelchair turning ($p < 0.0001$).

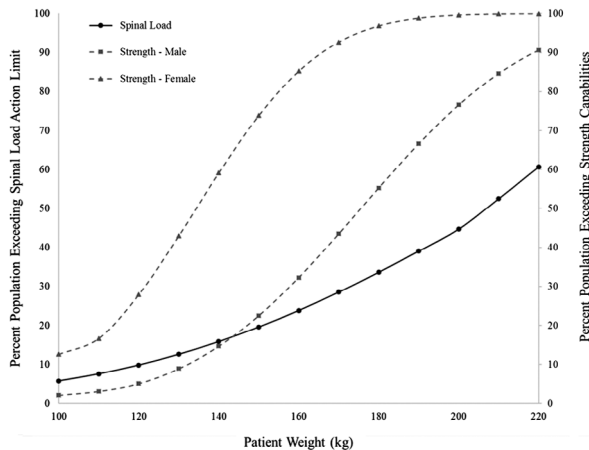## 3.3. Maximum hand force and torque

Males produced increased resultant hand force relative to female subjects ($p = 0.0004$), where maximum resultant hand forces produced by males and females during straight wheelchair pushing were 295 ± 72 (SD) N and 206 ± 53 N, respectively. The direction of force application for straight wheelchair pushing was −39.8 ± 15.6 degrees

Table 4. Parameters and fit of multiple linear regression analyses.

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

Table 5. Biomechanically determined maximum acceptable hand force.

| Percent of population protected | Straight wheelchair pushing | | | Wheelchair turning |
	Biomechanically determined resultant hand force limit (N)	Psychophysically determined equivalent limit (N)	Per cent difference	Biomechanically determined turning torque limit (Nm)
90	226	275	17.8	36.2
75	291	353	17.6	50.4
50	363	441	17.7	66.2
25	435	530	17.9	82.0
10	501	608	17.6	96.2

Figure 4. Injury risk pertaining to pushing patients of weights up to 227 kg. The biomechanical risk curve for spinal loading represents whichever estimate was more conservative between a straight wheelchair push and wheelchair turning. Superimposed onto this curve is strength data gathered based upon the MVEs collected in this investigation for male and female populations.

relative to horizontal, denoting that subjects pushed down on the handles while walking.

Similar to straight pushing, males produced increased torque relative to females ($p = 0.0135$); males produced 70 ± 20 Nm of torque for wheelchair turning trials, while females produced 51 ± 19 Nm. No other individual factors reached significance. Mean direction of force application relative to horizontal in the left hand was $27.9.8 \pm 17.9$ 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.

3.4. Biomechanically determined pushing thresholds

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 4. Adjusted R^2 values for the models ranged between 0.825 and 0.874.

Biomechanically based maximum acceptable resultant hand force or torque thresholds were determined based on model outputs for male subjects because males crossed action limits for spinal loading at lower hand force and net torque values than females. The Hand Force versus L3/L4 Inf. Compression model was used to determine maximum acceptable resultant hand force for straight wheelchair pushing, while the Torque versus L5/S1 Inf. A/P Shear model was used to determine maximum acceptable torque for wheelchair turning. The maximum acceptable forces and turning torques that resulted are shown in Table 5; the table presents multiple risk limits, each correlating to the particular percentage of the working populltion that is expected to be protected from biomechanical risk if the risk limit were set at that particular level. The biomechanically based hand force limits from this study were 17–18% lower than the closest psychophysically determined comparison from Snook and Ciriello (1991).

The pilot investigation correlating patient weight with the resultant hand force or turning torque required to initiate motion in a wheelchair was accompanied by good

model fit ($R^2 = 0.96$ for straight pushing and $R^2 = 0.86$ for turning). These models were used to show the percentage of the population tested that is expected to encounter biomechanical risk to the low back or exceed strength capacity in terms of patient weight rather than hand force or turning torque, as is shown in Figure 4.

4. Discussion

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, Haight, and Freivalds 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.

Only one study has estimated biomechanical risk of manual wheelchair pushing in terms of spinal load (van der Woude et al. 1995). This particular study estimated spinal loads for eight females pushing a wheelchair with a total weight of 105 kg 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). This investigation showed 75% higher compressive spinal loads and shear forces over three times the magnitude of the values reported by van der Woude et al. (1995) for its female subjects alone, with even higher spinal loads observed for males. Differences in spinal loading between studies can be attributed to this investigation employing a three-dimensional EMG-assisted biomechanical and the prior investigation employing a two-dimensional linked-segment model with a single muscle equivalent (van der Woude et al. 1995). 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).

Spinal loads exceeding action limits for spinal loading are hypothesised to cause endplate micro-fractures that have been shown to lead to the development of scar tissue that disrupts nutrition to the IVD (Gallagher et al. 2006). In the present investigation, both straight wheelchair pushing and wheelchair turning posed biomechanical risk in terms of spinal load. 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 strength capacity. Contrary to a prior investigation employing the same EMG-assisted model that showed that most problematic spinal loads during pushing are

associated solely 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 modelling 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. 2016a).

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. 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 a prior study 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).

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; Lin, McGorry, and Chang 2012). 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. 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.

No other studies have examined maximum hand force production for wheelchair pushing, nor have any studies examined maximum net turning 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 under wheelchair pushing conditions by 29% or more when compared to the closest strength comparisons available within the literature (Chow and Dickerson 2016).

While current pushing and pulling recommendations were determined psychophysically, Le et al. (2012) discovered no association between spinal load and psychophysically determined maximum acceptable forces. It is apparent via the results from this investigation that current psychophysically determined maximum acceptable push forces are not protective enough of biomechanical risk, as biomechanically determined maximum acceptable hand force values were 17–18% lower than the closest psychophysically determined values reported by Snook and Ciriello (1991). Snook and Ciriello (1991) also presented results from only 22 subjects (10 male, 12 female), whereas this study collected data on a more representative population of 62 subjects. 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.

Finally, the forces required to push patients of various weights are expected to be variable based upon other environmental conditions such as the floor type or wheelchair design; as such, hand force and turning torque were used when creating the biomechanically determined guidelines presented herein because these measures can be consistently measured across all patient handling environments. Still, it was helpful to relate measures collected for simulated wheelchair exertions (as was the main focus of this study) to real-life wheelchair pushing and turning situations so as to place the results of this study in context. This step was achieved in this study via determining relationships between hand force and turning torque and patient weight in a small follow-up pilot study.

As shown by Figure 4, manual wheelchair pushing at heavy patient weights is risky to both males and females, but for different reasons. Heavy patient weights above 100 kg may not leave females susceptible to risk from high spinal loads, but the force required to push or turn patients of these weights quickly surpasses female strength capability. On the other hand, males are capable of exerting more force. Even at high patient weights, a high proportion of males can still perform the task, leaving them

susceptible to imposing high risk to the lumbar spine. It is important to note that while these patient weights may seem extreme, trends suggest that the prevalence of clinically severe or morbid obesity (BMI over 40 or 50) is increasing quickly, with more rapid rates of growth in the higher weight groups (Sturm and Hattori 2013). Data for 2013–2014 showed the prevalence of extreme obesity (BMI over 40) in US adults age 20–74 to be 10.5% (Fryar, Carroll, and Ogden 2016).

4.1. Limitations

It is vital to consider the results of this work in context with its limitations. Participants were recruited mainly from a young college-aged population (mean age 26.1 years) inexperienced in patient handling. While this population is not perfectly representative of populations expected in health care work, it may represent health care workers first starting on the job.

The study was also run under laboratory conditions. Dimensions of standard wheelchairs were imitated, but the task performed was nonetheless simulated wheelchair pushing with an overhead rig and braking system. Similar to collection of MVCs 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). As a result, it was impossible to determine if true MVEs were actually obtained. Additionally, because trials were screened by the researcher responsible for data collection to ensure subjects remained upright throughout each exertion, it is possible that trials with extensive torso flexion (and subsequently reduced moment exposure) were included incidentally within the final data-set.

It is expected that allowing subjects to lean while pushing and turning would have decreased moments placed onto the lumbar spine and thereby decreased spinal loads. An upright posture was chosen for this investigation so that wheelchair pushing and turning risk limits could be created for the 'worst-case' scenario; thus, risk limits presented are protective of any self-selected posture by workers in industry. However, practitioners should be careful when generalising some of the other results of this investigation (peak spinal loads, shoulder moments) because these values come from extreme postures and were likely encountered at subjects' MVE.

During data analysis, spinal loads were interpreted relative to compression and shear thresholds that are neither sex- nor age-dependent. Jager, Luttmann, and Laurig (1991) previously reported decreased compressive strength in the lumbar spine for females relative to males and with increased age. As such, it is 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 tissue tolerance value.

Finally, there remain no accepted damage thresholds for net shoulder moments within the scientific literature that can be applied to realistic pushing tasks. This investigation used shoulder moments resulting from standing maximum isometric push exertions presented by Chow and Dickerson (2016) 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, Armstrong, and Young 2010), lower handle heights (Kumar 1995), and for dynamic exertions (Koski and McGill 1994).

4.2. Future work

Because current manual wheelchair design could pose biomechanical risk to the low back and shoulders, future work could focus on mitigating biomechanical risk via improved wheelchair design. If possible, the most helpful design intervention would be motor assists installed within bariatric wheelchairs that could be utilised while pushing heavy patients. At the least, the most feasible 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 0.9–1.1 m; current manual wheelchair push handles are set at 0.914 m, at the low end of the preferred range. 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, 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 maximise force application (Abel and Frank 1991; De Looze et al. 2000; Knapik and Marras 2009).

Because maximum isometric strength values presented within the literature were not directly matched to the experimental set-up 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 also 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.

Future work should also investigate how the hand force versus 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 incline could change factors such as the direction of applied hand force that were missed in this investigation.

5. Conclusion

Collectively, the results of this study suggest that manual wheelchair pushing tasks (including straight wheelchair pushing and wheelchair turning) can 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 suggest 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 biomechanically relevant than the current subjective, psychophysically determined thresholds commonly in use and should be implemented in subsequent evaluations of manual wheelchair pushing.

Disclosure statement

The authors report no financial interest or benefit derived from this work.

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