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ARTICLE



One versus two-handed lifting and lowering: lumbar spine loads and recommended one-handed limits protecting the lower back

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ABSTRACT

The objectives of this study were to quantify loads imposed upon the lumbar spine while lifting/lowering with one versus two hands and to create guidelines for one-handed lifting/lowering that are protective of the lower back. Thirty subjects (15 male, 15 female) performed one- and two-handed exertions in a laboratory, lifting from/lowering to 18 lift origins/destinations using medicine balls of varying masses. An electromyography-assisted model predicted peak spinal loads, which were related to tissue tolerance limits to create recommended weight limits. Compared to two-handed exertions, one-handed exertions resulted in decreased spinal compression and A/P shear loading ($p < 0.001$) but increased lateral shear ($p < 0.001$). Effects were likely driven by altered moment exposures attributable to altered torso kinematics. Differences between spinal loads for one- versus two-handed exertions were influenced by asymmetry ($p < 0.001$) and amplified at lower lift origin/destination heights, lower object masses and larger horizontal distances between the body and the load ($p < 0.001$).

Practitioner summary: A biomechanical model was utilised to compare spinal loading for one versus two-handed lifting/lowering. Spinal loads in compression and A/P shear were reduced for one-handed relative to two-handed exertions. As current lifting guidelines cannot appropriately be applied to one-handed scenarios, one-handed weight limits protecting the lower back are presented herein.

Abbreviations: LBD: low back disorder, EMG: electromyography, A/P: anterior/posterior, MVC: maximum voluntary contraction

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1. Introduction

Low back disorders (LBDs) and low back pain represent the number one cause of disability globally worldwide (Hoy et al. 2014). In the United States, more than 80% of the population experiences a LBD or low back pain at least once in their lifetime (Luo et al. 2004). These LBDs come with immense economic cost, in which the economic burden of LBDs in the United States encompassing both direct and indirect costs ranges from \$84.1 billion to \$624.8 billion (Dagenais, Caro, and Haldeman 2008). Additionally, low back (and neck) pain was one of two conditions for which medical expenditures increased the most between 1996 and 2013 in the United States (Dieleman et al. 2016).

In the early part of this century, a literature review by the National Research Council and the Institute for Medicine concluded that LBDs are linked to work

exposures, with between 11% and 80% of the occurrences attributable to physical work factors (Nrc 2001). One of the most highly studied risk factors for LBDs remains lifting. Lifting has been observed to place high mechanical loads on the low back, which could lead to LBDs over time if these loads surpass the mechanical tolerance of the tissue (Marras 2012). Several systematic literature reviews and meta-analyses confirm an association between lifting and the incidence of LBDs (R and Ramos 2010; Griffith et al. 2012; Coenen et al. 2014), though the vast majority of the studies that have been conducted focus on lifting scenarios in which two hands are used to lift a box or a load. Far fewer investigations have examined one-handed lifting scenarios, as might be observed when lifting objects from industrial storage bins, stocking products onto shelves, lifting objects with only one handle, and more.

Early laboratory assessments related to one-handed lifting employed a psychophysical approach to determine the maximum acceptable frequency of lifting during horizontal one-handed lifts, such as what may be observed during assembly line work (Garg and Saxena 1982; Mital and Asfour 1983). These studies were limited, however, in that they did not provide an appropriate comparison between the maximum acceptable frequencies of lifting for one versus two-handed exertions, nor did they provide information about the maximum acceptable frequency of lifting when the lift origin height did not match the lift origin destination. In general, psychophysical studies determining maximum acceptable weights may also be limited by the fact that psychophysically-determined limits may not correspond to biomechanical tolerance. Prior literature shows little association between spinal loads and psychophysically-determined maximum acceptable forces (Jorgensen et al. 1999; Davis, Jorgensen, and Marras 2000; Le et al. 2012), presumably because individuals are unable to sense biomechanical loading due to the lack of nociceptors in the intervertebral disc (Adams, McNally, and Dolan 1996).

Later studies investigating one-handed lifting were biomechanical rather than psychophysical in nature, but the results of these studies remain contradictory. Cook, Mann, and Lovsted (1990) asked subjects to lift loads out of industrial storage containers in a stooping posture and showed reduced muscle activity in the erector spinae during a one-handed lift relative to a two-handed lift, at least when the subjects could support the opposite hand on the top edge of the container. On the contrary, Allread, Marras, and Parnianpour (1996) showed that trunk kinematic measures traditionally associated with LBD development were more pronounced for one-handed lifting tasks relative to two-handed ones. However, both studies lacked a comprehensive picture of biomechanical loading (and therefore biomechanical risk) on the low back while performing one-handed lifts compared to two-handed ones.

A few studies have successfully predicted complex loads acting on the lumbar spine during both one- and two-handed lifting using a biomechanical model. Collectively, these studies suggest that the effects of using one versus two hands for a lifting exertion on the biomechanical loading to the lumbar spine are subject to complex interactions with other lifting conditions, including asymmetry, lift origin and more. For example, Marras and Davis (1998) observed an interesting interaction between using one versus two

hands when lifting and the lift origin asymmetry, which suggested that one-handed lifting may be preferred to two-handed lifting under asymmetric conditions performed on the ipsilateral side of the body. Likewise, Ferguson et al. (2002) explored spinal loading for one versus two-handed lifting from an industrial storage bin, wherein spinal loading measures were subject to complex interactions between lift style (incorporating one versus two hands and one versus two feet) and storage bin design (incorporating various lift origin heights, reach distances). Finally, Kingma and van Dieën (2004) showed that the effects of using one versus two hands on torso kinematics, spinal moments and spinal loading when lifting a load over an obstacle were heavily influenced by whether or not the free hand was used to support the weight of the body during the one-handed lift. These results were recently replicated by Beaucage-Gauvreau et al. (2019), who showed that sagittally symmetric one-handed lifting with a braced arm-to-thigh technique reduces L4/L5 flexion moments, compression and anterior/posterior (A/P) shear relative to unsupported one- and two-handed lifting techniques.

However, while the biomechanical models used in the more dated prior research efforts were considered novel and state-of-the-art when these studies were performed, there remains a need to re-examine one versus two-handed lifting scenarios utilising a more sophisticated and accurate biomechanical model. For example, prior models using straight-line muscle geometry have worked reasonably well for simple tasks requiring little lumbar motion but have been shown to be less reliable when applied to complex asymmetric lumbar motions (Hwang, Knapik, Dufour, Aurand et al. 2016). Biomechanical modelling capabilities today include improvements including estimations of passive muscle forces and curved muscle geometry (Hwang, Knapik, Dufour, Aurand et al. 2016; Hwang, Knapik, Dufour, Best et al. 2016). Both of these improvements are expected to more accurately estimate spinal loads during the complex asymmetric lumbar motions that commonly occur in the workplace (Marras et al. 1993).

Moreover, voids still exist in that no studies have evaluated lumbar spine loads for one versus two-handed lowering, nor have any studies effectively translated their conclusions regarding one one-handed lifting/lowering into clear, applicable weight limits that can be utilised in work environments to assess risk to and protect the lumbar spine. The widely-accepted lifting guidelines presented by NIOSH (Niosh 1981, Waters et al. 1993) specifically exclude one-handed

lifting exposures. Moreover, as has been shown previously (Marras and Davis 1998), known interaction effects between using one versus two hands when lifting and workplace exposure conditions such as asymmetry also suggest that it may be inappropriate to utilise any currently-available two-handed lifting guidelines when analysing a one-handed exertion.

Thus, the objectives of this study were twofold. First, this study aimed to quantify the biomechanical loads imposed upon the low back during one-handed lifting/lowering compared to two-handed lifting/lowering using an updated and more accurate lumbar spine model, specifically in relation to interaction effects present between the lifting technique and external lifting conditions. The second objective of this study was to identify weight limits for the low back during one-handed lifting that can be implemented as guidelines for practitioners.

2. Methods

2.1. Approach

A laboratory study was conducted to understand relationships between using one versus two hands to perform a lifting/lowering exertion, lifting conditions such as load origin/destination and load mass and dependent measures of spinal load. An electromyography (EMG) driven biomechanical spine model was implemented to evaluate lumbar spinal loads in compression, anterior/posterior (A/P) shear and lateral shear. This model relies on several dynamic inputs, including muscle activity for 10 power-producing muscles of the torso (measured via EMG), full body kinematics (derived via motion capture) and ground reaction forces (measured via a force plate). These dynamic inputs are also combined with more “static” inputs such as the anthropometry of the subject, a database of MRI-derived muscle locations and sizes (Jorgensen et al. 2001; Marras et al. 2001), and tissue material properties (such as muscle force–length and force–velocity relationships) from the scientific literature. For example, EMG data are combined with muscle size, length and contraction velocity information to predict dynamic outputs of muscle force. Muscle force data are then combined with body segment dynamics, muscle lines of action, muscle moment arms, vertebral angles and other geometric information to predict dynamic tissue loads on the intervertebral discs. Model structure and validation have been described extensively in previous publications (Marras and Sommerich 1991; Granata and Marras 1993; Dufour, Marras, and Knapik 2013), the

most recent being and Hwang, Knapik, Dufour, Aurand et al. (2016) (newest model structure) and Hwang, Knapik, Dufour, Best et al. (2016) (model validation).

2.2. Subjects

Thirty subjects (15 male: age 27.0 ± 6.4 years, stature 181.8 ± 8.0 cm, and mass 80.8 ± 16.1 kg; 15 female: 26.9 ± 3.9 years, stature 167.8 ± 9.8 cm, and mass 70.5 ± 15.4 kg) were recruited for this study. Using pilot data collected for the study as reference, 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. All subjects self-reported being asymptomatic for LBP and other upper extremity musculoskeletal injuries such as shoulder pain within the past three years. All but one of the male subjects were right-hand dominant. The study was approved by the University's Institutional Review Board.

2.3. Experimental design

A balanced $3 \times 3 \times 3 \times 2 \times 2$ mixed model design was implemented for this study. Independent variables included lift origin/destination height (ankle, knee, waist height), lift origin/destination asymmetry (0 degree position, 45 degrees, 90 degrees relative to the mid-sagittal plane), the mass of the load being lifted (2.7 kg, 7.6 kg, 11.3 kg), horizontal distance of the lift origin/destination from the body (40 cm, 70 cm) and using one versus two hands to perform the exertion. The order of the trials encountered by subjects were first blocked on lift origin/destination height. These three blocks (ankle, knee and waist) were counterbalanced across the 30 subjects to reduce the potential for order or fatigue effects. Within each of the three lift origin/destination height blocks for each subject, conditions were further blocked based on load mass. Instead of counterbalancing these blocks, they were randomised across trials. Finally, the remaining conditions (combinations of asymmetry, horizontal distance, one versus two hands) within each lift origin/destination height/load mass block were fully randomised.

Lumbar spine loads were assessed using the aforementioned EMG-driven lumbar spine model. The model predicts three-dimensional spinal loads (compression, A/P shear, lateral shear) at the superior and inferior endplates of each spinal level extending from T12/L1 to L5/S1. However, because loads along the length of the lumbar spine are generally correlated,

the primary dependent measures of interest in this study were the peak spinal loads observed at the particular lumbar level in which the highest loading was observed within each dimension of loading. Peak spinal loads were derived separately for the lifting and lowering portion of each exertion.

2.4. Apparatus and instrumentation

As shown in Figure 1, this study employed custom height configurable lift tables that were placed in front of the subjects at 0°, 45° and 90° relative to the sagittal plane on the dominant side of each subject's body. Subjects lifted one of three dual-grip medicine balls (Body-Solid, Forest Park, IL, USA) for each experimental condition. These medicine balls had two handles to accommodate either one- or two-handed lifting conditions and were constructed with a solid



Figure 1. Experimental apparatus and instrumentation. Subjects lifted and lowered dual-grip medicine balls of various masses accommodating both one- or two-handed exertions. Height configurable lift tables (as shown) were placed in front of the subjects for lifting and lowering exertions performed from knee and waist, while lifting and lowering exertions performed at ankle height were performed from the floor. Subjects were outfitted with EMG sensors on the 10 power-producing muscles of the trunk and optical motion capture markers from head to toe. All exertions were performed on a load cell.

weighted core such that the same size and shape medicine ball (30.5 cm diameter) was lifted by the subjects regardless of load mass. The medicine balls sat in plastic, 3D-printed bases that were placed onto the configurable lift tables at the correct lift origin as defined by the experimental design.

Kinematic data were captured via a 30 camera OptiTrack Prime 17W motion capture system (NaturalPoint, Corvallis, OR, USA) at a sampling frequency of 120 Hz. The data were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 10 Hz. Kinetic data were captured during model calibration trials and for all experimental conditions via a FP6090-15 force plate (Bertec, Worthington, OH, USA). The kinetic data measured tri-axial forces and moments and were captured at 1000 Hz. Finally, EMG data for the 10 power-producing muscles of the trunk were collected using a wireless Trigno™ system (Delsys, Natick, MA, USA) from bipolar surface electrodes placed bilaterally onto the erector spinae, latissimus dorsi, rectus abdominis, external oblique and internal oblique muscles. EMG data were collected at 1925.93 Hz, and signals were notch filtered at 60 Hz and its aliases, bandpass filtered between 30 and 450 Hz, rectified and smoothed, and low-pass filtered using a second-order Butterworth filter with a cut-off frequency of 1.59 Hz (chosen from a time constant of 100 ms), consistent with standards for reporting EMG data (Merletti 1999).

2.5. Procedure

Upon arriving at the laboratory, subjects were briefed on the study design and provided their informed consent. Several demographic (age) and anthropometric measures (stature, mass, width and depth of the torso at the xiphoid process and navel, and circumference of the torso at the navel) were collected from each subject; these anthropometric measures were ultimately used to scale the lumbar spine model to each individual. Subjects were then outfitted with EMG surface electrodes placed onto the aforementioned power-producing muscles of the trunk according to standard placement procedures (Mirka and Marras 1993) and 41 motion capture markers placed over the entire body in accordance with a custom marker set/kinematic model prescribed by the OptiTrack motion capture software. Then, subjects stood on the force plate and performed a calibration procedure as described previously in Dufour, Marras, and Knapik (2013), in which subjects performed sagittal and lateral bending exercises while holding a 9.07 kg medicine

ball. This procedure eliminates the need to collect maximum voluntary contractions (MVCs) for EMG normalisation, given that MVCs can be sensitive to sincerity of effort, fatigue, training, posture, exertion type and pain (Mirka 1991; Baratta et al. 1998; Keller et al. 1999; Ng et al. 2002; Vera-Garcia, Moreside, and McGill 2010). Instead, muscle and other model parameters are reverse-engineered by minimising predicted (model-derived) moments relative to measured moments in each plane.

After model calibration, the experimental conditions for the study could be collected. During each trial, subjects lifted the medicine ball with either one or two hands from the lift origin predefined by the study design to a sagittal-symmetric common lift destination directly in front of the body at chest height. Subjects were asked to hold the medicine ball here for 1–2 s before lowering the medicine ball, placing it on the lift table in the same location from which it was lifted; the lowering phase was also performed with either one or two hands consistent with the lifting portion of the exertion. All one-handed exertions were performed with the dominant hand, and asymmetric conditions were tested on the dominant side of the body. Subjects were instructed to perform each lift and lower at a natural pace but were not provided instructions regarding their posture (i.e. stoop vs. squat lifts) were also instructed to keep their non-dominant hand unsupported (i.e. not rested on their thigh or the lift tables) during one-handed exertions and to perform each exertion without moving their feet. Constraining foot placement and ensured consistency across trials.

Before collecting each trial, the location of the load to be lifted was precisely measured and controlled by the researchers to maintain consistency across trials. For example, the vertical height of the load was precisely controlled such that the handles of the medicine ball were always located at the same vertical height regardless of whether the exertion was performed with one or two hands. Additionally, the horizontal distance of the load from the body (i.e. L5/S1 joint) was measured before collecting each trial with a tape measure, ensuring consistency with the level of the variable that was to be tested for each given trial. Finally, the asymmetry of the load relative to the sagittal plane was also precisely measured and controlled before collecting each trial *via* aligning the load above strings taped to the ground for each asymmetry level (0°, 45°, 90°), measured relative to the sagittal plane for each subject *via* a protractor.

Due to the large number of conditions collected for each subject (108 conditions), only one repetition of

each condition was collected for each subject. However, the researcher responsible for collecting the data checked for poorly executed movements or poor data quality during/after each trial before collecting the next experimental condition. Any poorly executed trials or trials with poor data quality were repeated by the subject before moving forward. The time spent by the researchers investigating data quality also provided the subjects with a sufficient rest period between exertions. Longer rest periods of 10–15 min were also provided to each subject when switching among the lift origin heights being tested or if requested by the subjects. Any instances in which subjects could not perform the exertion (i.e. it fell outside their strength capability) were recorded by the investigators.

2.6. Analysis

Peak spinal loads were log-normalized where appropriate to meet normality assumptions for statistical analysis. Lift origin/destination height, lift origin/destination asymmetry, load mass, horizontal reach distance to the lift origin/destination, use of one versus two hands to perform the exertion and all two-way interactions on peak spinal loads were input into a generalised linear mixed model with subject and interaction effects involving subject as random effects. Higher order (three-, four- and five-way) interactions were included within the residual/error term of the statistical model. Likewise, the lifting and lowering portion of each exertion were analysed separately as a 'By' variable. Though all two-way interactions were included in the statistical model, only fixed effect tests for main effects and two-way interactions involving the use of one versus two hands to perform the exertion will be reported herein. This is because the primary objective of this study was to compare the biomechanical loads imposed upon the low back during one versus two-handed lifting/lowering, specifically in relation to interaction effects present between one versus two-handed exertions and external lifting conditions. For instances in which a significant p value was observed in a fixed effect test, effect details (i.e. significant differences between/among levels for each independent variable) were assessed using either a least squares means differences Student's t test (for independent variables with two levels) or a least squares means Tukey HSD test (for independent variables with three or more levels). All data were interpreted relative to a significance level $\alpha = 0.05$. Statistical analysis was

performed using JMP 13.0 Pro software (SAS Institute, Cary, NC, USA).

Central to the assessment of biomechanical risk to the critical tissues of the lumbar spine is the relationship between the loads imposed on the tissue and the tolerance of that structure. This concept, which has been recognised as load-tolerance, suggests that when loads experienced by the tissue structure exceed the tolerance of that particular structure, mechanical damage is expected to occur (Marras 2012). Force limits of 3400 N for compression and 700 N for shear are generally accepted tissue tolerance limits for the lumbar spine (Waters et al. 1993; Gallagher and Marras 2012). Thus, peak spinal loads were placed in context of biomechanical risk by comparing peak spinal load magnitudes to these widely accepted risk thresholds. These risk thresholds were also used in the derivation of biomechanically-based weight limits consistent with the second objective of this study.

2.7. Mixed model for one-handed weight limits

To develop biomechanically-determined weight limits for one-handed lifting, mixed modelling techniques were also used to derive an estimate of peak spinal loading in the population as a function of the independent measures and their interactions. Unlike the aforementioned statistical analyses performed in this study, each of the effects entered into the mixed model was treated as continuous rather than categorical. These included: lift origin/destination height (expressed in cm from the ground derived from relationships described by Drillis and Contini (1966), where 15 cm approximately corresponds to ankle height, 50 cm approximately corresponds to knee height and 93 cm approximately corresponds to waist height, range 15–100 cm), lift origin/destination asymmetry (range 0–90°), horizontal reach distance (range 40–70 cm) and object mass (range 0–11.3 kg). ‘Subject’ and two-way interactions with ‘Subject’ were accounted for as a random effects in the model, while three-way interactions (or higher) were excluded as potential model terms and were included in the residual. The dependent measures in the mixed models were the peak spinal loads in compression or shear across the lumbar spine for each trial (i.e. combined lifting and lowering). Each dimension of spinal loading (compression and shear) was considered in its own statistical model. All outliers (data points with values above the upper quartile or below the lower quartile by any more than 1.5 times the interquartile range) were excluded before model selection. Then, model

selection utilised a forward stepwise method to introduce relevant main effects and interaction terms minimising the Akaike Information Criterion and maximising adjusted R^2 .

It was assumed that the point estimate for peak compression or shear derived from each mixed model represents the population mean/median peak spinal load under those experimental conditions (i.e. the 50th percentile). Using the standard normal distribution of each mixed model’s residuals, peak spinal loads for the 80th percentile of the population were subsequently also derived (using a Z score of 0.842). Peak spinal load estimates for the 50th population percentile and 80th population percentile were then related to risk thresholds for spinal loading (3400 N compression or 700 N A/P shear reported by Waters et al. (1993) and Gallagher and Marras (2012), respectively) in order to determine the risk of each particular exertion across either compressive or shear loading dimensions.

Low, medium or high-risk classifications were based on spinal load estimates for the 50th and 80th population percentiles. The 50th population percentile was chosen because it is consistent with the maximal permissible limit criterion logic initially proposed by NIOSH (1981), which approximately set the upper risk threshold at the 15th percentile for compressive tissue strength. Likewise, since current best practice in ergonomics is to protect the 90th percentile male and the 75th percentile female, the 80th population percentile was chosen to split the difference between these two values given that the risk classifications proposed herein do not vary based on sex. Instances in which the compressive or shear spinal load estimate for both the 50th and 80th population percentiles were under the risk thresholds were determined to be of low risk, meaning that 80% or more of the population is expected to be protected from injury under these conditions. Where the 80th population percentile estimate crossed the risk threshold but the 50th percentile estimate did not, the exertion was deemed medium risk, meaning 20–50% of the population is at risk for injury under these conditions. Finally, where both the 50th and 80th population percentile estimates were above the risk thresholds, the exertion was deemed high risk, meaning over 50% of the population is at risk for injury under these conditions. The exertion was also rated according to the subjects’ ability to complete the task due to strength considerations. Consistent with a previously published set of two-handed lifting guidelines for asymptomatic and low back injured populations (Ferguson, Marras, and

Table 1. Statistically significant results

		Hands	Height	Asymmetry	Mass	Horizontal distance	Hands * height	Hands * asymmetry	Hands * mass	Hands * horizontal distance
Peak L3/L4 inferior compression	Lift	***	***	*	***	***	***	***	***	***
	Lower	***	***	***	***	***	***	***	***	***
Peak L5/S1 superior A/P shear	Lift	***	***	*	***	***	***	**	***	**
	Lower	***	***	***	***	***	***	***	*	***
Peak L5/S1 superior A/P shear	Lift	***	—	—	—	—	—	—	—	—
	Lower	***	—	—	—	—	—	—	—	—

Height, asymmetry, and horizontal distance describe the lift origin in lifting and the lift destination in lowering. Within the context of this study, lateral shear loads were deemed to pose little biomechanical risk for injury, so just the main effect of using one versus two hands has been presented for this measure.

* $p < 0.05$.

** $p < 0.01$.

*** $p < 0.001$.

Burr 2005), an exertion was deemed medium risk if 25% or more of the male or female subjects were unable to complete the task and high risk if 50% or more were unable to complete the task. The overall risk for each exertion was defined based on the highest risk level observed across each dimension (compression, shear, strength). However, instances in which the exertion was assigned medium risk across two or more dimensions were also subsequently assigned as high risk, also consistent with methods used by Ferguson, Marras, and Burr (2005).

Recommended weight limits were derived as the transition point between low and medium and medium and high risk classifications. These transition points were calculated via interpolation of the object mass variable, while holding all other factors constant. As the heaviest mass tested was 11.3 kg, point estimates were not extrapolated beyond this point. Thus, some recommended weight limits were deemed safe 'Up to 11.3 kg' but not beyond. However, point estimates were extrapolated for masses lower than the 2.7 kg mass tested. Given that lighter object masses led to reductions in spinal loading, we felt extrapolation in this direction was acceptable.

The effects of lift frequency were not investigated directly in the present study. However, lift frequency has been shown to interact synergistically with force level, contributing to a higher risk for injury at high forces with high repetition (Gallagher and Heberger 2013). Prior studies also implicate increased loading frequency of the lumbar spine with reduced cycles to failure for the tissue (Gallagher and Schall 2017). Thus, the authors felt it appropriate to consider lift frequency in the development of one-handed lifting and lowering guidelines presented herein. In a prior study examining three-dimensional trunk motion and low back pain outcomes across 403 jobs and 48 manufacturing companies, Marras et al. (1993) developed a multiple logistic regression model that used lift

frequency and various trunk motion parameters to predict the probability of a particular job being classified as high risk (at least 12 injuries per 2,00,000 h of exposure). Holding the trunk motion characteristics constant, it was determined that there is an 80% probability of high-risk group membership for lift rates exceeding 255 lifts/h (or 4.25 lifts/minute). It was decided that lift frequencies exceeding 255 lifts/hr should also be denoted as either medium risk (or high risk if any of the other dimensions are also denoted as medium risk), holding to the same logic as described above.

3. Results

Statistically significant main and interaction effects are shown in Table 1. The use of one versus two hands to perform the exertion, lift origin/destination height, lift origin/destination asymmetry, load mass and horizontal distance all affected spinal loading measures. While these effects may not be surprising given that these variables have all previously been shown to affect spinal loading, peak spinal loading measures were more interestingly also affected by interaction effects involving the use of one versus two hands to perform the lift, including hands * height, hands * asymmetry, hands * mass and hands * reach distance effects.

3.1. Spinal loads

Peak spinal loads during lifting generally occurred as subjects began to stand upright after picking up the mass, about one-third of the way towards the neutral position; during lowering, peak spinal loads generally occurred immediately after setting down the mass. Peak spinal loads were generally comparable between lifting and lowering, though peak spinal loads were generally higher in lifting than in lowering (5.2% higher compressive loads, 2.9% higher A/P shear loads

and 9.6% higher lateral shear loads on average). In general, statistical trends were consistent across lifting and lowering scenarios, though exceptions will be noted herein. As higher spinal loads pose a higher risk for injury, peak spinal loads for lifting are shown in [Table 2](#) and in subsequent figures.

The biomechanical model predicted peak values for spinal compression at the L3/L4 Inferior endplate and peak values for A/P and lateral shear at the L5/S1 Superior endplate. To correct for positive skew in the data, all three variables (L3/L4 Inferior Compression, L5/S1 Superior A/P Shear and L5/S1 Superior Lateral Shear) were log-normalized prior to statistical analysis. Across all conditions, comparisons of matched conditions (height, asymmetry, mass, reach distance) showed that compressive spinal loads were decreased by an average of 7.4% for one-handed lifting exertions compared to two-handed lifting exertions ($p < 0.001$) and decreased by an average of 11.5% for one-handed lowering exertions compared two-handed lowering exertions ($p < 0.001$). Reductions for one-handed lifting and lowering relative to two-handed exertions were also noted for A/P shear loading. On average, one-handed lifting exertions recorded 15.6% lower magnitudes of peak A/P shear spinal loading than two-handed lifting exertions ($p < 0.001$), and one-handed lowering exertions recorded 15.9% lower magnitudes of peak A/P shear spinal loading than two-handed lowering exertions ($p < 0.001$). In contrast, peak lateral shear loads were increased on average by 22.7% for one-handed lifting conditions relative to two-handed lifting conditions ($p < 0.001$) and were increased on average by 12.4% for one-handed lowering conditions relative to two-handed lowering conditions ($p < 0.001$). However, the magnitudes of peak lateral shear loads were rather low, exceeding the 700 N tolerance limit in just 0.5% of all the lifting exertions and 0.4% of all the lowering exertions. Thus, within the context of this study with a maximum mass of 11.3 kg, lateral shear loading was deemed to pose little biomechanical risk for injury to the lumbar spine, and results were not interpreted further.

As noted previously, peak spinal loading measures were also affected by main effects of the other independent variables tested. In both lifting and lowering exertions, peak compressive and peak A/P shear spinal loads were increased as lift origin height decreased (i.e. ankle > knee > waist) (both lifting and lowering $p < 0.001$), increased as load mass was increased (i.e. 11.3 kg > 7.3 kg > 2.7 kg) (both lifting and lowering $p < 0.001$), and were increased for the far reach distance relative to the close reach distance (both lifting

and lowering $p < 0.001$). Finally, a main effect of asymmetry, was deemed significant for peak compression and peak A/P shear. However, these main effects must be explained relative to significant hands * asymmetry effects that were also observed for both variables. For spinal compression, peak loads were generally increased with increasing lift origin/destination asymmetry in two-handed exertions (i.e. 90 > 0 and 45) but were generally decreased with increasing lift origin/destination asymmetry in one-handed exertions (i.e. 0 > 45 and 90 in lifting and 0 > 45 > 90 in lowering, both with $p < 0.001$). Likewise, lift origin asymmetry did not affect peak A/P shear loading in two-handed lifting or lowering, though peak shear loads were increased with increased asymmetry in one-handed exertions ($p = 0.005$ for lifting and $p < 0.001$ for lowering).

As noted previously, this and other interaction effects present between the lifting technique and external lifting conditions were of particular interest in this study. Thus, means for one- versus two-handed lifting conditions are shown stratified by each level of the other independent variables in [Figure 2](#) (compression) and [Figure 3](#) (A/P shear). For both compression and A/P shear, a hands * height effect suggested that differences in peak spinal loading between one- and two-handed lifting and lowering were more pronounced at lower lift origin heights (i.e. ankle, knee) (all $p = 0.001$). In fact, differences in peak A/P shear loads between one- versus two-handed exertions were not apparent whatsoever (in lifting) or less drastic (in lowering) at waist height ($p < 0.001$). In terms of mass, a significant hands * mass effect was also noted for both peak spinal compression (during lifting) and A/P shear loading (during lifting and lowering), but with slightly different effect details. While differences in peak spinal compression between one- and two-handed lifting became more pronounced with decreasing load mass ($p < 0.001$), the difference in peak A/P shear between two-handed and one-handed lifting was most pronounced with the 7.3 kg mass as compared to the other two (2.7 kg or 11.3 kg) ($p = 0.0123$). Finally, significant hands * reach distance interactions for both compression and shear suggested that horizontal reach distance had a greater impact on peak spinal loading in both dimensions in two-handed lifting and lowering than in one-handed lifting and lowering scenarios ($p < 0.001$).

3.2. One-handed weight limits

Coefficients and significance levels for the mixed models derived to predict peak spinal loading for

Table 2. Median (interquartile range) L3/L4 Inferior compression and L5/S1 Superior anterior/posterior shear for lifting exertions stratified by condition.

		Reach distance (cm)	Asymmetry (°)	One-handed lifts			Two-handed lifts		
				2.7 kg	7.3 kg	11.3 kg	2.7 kg	7.3 kg	11.3 kg
Peak L3/L4 superior compression (N)	Ankle	40	0	2683 (756)	2993 (873)	3316 (907)	2815 (848)	3042 (728)	3189 (826)
		70		2805 (721)	3334 (718)	3865 (1253)	3054 (840)	3484 (1190)	3740 (1321)
		40	45	2650 (819)	2873 (671)	3284 (783)	2881 (1035)	3085 (1004)	3525 (751)
		70		2625 (519)	3119 (747)	3593 (679)	3000 (561)	3493 (1059)	3767 (867)
	Knee	40	90	2604 (687)	2984 (725)	3219 (817)	2979 (910)*	3298 (1046)	3665 (989)
		70		2687 (738)	3225 (754)*	3562 (817)	3226 (1290)	3570 (849)*	3928 (1402)*
		40	0	2338 (580)	2732 (878)	2990 (1338)	2357 (666)	2644 (818)	2966 (951)
		70		2401 (697)	3025 (904)	3432 (1311)	2640 (584)	3150 (878)	3616 (965)
	Waist	40	45	2148 (661)	2667 (780)	2984 (910)	2447 (678)	2665 (816)	2947 (759)
		70		2261 (521)	2758 (772)	3039 (1056)	2631 (650)*	3125 (876)*	3205 (669)
		40	90	2046 (579)	2377 (725)	2858 (1012)	2458 (686)	2707 (677)	2927 (842)
		70		2168 (633)	2671 (722)	3121 (1015)	2714 (758)	3250 (899)	3354 (960)
Peak L5/S1 superior A/P shear (N)	Ankle	40	0	1366 (424)	1953 (652)	2305 (639)*	1345 (491)	1677 (611)	2042 (562)
		70		1660 (387)	2369 (748)	3000 (696)*	1795 (619)	2621 (790)	3096 (873)
		40	45	1365 (402)	1914 (519)	2280 (617)*	1436 (479)	1775 (494)	2170 (583)
		70		1438 (361)	2078 (574)	2502 (820)*	1853 (697)	2400 (670)	2914 (927)
	Knee	40	90	1345 (497)	1799 (363)	2211 (564)*	1525 (496)	1899 (581)	2121 (654)
		70		1374 (448)*	2007 (478)	2593 (513)*	2006 (515)	2505 (852)	2889 (784)
		40	0	629 (360)	700 (443)	749 (425)	803 (532)	854 (510)	934 (567)
		70		808 (437)	986 (432)	1010 (560)	1010 (629)	1172 (664)	1134 (690)
	Waist	40	45	738 (380)	755 (383)	804 (455)	853 (538)	768 (480)	950 (621)
		70		804 (362)	900 (427)	921 (508)	851 (475)	1042 (583)	1118 (557)
		40	90	671 (354)	777 (287)	814 (353)	734 (447)*	932 (541)*	959 (449)
		70		711 (298)	822 (345)*	938 (356)	967 (426)	986 (475)	1076 (687)*
	Ankle	40	0	269 (200)	279 (155)	329 (153)	303 (231)	310 (183)	391 (228)*
		70		334 (164)	347 (260)	376 (264)	418 (223)	433 (371)	483 (402)
		40	45	240 (146)	269 (177)	295 (144)	328 (199)	320 (205)	356 (361)
		70		317 (196)	369 (204)	395 (216)	390 (218)*	441 (355)*	429 (352)
	Knee	40	90	220 (144)	274 (180)	307 (117)	316 (204)	381 (236)	393 (262)
		70		293 (180)	330 (186)	379 (144)	419 (283)	497 (336)	470 (336)
		40	0	115 (61)	167 (103)	190 (94)*	123 (53)	162 (106)	210 (117)
		70		134 (64)	172 (142)	180 (221)*	153 (71)	194 (203)	243 (187)
	Waist	40	45	115 (62)	153 (102)	213 (128)*	122 (67)	146 (72)	173 (92)
		70		109 (50)	163 (99)	209 (132)*	136 (60)	160 (50)	212 (113)
		40	90	103 (56)	155 (44)	201 (87)*	119 (54)	146 (58)	183 (93)
		70		113 (50)*	146 (71)	191 (101)*	142 (77)	153 (53)	205 (103)

*A missing value for one or more subjects for that condition due to strength considerations or data quality issues.

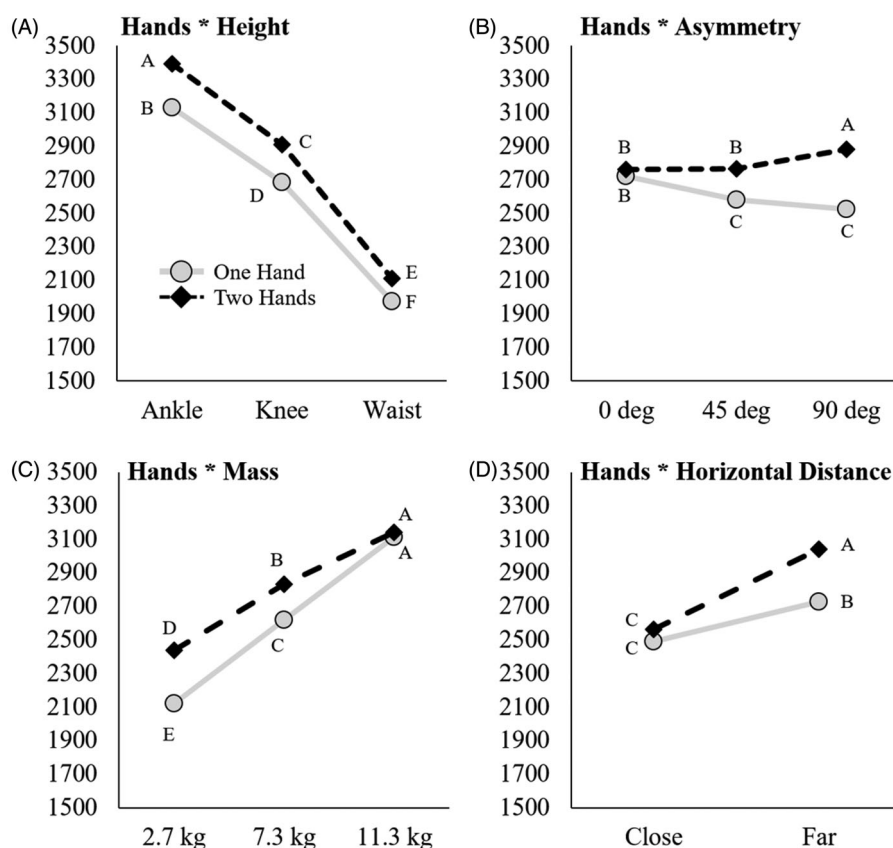


Figure 2. Group means for peak L3/L4 Inferior compression as a function of using either one or two hands to perform the exertion and each of the other independent variables. Groups not connected by the same letter within each plot are significantly different.

compression and A/P shear in the study population are shown in Table 3. A mixed model was not derived for peak lateral shear given the low magnitude of loading observed. Model performance was good in both models, though better for compression (adjusted $R^2 = 0.91$) than for peak A/P shear (adjusted $R^2 = 0.77$). Model residuals were 642 N in compression and 143 N in A/P shear. Using these standard deviations and a Z score of 0.842, it was determined that peak compression and A/P shear are expected to be 541 and 120 N higher (respectively) for the 80th percentile of the population than in the 50th percentile of the population.

Overall risk assignments for each of the conditions tested are shown in Table 4, while recommended weight limits (representing the low-medium and medium-high risk transition points) are shown in Table 5. Most medium and high-risk assignments were driven by shear loading or combinations of compression and shear loading. However, strength did play a role in one-handed lifting and lowering performed with the heavy (11.3 kg) medicine ball at waist height (regardless of lift asymmetry or horizontal distance),

leading to a medium or high risk assignment in the table. While all male subjects were capable of performing all of the one-handed lifting and lowering tasks, 4–7 (or 27–47%) of the female subjects were unable to complete these exertions, depending on the condition. This totalled to 28 trials that were not collected due to strength considerations.

4. Discussion

While a vast body of literature has studied two-handed lifting scenarios from epidemiological, biomechanical and psychophysical perspectives, far fewer studies have examined one-handed lifting scenarios. Moreover, many of the lifting guidelines available in the literature (Niosh 1981; Waters et al. 1993; Ferguson, Marras, and Burr 2005) are unsuitable for assessing one-handed lifting scenarios. This study employed a state-of-the-art biomechanical model to provide estimates of three-dimensional spinal loads while lifting and lowering with one versus two hands. Comparison of peak spinal loads to widely-accepted spinal load risk thresholds allowed for the

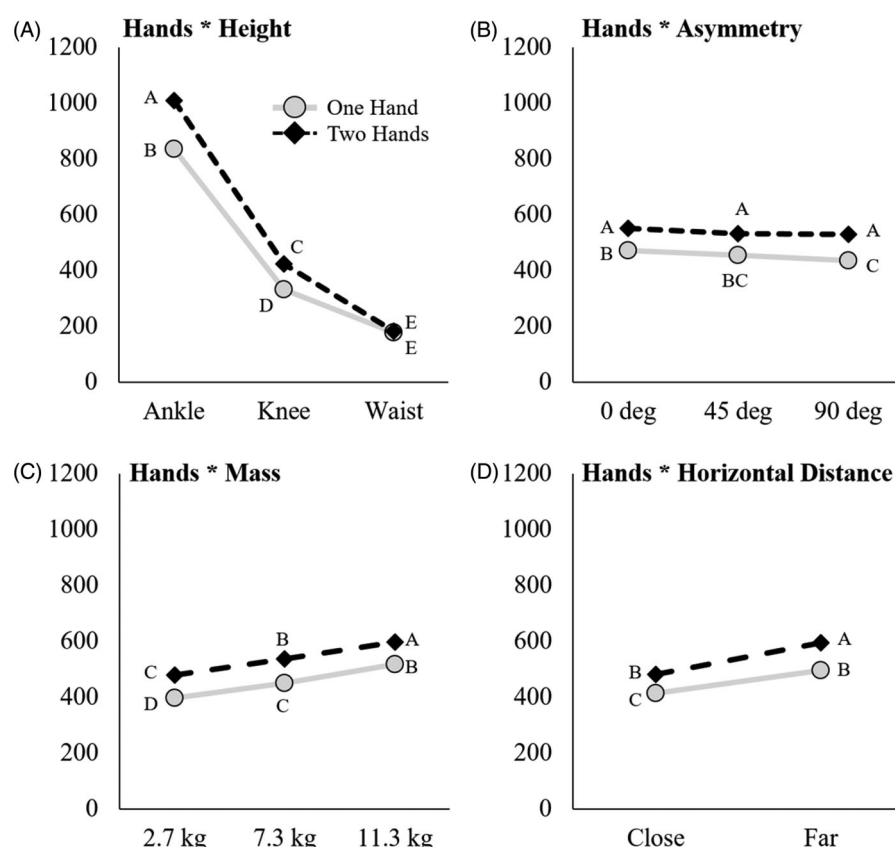


Figure 3. Group means for peak L5/S1 Superior anterior/posterior shear as a function of using either one or two hands to perform the exertion and each of the other independent variables. Groups not connected by the same letter within each plot are significantly different.

Table 3. Parameters and fit of mixed models predicting peak spinal loads for one-handed lifting.

Model	Term	Coeff.	SE Coeff.	t Ratio	Prob > t	R ² _{Adj}	Residual std. dev.
Peak compression	Constant	2295.3	135.2	17.0	<.0001	0.91	642 N
	Height	-15.3	1.1	-13.5	<.0001		
	Asymmetry	-2.8	0.4	-6.8	<.0001		
	Mass	115.9	8.7	13.3	<.0001		
	Horizontal Distance	9.1	0.8	11.7	<.0001		
	(Height-51.8) * (Mass-7.1)	0.6	0.07	8.4	<.0001		
	(Mass-7.1) * (Horizontal Distance-54.5)	1.0	0.1	7.0	<.0001		
Peak A/P shear	Constant	554.1	27.8	20.0	<.0001	0.77	143 N
	Height	-3.5	0.3	-13.4	<.0001		
	Mass	16.1	1.4	11.2	<.0001		
	Horizontal Distance	1.8	0.2	8.9	<.0001		
	(Height-55.6) * (Mass-7.1)	0.1	0.02	4.2	<.0001		
	(Mass-7.1) * (Horizontal Distance-54.5)	0.3	0.04	5.8	<.0001		

Height represents the distance of the lift origin/destination from the ground in cm (range 15–100 cm). Asymmetry represents the asymmetry of the lift origin/destination in degrees relative to the sagittal plane (range 0–90°). Mass represents the load mass in kg (range 0–11.3 kg). Finally, Horizontal Distance represents the horizontal distance between the lumbar spine and the lift origin/destination in cm (range 40–70 cm).

development of biomechanically-determined one-handed lifting guidelines that can be broadly implemented in occupational environments. Consistent with results from a recent biomechanical investigation (Beaucage-Gauvreau et al. 2019), the lifting and lowering scenarios tested in this study noted lower peak spinal compression and peak A/P shear loads on the lumbar spine for one-handed exertions than corresponding two-handed exertions. Thus, from a low

back loading perspective, one-handed lifting may be preferred to two-handed lifting for some lifting and lowering scenarios. A trade-off may exist, however, in terms of risk for shoulder injury, which was not examined in this investigation.

It is expected that differences in compressive and shear loading between one- and two-handed lifting conditions are mainly attributable to varied moment exposure on the lumbar spine from the weight of the

Table 4. Overall risk assignments for each of the one-handed lifting conditions tested relative to spinal loading and subject strength, assuming lift frequency is below 255 lifts/h.

Height	Asymmetry (°)	Distance	Object mass		
			Light (2.7 kg)	Medium (7.3 kg)	Heavy (11.3 kg)
Ankle (~15 cm)	0	Close (40 cm)	High* (C, A/P)	High* (C, A/P)	High (C, A/P)
		Far (70 cm)	High* (C, A/P)	High (C, A/P)	High (C, A/P)
	45	Close (40 cm)	Medium (A/P)	High* (C, A/P)	High (C, A/P)
		Far (70 cm)	High* (C, A/P)	High (C, A/P)	High (C, A/P)
	90	Close (40 cm)	Medium (A/P)	High* (C, A/P)	High (A/P)
		Far (70 cm)	Medium (A/P)	High (A/P)	High (C, A/P)
Knee (~50 cm)	0	Close (40 cm)	Low	Low	High* (C, A/P)
		Far (70 cm)	Low	High* (C, A/P)	High (C, A/P)
	45	Close (40 cm)	Low	Low	High* (C, A/P)
		Far (70 cm)	Low	High* (C, A/P)	High (C, A/P)
	90	Close (40 cm)	Low	Low	High* (C, A/P)
		Far (70 cm)	Low	Medium (A/P)	High (A/P)
Waist (~93 cm)	0	Close (40 cm)	Low	Low	Medium (S)
		Far (70 cm)	Low	Low	High* (C, S)
	45	Close (40 cm)	Low	Low	Medium (S)
		Far (70 cm)	Low	Low	Medium (S)
	90	Close (40 cm)	Low	Low	Medium (S)
		Far (70 cm)	Low	Low	Medium (S)

The individual risk dimensions that led to that overall risk assignment for that condition are represented as (compression – C, shear – A/P, strength – S). Conditions which were designated medium risk along two dimensions and were subsequently rated high risk overall are denoted with *. Lift frequency above 255 lifts/h would designate low risk exertions in the table as medium risk and medium risk exertions in the table as high risk.

Table 5. Recommended weight limits protective of the lower back for each one-handed lifting condition tested, assuming lift frequency is below 255 lifts/h.

Height	Asymmetry (°)	Distance	Recommended weight limit (kg)		
			Low risk	Medium risk	High risk
Ankle (~15 cm)	0	Close (40 cm)	Unacceptable	0–2.2 kg	2.3 kg or more
		Far (70 cm)	Unacceptable	0–1.0 kg	1.1 kg or more
	45	Close (40 cm)	Unacceptable	0–3.8 kg	3.9 kg or more
		Far (70 cm)	Unacceptable	0–2.2 kg	2.3 kg or more
	90	Close (40 cm)	Unacceptable	0–5.4 kg	5.5 kg or more
		Far (70 cm)	Unacceptable	0–3.3 kg	3.4 kg or more
Knee (~50 cm)	0	Close (40 cm)	0–8.0 kg	8.1–8.5 kg	8.6 kg or more
		Far (70 cm)	0–4.9 kg	5.0–6.0 kg	6.1 kg or more
	45	Close (40 cm)	0–8.0 kg	8.1–9.8 kg	9.9 kg or more
		Far (70 cm)	0–4.9 kg	5.0–7.0 kg	7.1 kg or more
	90	Close (40 cm)	0–8.0 kg	8.1–11.0 kg	11.1 kg or more
		Far (70 cm)	0–4.9 kg	5.0–8.0 kg	8.1 kg or more
Waist (~93 cm)	0	Close (40 cm)	0–9.3 kg	Up to 11.3 kg	–
		Far (70 cm)	0–9.3 kg	9.4–10.5 kg	10.6 kg or more
	45	Close (40 cm)	0–9.3 kg	Up to 11.3 kg	–
		Far (70 cm)	0–9.3 kg	Up to 11.3 kg	–
	90	Close (40 cm)	0–9.3 kg	Up to 11.3 kg	–
		Far (70 cm)	0–9.3 kg	Up to 11.3 kg	–

For exertions with a lift frequency greater than 255 lifts/h, practitioners should regard the low risk column as medium risk and the medium risk column as high risk.

torso. One-handed lifting and lowering scenarios do not require subjects to twist as far during asymmetric lifting conditions, nor do they require extensive torso flexion to reach the load. This is because other joints such as the shoulder can compensate in the one-handed case, which is not otherwise possible during two-handed exertions. During one-handed exertions, a less flexed torso places the centre of mass of the torso closer to the lumbar spine, thereby decreasing moment exposure and spinal loading. Altered moment exposure attributable to kinematic differences are supported by Kingma and van Dieën (2004), who showed

reduced torso flexion for unsupported one-handed lifting compared to two-handed lifting. Though kinematic measures have not been reported herein, reductions in torso flexion for one-handed exertions relative to two-handed ones were also noted in this study.

The main effects of lift origin/destination height, lift origin/destination asymmetry, load mass or horizontal moment arm between the body and the load being lifted/lowered observed for spinal loading measures are not particularly surprising. It is generally accepted, for example, that lower lift origins, increased asymmetry, increased load and an increased moment arm

between the body and the mass being lifted increase spinal loading. Thus, interaction effects involving the use of one versus two hands to perform the exertion and the other independent variables were more interesting. In fact, the use of one versus two hands to perform the exertion interacted with every other independent variable studied in some capacity. These interaction effects help point to scenarios in which (relative to low back loading) one-handed exertions should be preferred to two-handed exertions. The benefits of using one hand relative to two hands to perform a lift/lower were generally greatest at lower lift origins or when there was a large horizontal distance between the body and the load. The results of this study also suggest that there is some benefit to one-handed exertions over two-handed exertions with asymmetric lift origins/destinations. Consistent with results from Marras and Davis (1998), compressive spinal loads generally increased as the lift origin asymmetry increased in two-handed lifting scenarios, but compressive spinal loads generally decreased with increased lift origin asymmetry in one-handed lifting scenarios (Figure 1(B)).

However, interaction effects also pointed towards scenarios in which using one versus two hands to complete a lift matters very little in terms of spinal loading. In these scenarios, two-handed lifting and lowering may be preferable to one-handed lifting and lowering so as to reduce biomechanical loading on the shoulder. Instances in which one versus two hands made little difference in spinal loading generally included exertions performed with the heavy (11.3 kg) mass or at the waist lift origin/destination (Figures 2(A,C) and 3(A)). Under these circumstances, it is expected that the weight of the torso was not the dominant contributor of external moment on the lumbar spine, as described previously. Rather, the weight of the load being lifted or lowered was the dominant contributor (for the 11.3 kg mass case) or torso flexion was minimal for exertions regardless of whether the lifting and lowering exertion was performed with one or two hands (for the waist lift origin/destination case).

Two-handed exertions may also be preferable for instances in which strength is an issue. For example, some female subjects (27–47% dependent on the condition) were unable to perform the one-handed exertions at waist height. Lifts performed with one hand from this lift origin required too much bicep strength, and these subjects could not use their body's momentum to help with the lift, as could be done at knee or ankle lift origins.

As previously noted, there are very few guidelines that can be utilised in work environments to assess risk to the low back for one-handed exertions. Sesek et al. (2003) modified the Revised NIOSH Lifting Equation (RNLE) proposed by Waters et al. (1993) for one-handed lifting scenarios. A lifting index was calculated for each hand independently using a load constant of 11.5 kg (half the load constant for two hands), and an effective lifting index for the task was calculated by either averaging the lift indices for both hands or taking the maximum lift index from either hand. Applying their modified RNLE to a database of automobile manufacturing jobs and associated injury data produced significant odds ratios. Thus, these authors concluded that the RNLE can be modified to allow for analysis of one-handed lifting without hindering performance. However, the results of this study suggest that halving the load constant for two hands (23 kg) may be an incorrect simplifying assumption.

Despite the fact that our results suggest one-handed lifting/lowering may be preferable to corresponding two-handed exertions in terms of spinal loading, the weight limits prescribed herein are actually more restrictive than frequency independent weight limits calculated by the RNLE for corresponding two-handed lifting/lowering exertions (Table 6). This led to a rather drastic disagreement between the one-handed weight limits presented herein and two-handed limits calculated *via* the RNLE, especially at lowest (i.e. ankle) lift origin. The rather large variation between the one-handed limits presented here and those for the RNLE may be attributable to how each set of guidelines were derived. The RNLE was developed only partially based upon biomechanical data (it also used epidemiological, physiological and psychophysical data), and the biomechanical criterion in these guidelines accounted for spinal compression but not shear loading. In contrast, the one-handed lifting guidelines derived in this study were primarily based on biomechanical factors, including both compressive and A/P shear loading of the lumbar spine. Herein, damaging A/P shear forces played a large role in the reduction of the weight limits relative to the RNLE. For example, previous work has shown that damaging A/P shear forces may be present when lifting in a stooping postures, as is often observed for lifts performed from lower lift origins (Mcgill 1999), which would explain why the large discrepancies were observed at ankle height. Moreover, comparison of our one-handed lifting guidelines to a more recent set of two-handed lifting guidelines that also accounted for A/P shear loading led to a closer agreement between the

Table 6. Best comparison of one-handed weight limits to existent two-handed lifting guidelines.

Height (cm)	Asymmetry (°)	Horizontal distance	One-hand limit (med.-high risk transition point) (kg)	Revised NIOSH lifting equation frequency-independent RWL (kg)	Ratio (one/two)
15	0	Close (40 cm)	2.2	10	0.22
		Far (63 cm)	1.2	6.3	0.19
	45	Close (40 cm)	3.9	8.6	0.45
		Far (63 cm)	2.5	5.4	0.46
	90	Close (40 cm)	5.5	7.1	0.77
		Far (63 cm)	3.7	4.5	0.82
50	0	Close (40 cm)	8.6	11.7	0.74
		Far (63 cm)	6.5	7.4	0.88
	45	Close (40 cm)	9.9	10.1	0.98
		Far (63 cm)	7.6	6.4	1.19
	90	Close (40 cm)	11.1	8.3	1.34
		Far (63 cm)	8.6	5.3	1.62
93	0	Close (40 cm)	11.3	13.5	0.84
		Far (63 cm)	11.0	8.6	1.28
	45	Close (40 cm)	11.3	11.6	0.97
		Far (63 cm)	11.3	7.4	1.53
	90	Close (40 cm)	11.3	9.6	1.18
		Far (63 cm)	11.3	6.1	1.85

Note that one-handed limits are updated to reflect a 'far' horizontal distance of 63 cm (as opposed to 70 cm) for a more direct comparison. The Revised NIOSH Lifting Equation limits assume good coupling and a lift duration of 1 hour.

guidelines presented herein and those presently available in the literature (Ferguson, Marras, and Burr 2005).

The results of this study should, of course, be placed in context with the study's limitations. This study was performed in a controlled laboratory setting, and only one repetition of each experimental condition were collected for each subject. Additionally, study participants were recruited from the local university population and surrounding community and may be younger than the general working population. In regard to the experimental design, combinations of lift origin height, lift origin asymmetry, load mass and horizontal reach distance were chosen to cover a full range of lift origins, rather than specifically replicating particular job tasks. Foot placement was constrained for all lifting and lowering exertions and did not encompass wide, split stance or perpendicular stances, which could all influence posture and spinal loading. Additionally, because lifting and lowering exertions were both of interest, the load was handled without displacement (i.e. picked up and placed down in the same location). Constrained foot placement and handling the load without displacement may not be representative of work context and could affect the generalizability of the results; however, both were required in order to develop a comprehensive set of lifting and lowering weight limits, consistent with the second objective of the study. Additionally, all exertions were performed standing, so the trends and guidelines presented herein may not be applicable to one-handed exertions performed in other postures like sitting or kneeling. One-handed lifting scenarios were also idealised, in that the lifting and lowering

exertions were performed with excellent coupling between the hands and the load (i.e. handles on the medicine balls) and with the dominant hand to asymmetries on the dominant side of the body (i.e. ipsilateral, not crossing the body). One-handed lifting scenarios in which lifts are performed to the opposite side of the body would be expected to increase spinal loading significantly because these exertions would require extensive torso flexion and twisting (thus, the weight limits proposed herein should not be applied to this lifting scenario). Additionally, all one-handed lifts were unsupported, allowing subjects to support their opposite hand on a table, bin or their thigh while performing a one-handed lift would be expected to decrease spinal loading for one-handed exertions relative to two-handed exertions even more than what was observed presently, consistent with results from prior investigations (Ferguson et al. 2002; Kingma and van Dieën 2004; Kingma, Faber, and van Dieën 2016; Beaucage-Gauvreau et al. 2019). Thus, the weight limits presented herein represent a worst-case scenario.

Most importantly, results and one-handed limits were assessed from a biomechanical spinal loading perspective only. Thus, though one-handed lifting appears to be favourable to two-handed lifting for the low back, future studies should study the impact of one-handed versus two-handed lifting on the upper extremity (shoulder, elbow or wrist). While some prior studies have presented comparisons of strength capabilities for one versus two handed exertions (Fothergill, Grieve, and Pheasant 1991), or normative data for shoulder and/or arm strength (Andrews, Thomas, and Bohannon 1996; Hughes et al. 1999; La Delfa and Potvin 2017), there remain no studies

investigating the influence of one- versus two-handed lifting on shoulder loading or how shoulder loads relate to risk for injury.

Moreover, the risk model presented herein is population-based and should not be used to make inferences about risk for injury for individuals. Spinal loads predicted with the EMG-assisted biomechanical model were compared against risk thresholds that are neither gender nor age dependent. As both sex and age have been shown to influence the ultimate compressive strength of lumbar motion segments (Jäger, Luttman, and Laurig 1991), it is likely that biomechanical risk imposed onto the lumbar spine during one-handed lifting is higher for females and older individuals than would be predicted herein.

Finally, it is well established that increased lift frequency leads to reductions in the maximum acceptable weight of a particular exertion, from biomechanical, physiological and psychophysical perspectives (Waters et al. 1993; Pinder and Boocock 2014; Gallagher and Schall 2017). The present study aimed to include the impact of lift frequency within the proposed guidelines by incorporating results from a prior epidemiological study by Marras et al. (1993). However, it should be noted that the impact of lift frequency on spinal loading was not directly assessed in this study. Practitioners should interpret this aspect of the guidelines with more care than the information that can be derived from the mixed models for spinal loading.

5. Conclusion

Within the context of the experimental conditions tested, the results of this study suggest that one-handed lifting may be preferred to two-handed lifting from a spinal loading perspective; however, this study did not assess the influence of using one versus two hands on other joints such as the shoulders. Collectively, one-handed lifting and lowering conditions tested herein resulted in decreased peak loading on the lumbar spine in both compression and A/P shear. While lateral shear loads were increased for one-handed exertions relative to two-handed exertions, the magnitude of these loads remained low relative to biomechanical tissue tolerance values and subsequent risk for injury, at least within the context of the maximum 11.3 kg mass tested in this study. The choice to use one versus two hands for a particular lifting or lowering exertion should be considered alongside the strength capabilities of the worker (subjects are less capable of lifting with one hand from

waist height) in addition to external lifting characteristics including lift origin/destination height and asymmetry, load mass and horizontal reach distance. Differences between peak spinal loads for one versus two hands were most pronounced at lower lift origins/destinations (i.e. ankle or knee), with asymmetry to the ipsilateral side of the body, and at lower masses. Likewise, differences were less observable at higher lift origin/destination heights and heavier weights, suggesting lifting or lowering scenarios where use of one instead of two hands may not offer any benefit. Finally, this is one of the first studies to present weight limits for one-handed lifting and lowering that are protective of the lower back, as current lifting recommendations do not include one-handed cases.

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