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Spinal loading and lift style in confined vertical space

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ABSTRACT

The objective of this study was to investigate biomechanical loads on the lumbar spine as a function of working in a confined vertical space, consistent with baggage handling inside the baggage compartment of an airplane. Ten male subjects performed baggage handling tasks using confined (kneeling, sitting) and unconfined (stooping) lifting styles. Dependent measures of torso flexion and three-dimensional spinal loads were assessed with an electromyography-driven biomechanical model. Lifting exertions typical to airline baggage handling posed significant risk to the lumbar spine, regardless of lifting style. Statistically significant differences attributable to lift style (stooping, kneeling, sitting) were not observed for peak compressive, lateral shear, or resultant spinal loads, but lifting while kneeling decreased anterior/posterior (A/P) shear spinal loads relative to stooping ($p = 0.02$). Collectively, kneeling offers the greatest benefit when lifting in confined spaces because of the ability to keep the torso upright, subsequently reducing shear forces on the lumbar spine.

1. Introduction

It is well recognized that lifting can place considerable stress onto the lumbar spine and serves as a major risk factor for occupational low back disorders (LBDs) (Bernard, 1997; Griffith et al., 2012; Hoogendoorn et al., 2002; NRC, 2001). Risk factors that have been shown to exacerbate biomechanical risk associated with lifting include lifting heavy loads, high frequencies of lifting, lift asymmetry, and lifting in awkward or restricted postures or confined spaces (Coenen et al., 2014; Hoogendoorn et al., 1999; NRC, 2001; Picavet and Schouten, 2000). While the first three of these risk factors (object weight, lift frequency, and lift asymmetry) have historically been addressed via implementation of assessment techniques and guidelines designed to control LBD incidence (Ferguson et al., 2005; Marras et al., 1999; Snook and Ciriello, 1991; Waters et al., 1993), many are appropriate for standing lifts only and are not applicable to job tasks performed in confined spaces. Though some potential workplace interventions for lifting in vertical confined spaces have been proposed (Riley, 2009), the most common (cranes, hoists, etc.) typically designed to mitigate biomechanical risk, are often not feasible to implement (Gallagher, 2005).

To date, the majority of the scientific literature related to lifting in confined spaces has been performed in the mining industry and has focused on recommending appropriate maximum acceptable weights of lift (MAWL) using a psychophysical approach. This experimental method derives maximum acceptable weights for the working population while relying on subjects to adjust the weight of the object lifted to the maximum that they can handle without excessive fatigue. Notably, psychophysical studies have shown reductions in the MAWL of up to 20% while kneeling (Gallagher, 1991; Gallagher et al., 1988; Gallagher and Unger, 1990) or sitting (Gibbons, 1989) in confined vertical spaces relative to standing. Prior studies have also shown that lifting in restricted postures is accompanied by negative physiological changes including increased heart rate and increased oxygen consumption (Gallagher et al., 1988; Gallagher and Unger, 1990).

Despite the scientific knowledge related to lifting in confined spaces from the psychophysical approach, very few studies have investigated confined space lifting scenarios with biomechanical methods. Stalhammar et al. (1986) and Gallagher et al. (1988) were among the first to use biomechanical methods, collecting electromyography (EMG) signals from low back muscles while lifting in postures including squatting, stooping, kneeling, and sitting across both studies. These studies provided relative differences in muscle activity among lifting styles but did not provide a comprehensive picture of the biomechanical loads on the critical tissues of the lumbar spine (i.e., the intervertebral discs, or IVDs). Gallagher et al. (1994) and Splittstoesser et al. (2007) later used an EMG-assisted model to predict lumbar compression and shear for stooping and kneeling postures in confined space, while Middleton et al. (2016) used the Three-Dimensional Static Strength Prediction Program

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(3DSSPP, University of Michigan Center for Ergonomics, Ann Arbor, MI, USA) to predict lumbar spinal loads for squatting versus sitting in a military population. However, there remain no biomechanical studies that directly compare spinal loading for kneeling, sitting, and stooped lifting styles altogether in a single study. Moreover, biomechanical methods and models today are far improved from years ago and now include estimations of passive muscle forces and curved muscle geometry (Hwang et al., 2016a, 2016b). Both are expected to more accurately estimate spinal loads during the complex (i.e., highly flexed) lumbar motions that commonly occur when lifting in confined vertical space.

As individuals are unlikely to sense biomechanical loading on critical tissues in the spine due to the lack of nociceptors in the IVD (Adams et al., 1996), prior studies have shown little association between maximum acceptable forces derived using psychophysics and spinal loads derived via biomechanical measures (Le et al., 2012; Weston et al., 2018). Subsequently, confined space lifting scenarios have not been measured within the context of biomechanical risk. Given the scientific literature that is available on this topic, it is expected that lifting in vertically confined spaces increases biomechanical risk to the lumbar spine due to several factors. Disrupted muscular synergy of the trunk muscles is likely to occur in restricted postures due to changes in muscle moment arms and muscle lengths and the inability of other muscles in the hips and thighs to contribute to the lift (Gallagher, 2005). Confined vertical space may require subjects to assume more flexed torso postures, contributing to an increased moment exposure to the lumbar spine attributable to the weight of the torso (Gallagher et al., 2001). Finally, workers may be unable to bring their bodies close to the load that is to be lifted due to barriers created by the knees and legs during kneeling and sitting (Gallagher et al., 2001).

Lifting in confined vertical spaces is observed across several industries including airline baggage handling, mining, construction, maintenance, and shipbuilding. A very common job task of particular interest in this study was biomechanical loads encountered for airline baggage handling, where workers often kneel or sit inside the baggage compartments of narrow-bodied airplanes while loading or unloading bags in the tarmac area (i.e., the area where airplanes are parked for departure and arrival). Other awkward postures, such as stooping, are commonly used by baggage handlers for loading and unloading bags between a belt loader and baggage carts. Baggage handlers typically lift 5–10 bags per minute during loading/unloading to the airplane (Oxley et al., 2009). The average weight of bags handled by baggage handlers is about 15 kg, and a small percentage (3%) of bags handled on a typical day exceeds airlines' standard limit of 23 kg (Lu et al., 2018). Injury risks associated with these tasks are repetitive heavy lifting exertions in awkward and restricted postures in the cargo hold space ranging from 46 to 55 inches (i.e., 117–140 cm) in height for narrow-bodied airlines (Tapley and Riley, 2005; Tapley et al., 2007). Automatic container systems for loading and unloading checked baggage have been used in larger or wide-bodied airplanes. These lift assist systems, however, are not available for narrow-bodied airplanes (Lu et al., 2015). Most recent data show that approximately 55% of commercial fleet were narrow-bodied airplanes in 2017 (Fi-Aeroweb.com, access January 18, 2019).

There are nearly 180,000 airline baggage handlers working for commercial fleets in the US (Lu et al., 2015). They encounter annual incidence rates of work-related injuries at more than three times the rate of private industry as a whole (BLS, 2016). Moreover, 35% of baggage handlers have experienced a low back injury at some point in time (Tafazzol et al., 2016), leaving these workers more susceptible for a more costly recurrent injury (van Poppel et al., 1998).

Given that scientific investigations assessing biomechanical loading and risk to the low back while lifting in vertically confined spaces are limited and the particularly high incidence of LBDs related to airline baggage handling, the objective of this study was to quantify biomechanical loads during baggage handling using various lifting styles. Specifically, this study quantified loads placed onto the lumbar spine in a confined vertical space simulating the inside of a simulated narrow-bodied aircraft. It was hypothesized that lifting bags while kneeling and sitting significantly increase spinal loading relative to standing and therefore, increase biomechanical risk.

2. Methods

2.1. Approach

A laboratory study was conducted to evaluate biomechanical risks to the lumbar spine associated with lifting in confined space. The experimental vertically confined space lifting task was designed to represent airline baggage handling performed inside of a narrow-bodied aircraft. Dependent measures consisted of torso kinematics and the biomechanical forces imposed upon the lumbar spine. Lumbar spinal loads were derived from a biologically-assisted EMG-driven lumbar spine model; this model will subsequently be described in more detail in Section 2.6 (Signal Processing and Biomechanical Model).

2.2. Subjects

Ten male participants (age 25.2 ± 6.4 years (SD), stature 183.3 ± 8.6 cm, mass 81.8 ± 9.1 kg) recruited from The Ohio State University and surrounding community participated in this study. These participants did not have experience in manual materials handling tasks in workplace settings. This population size was deemed appropriate to detect effects in variables of interest with a power of 0.80 and a significance level $\alpha = 0.05$ prior to beginning the study. None of the subjects reported any musculoskeletal complaints in the year preceding the study. Two of the subjects recruited were left-hand dominant. All subjects provided informed consent to the research protocol as approved by the University's Institutional Review Board (study ID, 2017H0178).

2.3. Experimental design

A fully balanced 3 x 4 x 2 mixed model design was implemented for this study to assess the effects of lifting style (3), exertion type (4), baggage weight (2), and their interactions on trunk kinematics and spinal loads. Experimental conditions were blocked based on lifting style, with the combinations of exertion type and baggage weight randomized within each block; likewise, each block of lifting style was also randomized within the design. Two repetitions of each experimental condition were collected, and repetitions of each trial type were collected back-to-back.

2.3.1. Independent variables

Independent variables were selected to be consistent with lifting exertions observed via field observation and included lifting style, exertion type, and baggage weight. Lifting styles investigated included standing (i.e., stooping), kneeling, and cross-legged sitting. Exertion type included loading and unloading bags at low and high vertical heights, where the vertical heights described the vertical height of either the lift destination or lift origin. For low vertical heights, the origin or destination of the bag that was to be lifted was the floor (0 cm); likewise, for high vertical exertion heights, the bag that was to be lifted was placed on top of (i.e., stacked) or removed from (i.e., unstacked) two other bags at a vertical distance of 50.8 cm from the floor. Finally, baggage weight included bags of 14.5 kg (32 lbs.) and 22.7 kg (50 lbs.). Bags weighing 14.5 kg have been noted to represent the industry average (Tafazzol et al., 2016), while bags weighing 22.7 kg have been noted to represent bags at the 95th percentile for weight (Lu et al., 2018). In all exertions, subjects lifted a medium sized suitcase (66.0 cm long x 45.7 cm wide x 25.4 cm tall) at a relaxed pace; the suitcases were filled with sand bags to bring them to the appropriate weight and ensure the weight was evenly distributed. Subjects were instructed to use the handles at the top and bottom of each bag when lifting.

2.3.2. Dependent variables

Dependent measures included peak flexion, lateral bending, and twisting angles in the torso for each trial and spinal loads including compression, anterior/posterior (A/P) shear, lateral shear, and resultant spinal load. Lumbar loads were predicted for the lumbar levels extending from T12/L1 to L5/S1 for each trial. However, as spinal loads are generally correlated across lumbar levels, the dependent measures of interest in this study were the peak compressive, shear (A/P and lateral), and resultant loads assessed at the lumbar level in which the highest spinal loads were observed.

2.4. Instrumentation and apparatus

EMG activity was obtained bilaterally for the power-producing muscles of the torso, including the erector spinae, internal oblique, latissimus dorsi, external oblique, and rectus abdominis. These data were sampled at 1000 Hz using a Motion Labs MA300-XVI system (Baton Rouge, LA, USA). Kinetic data were obtained at a sampling frequency of 1000 Hz using a force plate 60 cm wide x 90 cm long x 15 cm tall (Bertec 6090–15, Worthington, OH, USA); however, kinetic data were only collected during model calibration, not during the experimental conditions. Finally, kinematic data were collected at a sampling frequency of 120 Hz using a 42-camera optical motion capture system (OptiTrack Prime 41, NaturalPoint, Corvallis, OR, USA). The accuracy of this system has been validated to be less than 200 μm in 97% of the capture volume (Aurand et al., 2017). All signals were gathered with custom laboratory software written in MATLAB (MathWorks, Inc., Natick, MA, USA) and synchronized using a data acquisition board (USB-6225, National Instruments, Austin, TX, USA).

2.5. Experimental procedure

Upon arriving at the laboratory, subjects were briefed on the study protocol and gave informed consent. Anthropometric measures including stature, mass, trunk width and depth, and trunk circumference were collected, as these measures would be used to scale the aforementioned biodynamic spine model individual to that subject. Subjects were then prepped with surface electrodes on the aforementioned power-producing muscles of the torso using standard placement procedures (Mirka and Marras, 1993). Forty-one motion capture markers were placed onto the subject consistent with a marker set custom to OptiTrack's motion capture software (Motive v1.10.3). A total of four redundancy markers added to the hip and shank segments to account for potential marker occlusion during sitting or kneeling trials. Subjects also donned plastic kneepads, similar to what a baggage handler would wear on the job.

After sensor placement, each subject was given the opportunity to practice a few of the experimental conditions. Once they felt comfortable, subjects performed standing, kneeling, and sitting lifts on the force plate, picking up the 22.7 kg bag from the floor and stacking it onto two other bags (effectively the "loading high" condition with the 22.7 kg bag). These lifts were performed at a comfortable pace and were designed to expose each subject to a dynamic range of muscle length, muscle velocity, and neuromuscular recruitment patterns representative of the other experimental conditions to be tested. Consistent with a previously described no-max calibration procedure (Dufour et al., 2013), the data derived from these lifts were used to optimize model parameters for each subject and lifting style and subsequently calibrate the biomechanical spine model.

After model calibration, the 48 experimental lifts were collected off of the force plate. In loading conditions, subjects lifted the bag from the ground on the dominant side of the body to the appropriate vertical destination (low or high) on the non-dominant side of the body (i.e., right to left in right-handed subjects and left to right in left-handed subjects). In unloading conditions, the direction of the lift was reversed; subjects lifted from the appropriate vertical origin (low or high) on the non-dominant side of the body to the ground on their dominant side. Subjects were allowed to self-select their horizontal distance from the bag and their lifting pace for all exertions so as to investigate lifting exertions that were representative of how workers would more naturally perform the exertions in an occupational environment. Subjects were also instructed to handle the bags realistically. Precision placement was not of utmost importance, but subjects were also asked to refrain from throwing the bags to the ground. Finally, potential fatigue effects were mitigated via short rest breaks (1–2 min) for the subjects between each lift and longer rest breaks (10–15 min) for the subjects between lifting style blocks.

The kneeling and sitting lifting trials were performed within a custom-built frame made from T-slotted aluminum (80/20 Inc., Columbia City, IN, USA) outfitted with a ceiling of mesh netting to allow for motion capture tracking from above (Fig. 1). In all kneeling and sitting exertions, subjects were confined within 1.22 m (48 inches) of vertical space, representative of the vertical constraint within a narrow-bodied airplane. In contrast, subjects performed all standing exertions outside of the custom-built frame with no constraints on vertical space. The standing condition represented baggage handling outside of the narrow-bodied aircraft and therefore served as a control condition. It was clear that all subjects self-selected a stooping lifting posture when lifting bags from the floor as opposed to keeping the torso upright and squatting to reach the load; thus, standing lifts will herein be described more clearly as "stooping."

2.6. Signal Processing and Biomechanical Model

Before being utilized as an input in the biomechanical spine model, EMG signals were notch filtered at 60 Hz and its aliases and band-pass filtered at 30–450 Hz; then, the signals were subsequently rectified and smoothed using a fourth-order low pass filter with a cutoff frequency of 1.59 Hz, which corresponds to a time constant of 100 ms. Likewise, kinematic data derived from motion capture were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 10 Hz.

Using the location and orientation of the torso and hip segments (expressed in quaternions) from motion capture, a custom script calculated the torso angle (via Euler angles) for each time point as the relative difference between the hip and torso segments, decomposed into sagittal, lateral, and axial components. The peak torso flexion angle,

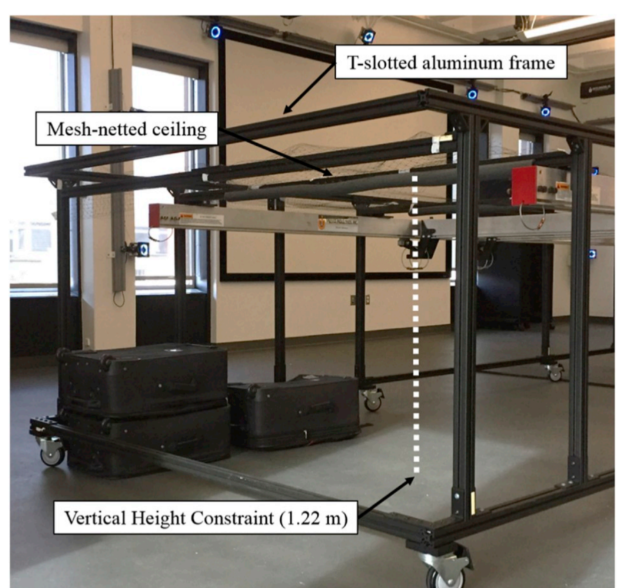

Fig. 1. Experimental setup used for kneeling and sitting lifting conditions.

lateral bending (either side), and twisting angle (either side) for each individual trial were subsequently extracted. Likewise, spinal loading measures were calculated using the aforementioned biomechanical spine model. EMG, kinetic data (gathered during model calibration), and kinematic data were processed together as dynamic inputs to the model. Anthropometric measurements collected for each subject through both optical motion capture and an anthropometer (for scaling), tissue mechanical properties, and MRI-derived muscle cross sectional areas and moment arms (Jorgensen et al., 2001; Marras et al., 2001) were also used as inputs. Ultimately, model outputs included 3-dimensional spinal loads (compression, A/P shear, lateral shear) estimated along the length of the lumbar spine from T12/L1 to L5/S1, calculated using a multibody dynamics software (Adams MSC Software, Santa Ana, CA, USA). This biomechanical spine model has been described extensively in the scientific literature (Dufour et al., 2013; Granata and Marras, 1993, 1995a; Marras and Sommerich, 1991a, b). Recent improvements also include wrapping lines of action for the torso muscles and the separation of active and passive muscle force components (Hwang et al., 2016a, 2016b).

2.7. Analysis

The lumbar level with the highest magnitude of spinal load in each dimension (compression, A/P shear, and lateral shear) served as the basis for assessing biomechanical risk. The spinal loads at these vertebral levels were also compared to documented tolerance limits for spinal loading (Gallagher and Marras, 2012; Waters et al., 1993).

Post processed kinematic and spinal load data were log-normalized where appropriate to meet the required normality of data distribution for testing for a statistical difference. The effects of lifting style, exertion type, baggage weight, and all potential interactions on torso kinematics and biomechanical loads assessed at the peak resultant spinal load were assessed via a within-subject, three-way analysis of variance (ANOVA) with significance level (α) 0.05. Post-hoc analyses were performed using Tukey HSD tests where appropriate. All results were analyzed using JMP 11.0 software (SAS Institute Inc., Cary, NC, USA).

3. Results

As shown in Table 1, main effects of lifting style, exertion type, and bag weight in addition to several interactions were noted to affect torso kinematics or peak spinal loads. Main effects of lifting style, exertion type, and baggage weight and interaction effects of lift style * exertion type and lift style * weight were recorded in some or all of the dependent measures. However, neither an exertion type * bag weight interaction effect nor a three-way lifting style * exertion type * bag weight interaction were observed for any of the dependent measures.

3.1. Torso kinematics

Torso kinematics for the experimental conditions tested in this study are represented in Fig. 2. The main effect of lifting style differed in each plane. Torso flexion was most extreme in sitting postures, followed next by stooping, and then kneeling ($p < 0.001$). Torso twisting was also most extreme in sitting postures, but was followed next by kneeling, and then finally stooping ($p < 0.001$). Finally, lateral bending was increased for kneeling relative to both stooping and sitting ($p = 0.003$).

Exertion type was only significant as a main effect for peak torso flexion, which was reduced significantly while unloading bags from the higher vertical height (i.e., unstacking), with no significant differences observed among the other three exertion types ($p < 0.001$). Peak torso flexion was also influenced by a lifting style * exertion type interaction effect; whereas peak torso flexion was reduced for stooping relative to sitting when unloading from a high vertical height, there was no statistically significant difference in peak torso flexion between stooping and sitting for the other exertion types ($p < 0.001$). It should be noted that a significant lift style * exertion type interaction was also observed for peak lateral bending angle, though post-hoc analysis revealed that when stratifying the data by exertion type, significant differences were only observed among the different lifting styles during loading and unloading from the high vertical height.

Finally, whereas bag weight did not significantly affect peak torso flexion, peak lateral bending and twisting angles in the torso were both increased when handling the heavy bag compared to the light bag ($p < 0.001$ for lateral bending, $p = 0.02$ for twisting). However, the lift style * bag weight interaction that was also observed for peak torso twisting suggested that the effect of bag weight on twisting angle was only significant when kneeling, not when sitting or stooping ($p = 0.004$).

3.2. Compressive spinal load

The highest compressive spinal loads were observed at the L4/L5 lumbar level. At L4/L5, peak compressive spinal loads surpassed the 3400 N action limit for spinal loading proposed by (Waters et al., 1993) in 86.6% of the stooping trials, 74.2% of the kneeling trials, and 65.8% of the sitting trials. Fewer peak compressive loads were noted to exceed the 6400 N maximum permissible limit for spinal loading (NIOSH, 1981), though it did occur in 7.0% of the stooping trials, 2.6% of the kneeling trials, and 3.8% of the sitting trials.

Compressive spinal load data is shown in Fig. 3. On average, stooping recorded mean peak compressive loads 12.3% higher than kneeling and 11.0% higher than sitting, though a main effect of lifting style on peak compressive spinal load was not observed due to large variability in the spinal compression data ($p = 0.48$). However, statistically significant main effects were observed for both exertion type ($p < 0.001$) and bag weight ($p < 0.001$). Post-hoc tests revealed that loading to the high vertical height (i.e., stacking) and unloading from the low vertical height had significantly higher peak compressive loads than loading to the low vertical height and unloading from the high vertical height (i.e., unstacking); compressive loads were also increased for the 22.7 kg bag relative to the 14.5 kg bag. A statistically significant lifting style * exertion type interaction effect was observed ($p = 0.02$), though post-

Table 1
Statistically significant results (*p < 0.05; **p < 0.01; ***p < 0.001). ↑ Represents data that were log-normalized prior to statistical analysis.

	Lifting Style (LS)	Exertion Type (E)	Baggage Weight (W)	LS * E Interaction	LS * W Interaction	E * W Interaction	LS * E *W Interaction
Torso Kinematics							
Peak Sagittal Torso Flexion	***	***		***			
Peak Torso Lateral Bending	**		***	**			
Peak Torso Twisting	***		*		*		
Spinal Load							
L4/L5 Compression ↑		***	***	*			
L5/S1 A/P Shear ↑	*	***	*	*	***		
L5/S1 Lateral Shear			*				
L4/L5 Resultant ↑		***	***	*			

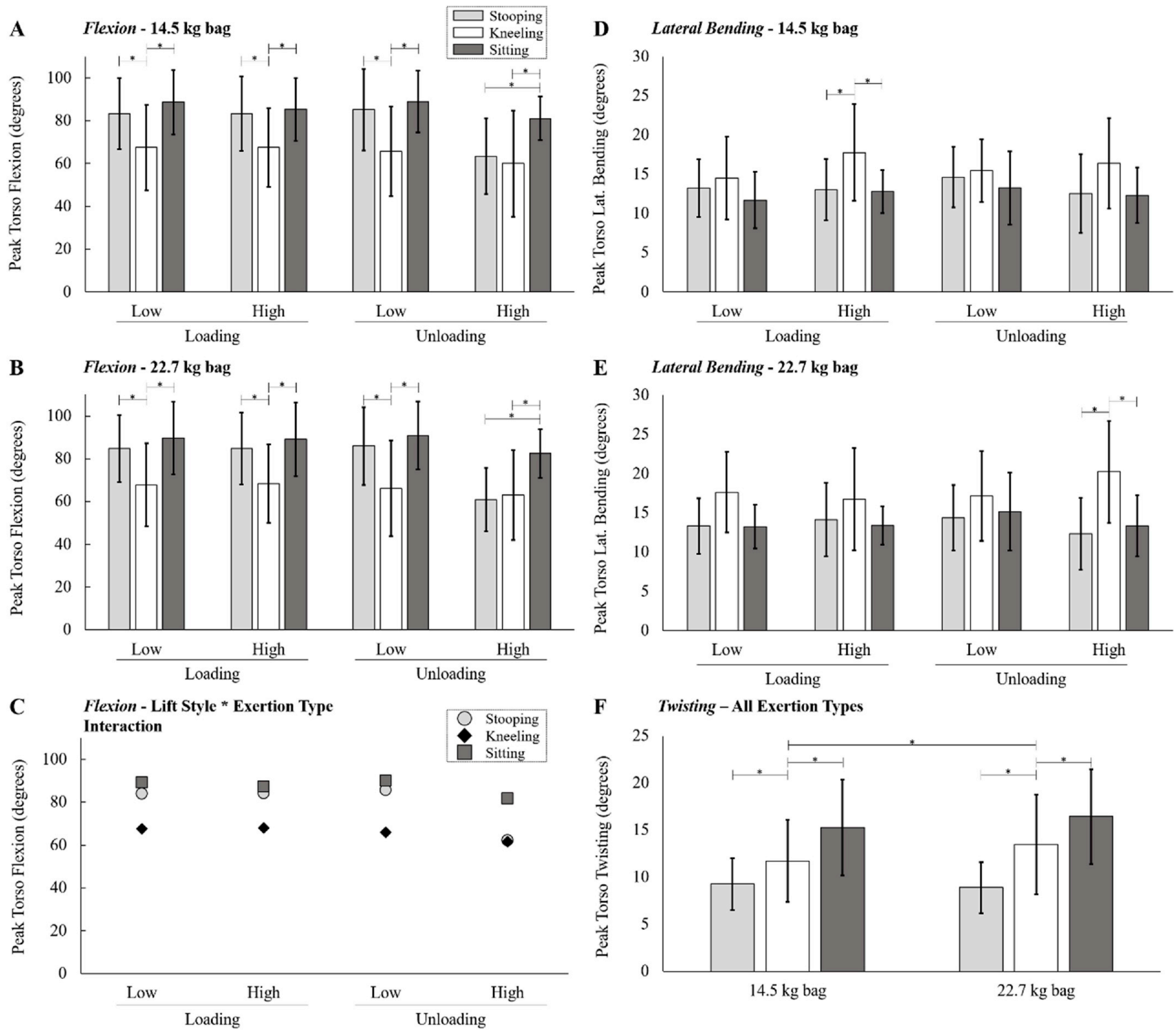

Fig. 2. Torso kinematics including: **(A)** peak torso flexion angles recorded while lifting the 14.5 kg bag, **(B)** peak torso flexion angles recorded while lifting the 22.7 kg bag, **(C)** the significant lifting style * exertion type interaction effect observed for peak torso flexion, **(D)** peak lateral bending angles recorded while lifting the 14.5 kg bag, **(E)** peak lateral bending angles recorded while lifting the 22.7 kg bag, and **(F)** peak trunk twisting angle across all exertions stratified by bag weight. Error bars denote standard deviation, while * denotes a statistically significant difference at an alpha level of 0.05.

hoc analysis revealed that when stratifying the data by exertion type, no significant differences were observed among the different lifting styles.

3.3. Anterior/posterior shear

The highest magnitude of A/P shear spinal loads were observed at the L5/S1 lumbar level. In the context of biomechanical risk, at this vertebral level A/P shear loads surpassed the 700 N threshold presented for frequent loading (>100 times per day) (Gallagher and Marras, 2012) in 87.9% of the stooping trials, 43.9% of the kneeling trials, and 72.2% of the sitting trials.

A/P shear loads at the L5/S1 lumbar level for the experimental conditions tested are represented in Fig. 4. Post-hoc tests showed that in terms of a main effect, peak A/P shear was increased for stooping relative to kneeling ($p = 0.02$), but peak A/P shear for sitting did not differ significantly from the other two lifting styles. However, the statistically significant lifting style * exertion type interaction suggests that this trend held consistent across all loading conditions, whereas no significant differences were apparent among the three lifting styles for the unloading conditions ($p = 0.018$).

As was observed for spinal compression, main effects of exertion type and bag weight were also observed for A/P shear; unloading bags from the high vertical height (i.e., unstacking) recorded significantly lower shear values than the other three exertion types ($p < 0.001$), and shear loads were increased for the 22.7 kg bag relative to the 14.5 kg bag ($p = 0.012$). Finally, a statistically significant lifting style * bag weight interaction was recorded in which the effect of bag weight was more pronounced in a sitting than for stooping or kneeling ($p = 0.0004$).

3.4. Lateral shear

The highest magnitude of lateral shear loads were also observed at the L5/S1 lumbar level. The magnitude of these loads were much smaller than the magnitude of loads observed for A/P shear. Though

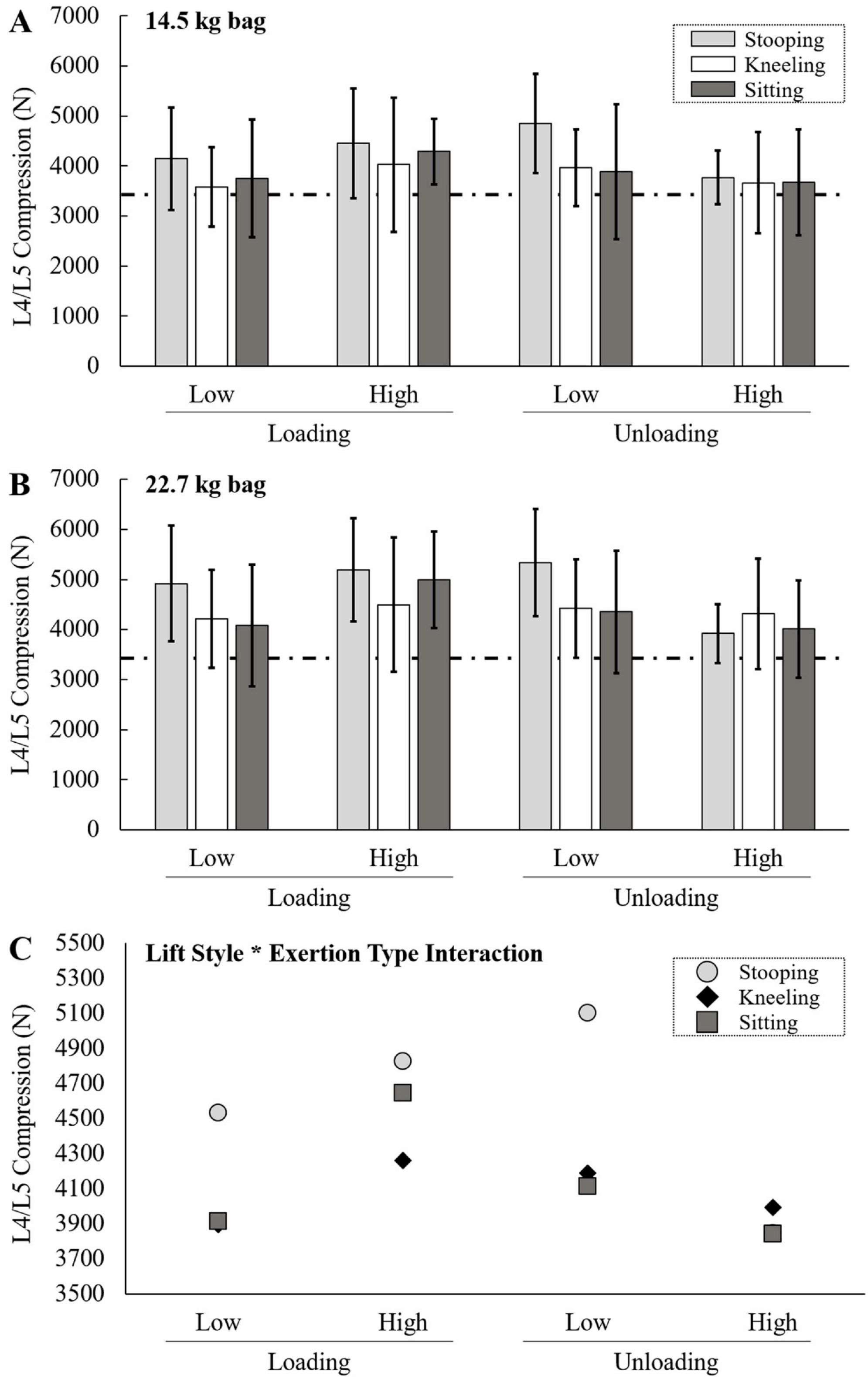

Fig. 3. Compressive spinal loads predicted for **(A)** lifting the 14.5 kg bag and **(B)** lifting the 22.7 kg bag. Spinal loads were compared to the 3400 N action limit proposed by Waters et al. (1993) when estimating biomechanical risk, as indicated by the dashed horizontal line. Error bars denote standard deviation. **(C)** Illustrates the significant lifting style * exertion type interaction effect for this variable.

lateral shear loads did surpass the 700 N threshold presented for shear loading (Gallagher and Marras, 2012) in a handful of the stooping trials (3.2%), kneeling trials (8.4%), and sitting trials (1.9%), it can be concluded that the lateral shear loads observed pose little biomechanical risk for injury.

Peak lateral shear loads are shown in Fig. 5. The only main effect observed for peak lateral shear was that of baggage weight, in which shear loads were increased for the 22.7 kg bag relative to the 14.5 kg bag

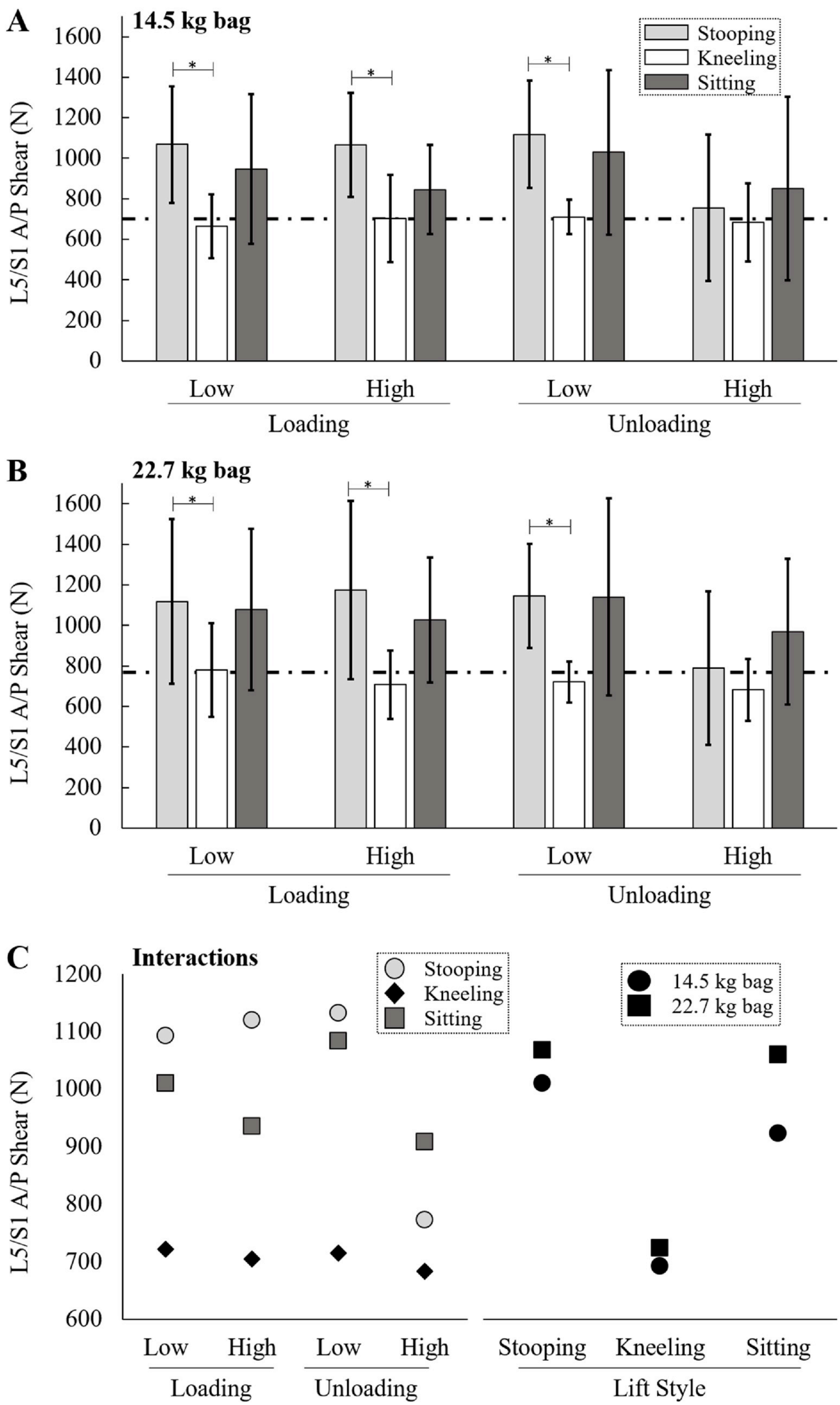

Fig. 4. Peak Anterior/Posterior (A/P) shear forces predicted for **(A)** the 14.5 kg bag and **(B)** the 22.7 kg bag conditions. Spinal loads were compared to the 700 N tissue tolerance limit proposed by Gallagher and Marras (2012), as indicated by the dashed horizontal line. Error bars denote standard deviation, while * denotes a statistically significant difference at an alpha level of 0.05. **(C)** Illustrates the significant lifting style * exertion type (left) and lifting style * baggage weight (right) interaction effects for this dependent measure.

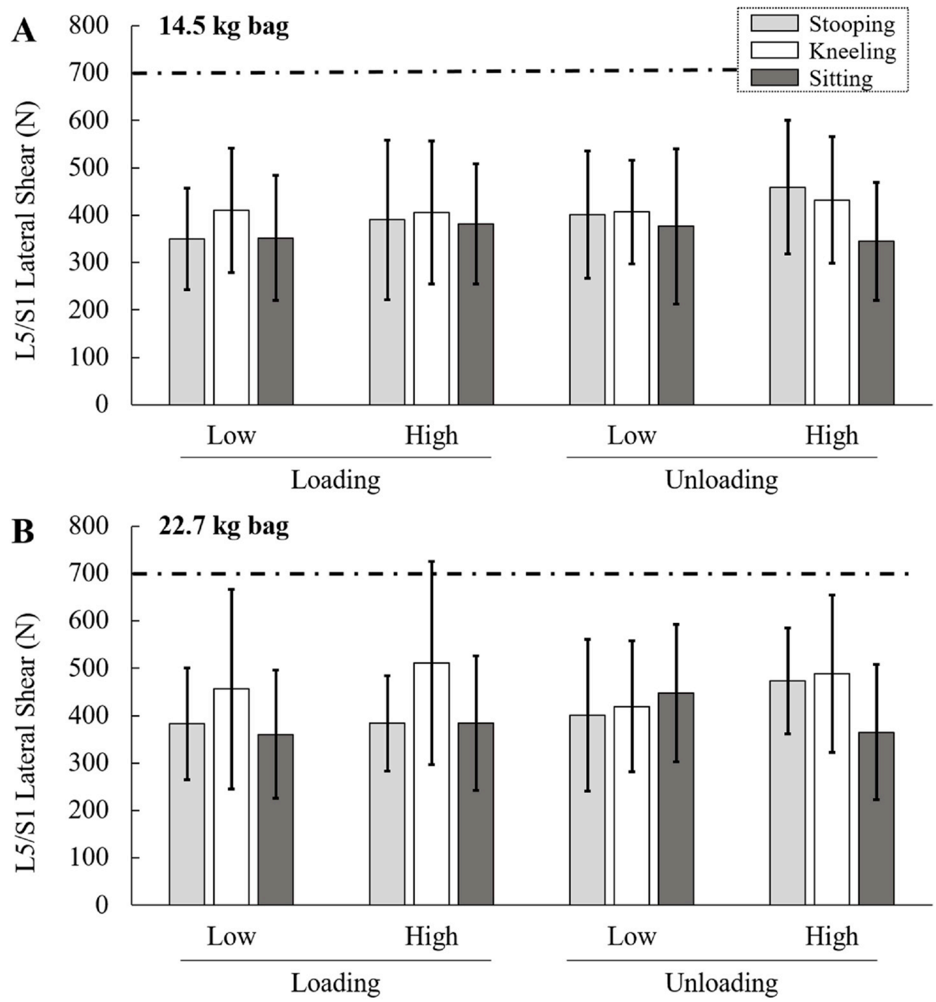

Fig. 5. Peak lateral shear forces for **(A)** the 14.5 kg bag and **(B)** the 22.7 kg bag conditions. Spinal loads were compared to the 700 N tissue tolerance limit proposed by Gallagher and Marras (2012), as indicated by the dashed horizontal line. Error bars denote standard deviation.

($p = 0.011$). No statistically significant interaction effects were observed for this dependent measure.

3.5. Resultant spinal load

Resultant spinal loads were driven mainly by compressive spinal loads in this investigation. The statistically significant effects observed for resultant spinal loads directly mirrored the previously mentioned effects observed for compression, though the p-values derived for each dependent measure did differ very slightly.

4. Discussion

Lifting in confined vertical space is a reality across several industries, including but not limited to airline baggage handling, mining, construction, maintenance, and shipbuilding. A simulated baggage handling environment was chosen to evaluate the effects of lifting in confined vertical space in kneeling and sitting postures relative to a stooping control condition. Though experimental conditions specific to baggage handling were examined, it is expected that the results of this investigation are applicable across all industries in which lifting in confined vertical spaces is observed, and perhaps applicable for industries where space is not confined but work is performed in similar restricted postures, such as agriculture.

Prior to this investigation, few studies utilized a biomechanical approach to investigate lifting in confined spaces. Moreover, there was a need to reexamine confined lifting scenarios with a new and improved biomechanical model considering both wrapping muscle lined of action in the torso and passive muscle forces (Hwang et al., 2016a, 2016b). Interestingly, no statistically significant differences in compression, lateral shear, or resultant spinal loads were observed attributable to lifting style in this investigation. In fact, stooping yielded mean peak compressive loads 11–12% higher than kneeling and sitting on average, contrary to our hypothesis. The results related to A/P shear loading on the spine were also contradictory to the hypothesis proposed, in which A/P shear loads were reduced for kneeling relative to stooping (loads when sitting fell somewhere between these other two lifting styles).

Though initially unexpected, these results can largely be explained by the differences in torso kinematics observed across the lifting styles. Though posture was unconfined for standing lifts, it is clear from examination of peak torso angles observed across the experimental conditions (Fig. 2) that subjects adopted a stooping lifting posture when lifting bags from the floor as opposed to keeping the torso upright and squatting to reach the load. A stooping posture has previously been noted to be used in 50% of all luggage transfers (Stalhammar et al., 1986) and is commonly adopted by workers across many industries when initiating a lift off the floor. This preference for stooping is likely due to the ability to use the hip extensor muscles to aid in the lift (Gallagher, 2006). However, a highly flexed torso posture, as is consistent with stooping, contributes to a significant moment exposure for the lumbar spine, attributable to the weight of the torso (Gallagher et al., 2001, 2002); this increased moment exposure subsequently increased spinal compression for this type of lift. Moreover, previous work has shown damaging A/P shear forces may be present when lifting in a stooping posture (McGill, 1999). Stooping has also been shown to load the passive structures of the lumbar spine (Floyd and Silver, 1955), increasing the potential for damage to the ligaments. Finally, deep flexion of the torso has been shown to disengage the facet joints and decrease the effective moment arm of the lumbar erector muscles, which has implications for increased spinal loading onto the IVDs and reduced tissue tolerance (Gallagher et al., 2005; Jorgensen et al., 2003). However, despite these apparent drawbacks, stooping lift styles do offer one advantage that lifting while kneeling or sitting in confined space does not. The stooping lift style recorded the least degree of torso twisting, a known risk factor for LBDs. It is likely that subjects were able to avoid twisting the trunk by repositioning their feet as needed while standing. Gallagher (2005) argued that repositioning the body is more difficult when kneeling, especially when handling a load, and that workers will often tend to opt for a more efficient trunk twisting motion. Kinematic results from this study confirm this and suggest that sitting may be even worse than kneeling when it comes to torso twisting/asymmetry.

It was initially expected that the knees and legs would create a barrier that prevents subjects from lifting the load close to the body in kneeling and sitting. This barrier was expected to increase the moment arm between the spine and the load being lifted and thus also increase moment exposure to the lumbar spine. However, because the bags were loaded or unloaded from one side of the body to the other rather than the lift being performed purely in the sagittal plane, it is not likely that there was increased moment exposure on the lumbar spine attributable to the barrier created by the knees and legs, a finding that is corroborated by the lack of a statistically significant difference in compressive or resultant spinal loading attributable to lifting style.

That being said, the results of this study generally confirm the results of prior biomechanical assessments examining confined space lifting scenarios or more specifically baggage handling. Consistent with the results of this study, Gallagher et al. (1994) showed increased lumbar shear forces for stooping relative to kneeling. Additionally, several studies performed in relation to work postures seen in confined vertical space also noted increased spinal loading with increased object weight and an increase in vertical destination height, as was seen presently (Gallagher et al., 1994; Splittstoesser et al., 2007). Unloading bags, particularly from the higher vertical height, was also noted to be the most favorable experimental condition in terms of reduction in spinal load, consistent with results from Stalhammar et al. (1986) that suggested that unloading is easier on the back because bags can be slid or dropped when being moved. One major difference between the results of this study and those before it, however, is that no statistically significant differences were noted for spinal compression between stooping and kneeling, contradictory to Gallagher et al. (1994), who noted increased compressive spinal loads for kneeling relative to stooping. This difference could be explained by the addition of passive muscle forces in our updated biomechanical model, which increased muscle forces in the lower back (and subsequently spinal compression) more so during stooping than kneeling due to an increase in lumbar flexion.

Central to the assessment of biomechanical risk to 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 recognized 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). Though individualized tissue tolerances may vary dependent on the individual based on factors such as age, sex, and load frequency (Adams et al., 2000; Brinckmann et al., 1988; Gallagher and Schall, 2017; Jäger et al., 1991), tolerance thresholds presented by Waters et al. (1993) and Gallagher and Marras (2012) are commonly accepted as appropriate estimates of tissue tolerance limits for the general working population. Loads exceeding these values are expected to lead to endplate micro-fractures that have been shown to lead to scar tissue that disrupts nutrition to the intervertebral disc, ultimately leading to disc degeneration (Adams et al., 2000; Brinckmann et al., 1988; Gallagher et al., 2005; Jensen, 1980; Marras, 2008). The overall results of this investigation suggest that considerable biomechanical risk is placed on the lumbar spine during airline baggage handling. Even at the industry average bag weight (14.5 kg), all exertion types recorded mean compressive spinal loads above the 3400 N threshold proposed for compression, regardless of lifting style, in the population tested. Moreover, while excessive compressive spinal loading to the lumbar spine is traditionally seen as most problematic when lifting, the results of this study also suggest that biomechanical risk is also encountered in A/P shear, at least during both stooping and sitting. Stooping and sitting lifting styles recorded mean peak A/P shear loads above the risk limit in all four of the exertion types. Fortunately, the baggage handling task performed (bag transfer from one side of the body to the other) did not also come at the cost of high lateral shear loads placed onto the lumbar spine; only a small percentage of the trials noted lateral shear loads above the risk limit.

In relation to low back loading, the results of this investigation ultimately suggest that in confined lifting scenarios, it may be best for workers to lift while kneeling as opposed to sitting. Subjects lifted in a more upright posture when kneeling, which would be expected to reduce the moment exposure to the lumbar spine attributable to the weight of the torso and engage the facet joints, thereby offloading some of the load from the IVD. In contrast, biomechanical risk to the lumbar spine is likely higher for sitting. Sitting recorded higher peak torso twisting angles than kneeling, and though not statistically significant, A/P shear loads for sitting were generally higher than those for kneeling. Moreover, despite the insignificant differences in load magnitude observed, the same magnitude loads should be expected to do more damage to the tissues of the spine in sitting than in kneeling. In sitting (like in stooping), all of the stress is transferred to the IVDs in these postures as opposed to kneeling, where some of the load is distributed via facet joint contact, which is suggestive of reduced tolerance in highly flexed postures (Gallagher et al., 2005). Like stooping, lifting while sitting may also engage the passive tissue structures of the lumbar spine rather than maintaining active muscle activation in the erector spinae muscles (Stalhammar et al., 1986).

The recommendation for a kneeling lifting style is supported by field data collected inside of baggage compartments of narrow-bodied airplanes by the Spine Research Institute at The Ohio State University (though this data remains unpublished in peer-reviewed literature). A Lumbar Motion Monitor (LMM) was used to objectively gather trunk motion data and estimate LBD risk (see Marras et al. (2000) for details on the LBD risk model) as 42 baggage handlers (age range 25–63, job experience range 5–27 years, stature range 157–193 cm) handled over 600 bags while both sitting and kneeling. Consistent with the results of this study, baggage handling tasks in confined vertical space were deemed high risk regardless of lift style; though the risk for LBDs was comparable between the two lift styles, kneeling also fared better than sitting.

However, our recommendation for kneeling is also subject to its own set of stipulations. Namely, this study approached confined space lifting from a biomechanical perspective, not with a physiological or psychophysical approach. Conclusions drawn from a biomechanical approach employing load-tolerance logic may subsequently differ from those drawn using either of these other two methods. In addition, subjects that are of large stature may not benefit as much as smaller workers when kneeling in confined vertical space. Anecdotally, the two tallest subjects recruited for this investigation (each with stature 192 cm) reported increased discomfort with the kneeling condition. These subjects sat back on their heels slightly in order to fit within the vertical space allotted to them, subsequently also increasing their distance from the load. Finally, recommendations should not be drawn from the results of this study regarding preferred lifting styles in *unconfined* spaces or regarding stooping versus kneeling, as stooped lifting in confined vertical space was not directly assessed in this study (again, the stooping posture was observed as the preferred lifting posture in unconfined space).

The results of this study should also be placed in context with its other limitations. This study was performed as a laboratory investigation which only simulated a confined space environment. Subjects recruited for this investigation were young and inexperienced in baggage handling, and the subject population was comprised only of males. However, this decision in the experimental design was made with concern that lifting a 22.7 kg bag while either kneeling or sitting would exceed the strength capability of most females. Additionally, spinal loads were observed for a short duration of lifting, and neither productivity nor hurried work were assessed. Prior studies have shown that spinal loads vary over the duration of a work shift, as workers begin to vary their neuromuscular recruitment patterns in order to complete a task as a result of fatigue (Chany et al., 2006). Likewise, the psychosocial factor of job stress that would be expected in a hurried work situation has also been implicated in the increase of biomechanical risk to the lumbar spine (Bongers et al., 1993). Both lifting for a more extended duration and hurried work would be expected to increase coactivity in the power-producing torso muscles, thereby further increasing lumbar spinal loads (Granata and Marras, 1995b). Finally, the effects of lifting in confined vertical space were assessed only for the lumbar spine. Future studies should aim to assess the biomechanical risks associated with confined lifting scenarios to other regions of the body in which musculoskeletal disorders might be of concern. For example, as has been investigated in prior biomechanical investigations (Gallagher et al., 2011; Pollard et al., 2011; Porter et al., 2010) and the epidemiological literature (Baker et al., 2003; Coggon et al., 2000; Kivimaki et al., 1992, 1994), kneeling may pose risk of injury in the knee joint itself and subsequently contribute to osteoarthritis, prepatellar bursitis, and meniscal injury.

Given the biomechanical risk observed to the low back in particular for the experimental conditions tested, this study does collectively highlight the need to consider potential interventions that may feasibly be employed in confined space environments, especially given that traditional workplace interventions related to lifting such as a crane or hoist may not be feasible to implement in these environments. These interventions should be assessed from a biomechanical perspective, investigating their ability to reduce lumbar load below the commonly accepted risk thresholds or at the very least relative to the loads observed here.

5. Conclusion

Lifting in confined vertical space is a complex problem impacted by the lifting conditions (vertical height constraint, lift origin, destination, and object weight) in addition to other likely factors such as anthropometry and fatigability. Given the particularly high incidence of LBDs related to airline baggage handling, torso kinematics and spinal loads for exertions typical to this industry were investigated. Collectively, the results of this investigation suggest that lifting exertions typical to the airline baggage handling industry pose significant biomechanical risk to the lumbar spine in compression and in A/P shear in both unconfined (kneeling, sitting) and confined (stooping) lifting scenarios. Where possible, engineering interventions such as cranes, hoists, or vacuum lifts should be implemented in this industry in both confined and unconfined spaces to help mitigate biomechanical risk. However, given that these more traditional interventions may not be feasible to implement in confined space environments, more custom solutions may be necessary. When not practical to implement such an engineering intervention, kneeling lifting styles should be preferred to sitting when lifting in a confined vertical environment (from a low back loading perspective) because of the ability to keep a more upright torso when using this lifting style.

Declaration of competing interest

The authors have no competing interests or conflict of interest to declare.

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