



Impact of two postural assist exoskeletons on biomechanical loading of the lumbar spine

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ABSTRACT

This study evaluated loading on the low back while wearing two commercially available postural assist exoskeletons. Ten male subjects lifted a box from multiple lift origins (combinations of vertical height and asymmetry) to a common destination using a squatting lifting technique with and without the use of either exoskeleton. Dependent measures included subject kinematics, moment arms between the torso or weight being lifted and the lumbar spine, and spinal loads as predicted by an electromyography-driven spine model. One of the exoskeletons tested (StrongArm Technologies™ FLx) reduced peak torso flexion at the shin lift origin, but differences in moment arms or spinal loads attributable to either of the interventions were not observed. Thus, industrial exoskeletons designed to control posture may not be beneficial in reducing biomechanical loads on the lumbar spine. Interventions altering the external manual materials handling environment (lift origin, load weight) may be more appropriate when implementation is feasible.

1. Introduction

Low back disorders (LBDs) remain prevalent for workers around the globe and carry a significant social and economic burden (NRC, 2001). In the United States, 80% of the population will suffer low back pain (LBP) at some point in their lifetime (Andersson, 1997). LBDs have become a leading reason for physician visits, hospitalization, and utilization of other health care services (Andersson, 1999) and are the cause for approximately 149 million lost work days per year (Guo et al., 1999). LBDs also carry a large economic burden, with the annual direct cost of treatment totaling over \$100 billion in the United States alone (Katz, 2006).

Though LBDs are prevalent in a variety of occupational environments, jobs involving manual materials handling remain among the riskiest. As such, several interventions have been made available to assist workers in their occupational activities. Some of these include lift tables, cranes, balancers, and other lift assist devices (Lavender et al., 2013). These devices can be beneficial for the workers but may also have drawbacks in that they can be costly, space consuming, and underutilized if the loads to be lifted fall within the capabilities of the worker (Graham et al., 2009). In order to address some of these limitations, industrial exoskeletons have recently been designed and

integrated into various industry settings as a workplace intervention. These exoskeletons enable humans to more safely generate the physical power required for a given task (Bosch et al., 2016; de Looze et al., 2016).

While some “active” exoskeletons contain one or more actuators that assist the human body by actively augmenting power using batteries or electric cable connections (de Looze et al., 2016; Gopura and Kiguchi, 2009; Lee et al., 2012), the majority of exoskeletons designed for industrial work are considered “passive.” Passive devices are more readily adopted due to their lower cost and ease of implementation into occupational environments. The appearance and functions of the commercially available passive devices available are vast, dependent on the supported body part(s) and mechanism being used for support. For example, some passive exoskeletons rely on springs, dampers, or materials capable of storing energy from the movement of the body and releasing it when necessary (Bosch et al., 2016; de Looze et al., 2016). Common to most of these devices is their focus on imposing a restorative force, such as one that aims to return the user to a neutral posture when the torso is flexed.

To date, there have been numerous biomechanical studies using electromyography (EMG) data, kinematic measures, or complex biomechanical modeling techniques to evaluate passive exoskeletons that

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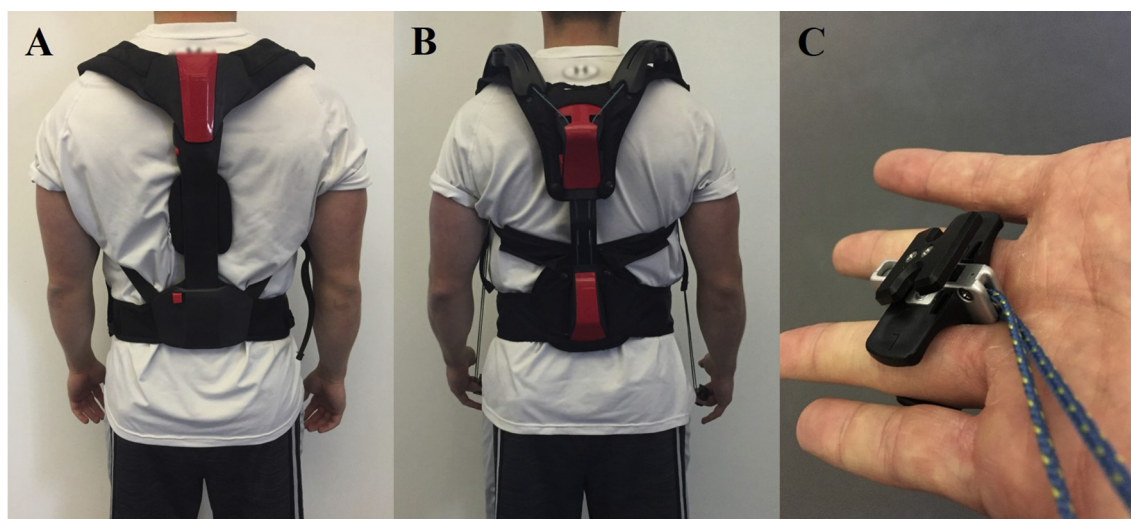


Fig. 1. Photos showing postural assist exoskeletons tested including (A) the FLx (B) the V22 and (C) the hand effector of the V22.

provide a restorative force to the user in some capacity (Abdoli-E et al., 2006; Abdoli-E and Stevenson, 2008; Abdoli-Eramaki et al., 2007; Bosch et al., 2016; Frost et al., 2009; Godwin et al., 2009; Graham et al., 2009; Heydari et al., 2013; Lotz et al., 2009; Ulrey and Fathallah, 2013a; b; Wehner et al., 2009; Weston et al., 2018). However, no studies have evaluated an even simpler class of passive exoskeletons, one that does not provide any restorative force to the user. Relying instead on assisting its users to adopt more favorable postures in an attempt to reduce biomechanical risk, this class of exoskeleton is less expensive than other passive exoskeletons on the market, making them a potentially attractive purchase for companies. However, implications surrounding the use of this latter type of passive exoskeleton are not as well understood. Given, too, that the influence of lifting technique (such as stoop vs. squat) on low back pain outcomes is still a matter of debate (Dreischarf et al., 2016; Hsiang et al., 1997; van Dieen et al., 1999), it remains unclear what benefit passive exoskeletons designed solely for postural guidance might have on biomechanical risk measures.

Thus, the objective of this study was to employ a complex biomechanical model to understand the effects of two postural support exoskeletons on subject kinematics and biomechanical loading of the lumbar spine during a controlled lifting task. Given that subjects would experience the same load regardless of whether either exoskeleton was being worn or not worn (no restorative force), it was expected that only subtle (if any) improvements to biomechanical measures would be observed.

2. Methods

2.1. Approach

Two different commercially available exoskeletons (details below) were evaluated and compared to a no exoskeleton-use condition as subjects lifted from several typical origins to a common destination in a laboratory setting. Basic biomechanical measures included joint flexion angles and the horizontal moment arm from the torso and load being lifted to L5/S1. Research has demonstrated that tissue loading logic provides a more clear picture of injury risk than assessing EMG activity or posture alone (Marras, 2012), so assessing exoskeleton effectiveness from a complex biomechanical tissue loading perspective was preferable in this investigation. Thus, an EMG-driven dynamic biomechanical spine model was also employed to evaluate peak spinal loads in compression, anterior/posterior (A/P) shear, and lateral shear along the length of the lumbar spine extending from T12/L1 to L5/S1. This

biomechanical model is well validated and has been described extensively in the literature (Dufour et al., 2013; Granata and Marras, 1993, 1995; Hwang et al., 2016a, 2016b, 2017; Marras and Granata, 1997). It relies on subject-specific anthropometry, MRI-derived muscle locations and sizes (Jorgensen et al., 2001; Marras et al., 2001), full body kinematics, kinetics, muscle activity for the power producing muscles of the torso, and tissue material properties as model inputs to ultimately predict dynamic tissue loads.

2.2. Subjects

Ten male subjects were recruited locally (mean age 24.9 ± 5.0 years (SD), range 22–38 years; mass 81.1 ± 16.1 kg, range 63.4–102.7 kg; height 179.4 ± 4.6 cm, range 172.1–186.4 cm). This sample size was deemed appropriate via a power analysis with a power of 0.95 using a one side *t*-test. Subjects recruited for this study reported neither any LBDs nor cases of low back pain in the past 3 years nor any prior low back surgeries. Subjects gave informed consent per a study protocol approved by the University Institutional Review Board.

2.3. Study design

As mentioned in 2.1 (Approach), two commercially available exoskeletons were evaluated in this investigation. The first device tested was a postural assist device properly named the FLx; this exoskeleton (size medium, length 40.7–48.8 cm, weight 1.08 kg) has a rigid plastic rod that extends up the length of the back of its wearer and is worn on the body similar to a backpack, with two torso straps and a hip strap (Fig. 1A). This device was designed to remind the wearer to use proper lifting techniques in order to reduce the risk of injury, primarily by discouraging both extensive torso flexion and twisting, which are known risk factors for LBDs (Marras et al., 1993, 1995). When standing in a neutral posture, the FLx remains unengaged; however, as its user bends forward or twists, the device applies pressure on the user's back as feedback and a reminder to return to a more neutral posture. The second device tested, properly named the V22 (size medium, length 41.9–52.1 cm, weight 1.29 kg), is similar to the FLx but the notable difference between the two is that the V22 also contains cables extending from the shoulders (Fig. 1B). These cables terminate at two effectors worn on the hands (Fig. 1C) between the middle and ring fingers that were designed to lock the cables into place as the user of the device lifts and carries a particular load. As such, the intent of this device is to not just serve as a postural assist device but also to transfer biomechanical loads from the upper body; the exoskeleton was

designed to transfer loads (up to 68 kg) via the cables through the shoulders and the rigid back of the device itself down to the legs. Both lift assist interventions tested are manufactured by StrongArm Technologies™ (Brooklyn, NY, USA.).

A $3 \times 3 \times 2 \times 2$ repeated measures design was employed in this study, in which all combinations of intervention (FLx exoskeleton, V22 exoskeleton, none), lift origin height (shin, knee, waist), lift origin asymmetry (0° and 45°), and load weight (9.07 kg and 18.14 kg, or 20 and 40 lbs.) were evaluated. To control for potential order effects, the intervention variable was counter-balanced (control, V22 and FLx) across subjects. Within each block of intervention, the remaining tasks were randomized based on lift origin, asymmetry, and weight. Two repetitions of each condition were collected, back to back.

2.3.1. Independent variables

Main effects of intervention, lift origin height, lift origin asymmetry, and load weight were assessed. As intervention was the primary variable of interest in this investigation, two-way interaction effects with intervention were also assessed as independent measures. Main and interaction effects involving intervention were deemed to be the most relevant discussion points, as it is already known that the other independent variables tested influence the dependent measures (Davis and Marras, 2005; Marras et al., 1997, 1999).

2.3.2. Dependent variables

Dependent measures included subject kinematics, horizontal moment arms from the L5/S1 joint, and three-dimensional spinal loads (compression, A/P shear, lateral shear) as predicted by the aforementioned lumbar spine model. Kinematic measures included peak sagittal flexion angles of the torso, hip, and knee throughout the duration of each exertion. Moment arm measures included the peak horizontal distances between both the center of the mass (COM) of the torso to the L5/S1 joint and the midpoint of the hands to the L5/S1 joint, serving as indicators of the moment placed onto the lumbar spine due to the mass of the torso and the weight being lifted, respectively. Finally, as spinal loads are generally correlated across lumbar levels, the dependent measures of interest in regard to spinal loading were the peak loads in compression, A/P shear, and lateral shear assessed at the endplate in which the highest spinal loads were observed for that dimension of loading.

2.4. Apparatus and instrumentation

2.4.1. Apparatus

As mentioned, the exoskeleton devices tested included the FLx and V22 exoskeletons manufactured by StrongArm Technologies™ (Brooklyn, NY, USA.). In terms of the experimental setup, a height configurable lift table was placed in front of all subjects, which served as the lift origin; this lift table was fan-shaped in order to accommodate both symmetrical and asymmetrical lift origins. Subjects lifted an open-topped box of dimensions 30 cm (width) x 28 cm (depth) x 22 cm (height) filled to the appropriate weight; the box lacked handles, as it was necessary to pick up the items without handles in order for the V22 exoskeleton's hand effectors to engage.

2.4.2. Instrumentation

EMG data was obtained bilaterally for the 10 main power producing muscles of the trunk, including the left and right latissimus dorsi, erector spinae, internal oblique, external oblique, and rectus abdominis. EMG data were captured at 1000 Hz using bipolar surface electrodes and a wireless Trigno™ system (Delsys, Natick, MA, USA). The EMG data signals were notch filtered at 60 Hz and its aliases and band-pass filtered between 30 Hz and 450 Hz. The signals were then rectified and smoothed using a fourth order low pass filter with a cutoff frequency of 1.59 Hz (chosen from a time constant of 100 ms). Body segment kinematics were recorded via a 42-camera Prime 41 OptiTrack optical

motion capture system (NaturalPoint, Corvallis, OR, USA) at a 120 Hz sampling rate. Subjects stood on two 6090-15 six-axis force plates (Bertec, Worthington, OH, USA) recording ground reaction forces for the entirety of each session. All signals were simultaneously collected using custom laboratory software developed in Matlab (Mathworks, Natick, Massachusetts, United States) and synced with a data acquisition board (USB-6225, National Instruments, Austin, TX, USA).

2.5. Procedure

Subjects arrived at the laboratory, were given a brief overview of why the study was being conducted, and were given the opportunity to ask any outstanding questions regarding the study. After providing informed consent, anthropometric measures were then collected, including stature, weight, waist circumference, and the breadth and width of the torso at the level of the sternum and umbilicus. EMG sensors were then placed in accordance with standard placement guidelines (Mirka and Marras, 1993) on the aforementioned 10 torso muscles. A total of 41 reflective optical motion capture markers were then placed on the body of each subject following a marker set prescribed by OptiTrack's motion capture software. Markers were also placed on both of the force plates in order to track their location relative to the subject throughout the session. After sensor placement was complete, the model was calibrated for each subject using data derived from a series of dynamic concentric and eccentric lumbar motions while holding a 9.07 kg medicine ball. This calibration technique has been described previously (Dufour et al., 2013) and does not require maximum voluntary contractions (MVCs). Before any experimental conditions were collected, subjects were also asked to stand on the force plate at a distance as close as possible to the box that was to be lifted without interference from the configurable lift table. Once this foot placement was determined for each subject, it was kept constant for the rest of the study.

Both the FLx and V22 exoskeletons were designed to be used in conjunction with (not to replace) proper training, which the manufacturer denotes to be a squatting lifting technique. Thus, all subjects were trained on this lifting technique prior to collection of any experimental conditions by one of the researchers (MP) and were encouraged to implement this squatting lifting technique throughout the duration of the study. It should be noted, too, that prior to donning either of the exoskeletons that were being tested, subjects also watched instructional videos on each particular device detailing how to don and wear the device correctly and how to properly lift while wearing the device; these videos were produced by the exoskeleton manufacturer. As subjects donned a new device, the researchers adjusted it based on manufacturer recommendations in both size and device flexibility. Once the device was equipped, the subject was able to become familiar with the device for approximately 10 min before any randomized lifting trials began.

Subjects were instructed to start each trial in a relaxed upright position with their arms at their sides. The experimental task involved lifting the box at a comfortable pace from the given lift origin determined from the study design (combination of height and asymmetry) to a common destination in front of the body at waist height. During asymmetric lifts, subjects' feet were kept at the previously mentioned location, and trunk rotation was required to lift the box; these instructions were chosen so that the asymmetry conditions represented more of a 'worst-case' scenario. The box which contained the load was also raised 3.81 cm (1.5 inches) off of the surface of the lift table for all trials. This allowed for the hand effectors of the V22 to easily slide under the box, which were required to be engaged to utilize the device correctly. The box was raised for all trials for appropriate comparability among conditions.

Table 1

The significance level of the effects of intervention, lift origin, asymmetry, load weight, and their interactions on subject kinematics, moment arms, and spinal loads (* $p < 0.05$, ** $p < 0.01$, *** $p < 0.001$).

	Intervention (I)	Lift Origin (O)	Asymmetry (A)	Load Weight (W)	I*O	I*A	I*W
Kinematics							
Sagittal Torso Angle	*	***		**	**		
Sagittal Hip Angle		***	**	***			
Sagittal Knee Angle		***	**	**			
Moment Arms							
Torso COM to L5/S1	**	***	**	**			
Hand COM to L5/S1		***		***			*
Spinal Load							
L3/L4 Superior Compression		***		***	*		
L5/S1 Superior A/P Shear		***		***			
L5/S1 Inferior Lateral Shear		**	*	***		*	

2.6. Data processing and statistical analysis

All dependent measures were derived via simulations in a multibody dynamics solver, Adams (MSC Software, Santa Ana, CA, USA), using the aforementioned EMG-driven lumbar spine model. Results were analyzed using JMP Pro 13 software (SAS Institute Inc., Cary, NC, USA). A repeated measures 2-way analysis of variance (ANOVA) with a significance level of 0.05 was used to assess the effects of the independent measures and two-way interactions on the dependent measures. Post-hoc analyses were conducted using a Tukey HSD test where appropriate.

3. Results

Statistically significant main and interaction effects for the dependent measures investigated are presented in Table 1. Lift origin and load weight impacted each of the dependent measures assessed, with fewer main and interaction effects involving intervention reaching statistical significance.

3.1. Kinematics

Kinematic measures consisted of the peak sagittal joint flexion in the torso, hip, and knee throughout the duration of each trial, as shown in Fig. 2 (as lift origin most affected the magnitude of the dependent measures, the data in this figure and in subsequent figures have been stratified by lift origin and intervention). Given that no significant differences between the right and left sides of the body were observed, results for just the right hip and right knee are presented herein. All three dependent measures were influenced by both lift origin and load weight, in which higher magnitudes of peak flexion were observed in all three joints for lower lift origins (i.e., shin > knee > waist) ($p < 0.001$) and higher magnitudes of peak flexion in all three joints were observed for the heavy load compared to the light load ($p < 0.006$). While asymmetry did not affect peak torso flexion, hip flexion was increased for the asymmetric lift origin (45°) relative to the sagittally symmetric lift origin ($p = 0.009$); however, the opposite effect was observed for peak knee flexion ($p = 0.007$).

No effects attributable to intervention (main or interaction effects) were observed for either peak hip flexion or peak knee flexion. However, there was both a main effect of intervention and an interaction effect of intervention with lift origin observed for peak torso flexion. The main effect of intervention that was observed suggested that peak torso flexion angles observed for the control condition did not vary significantly between either of the devices tested (rather, this effect was significant because increased torso flexion was observed for the V22 relative to the FLx, $p = 0.026$). The intervention * lift origin interaction effect observed showed that when stratifying the data by lift origin, peak torso flexion was decreased at shin height with use of the FLx relative to the control condition or V22 ($p = 0.0015$).

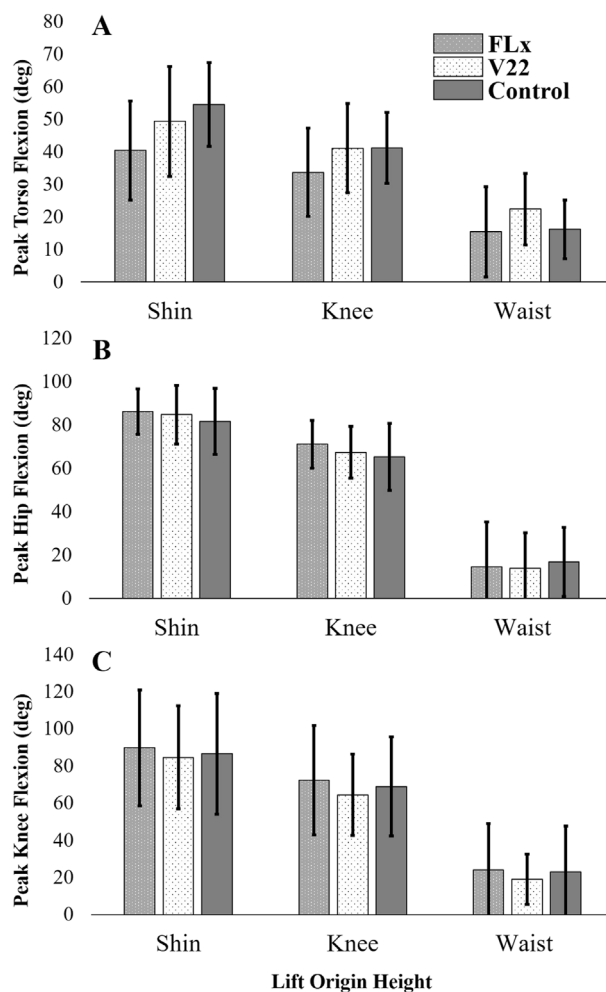


Fig. 2. Peak joint flexion angles in (A) the torso, (B) the right hip, and (C) the right knee as a function of intervention level (FLx, V22, control) and lift origin (shin, knee, waist). Error bars denote standard deviation.

3.2. Moment arm

The peak sagittal moment arm between the torso COM and L5/S1 throughout each trial was increased with use of the V22 exoskeleton relative to the control condition or FLx ($p = 0.010$) (Fig. 3). This dependent measure was also influenced by lift origin, in which the greatest moment arms were seen for the shin lift origin, then the knee lift origin, and finally the waist lift origin ($p < 0.001$). The peak moment arm between the torso COM and S1 was also increased under asymmetric lifting conditions ($p = 0.0016$) and increased for the heavy

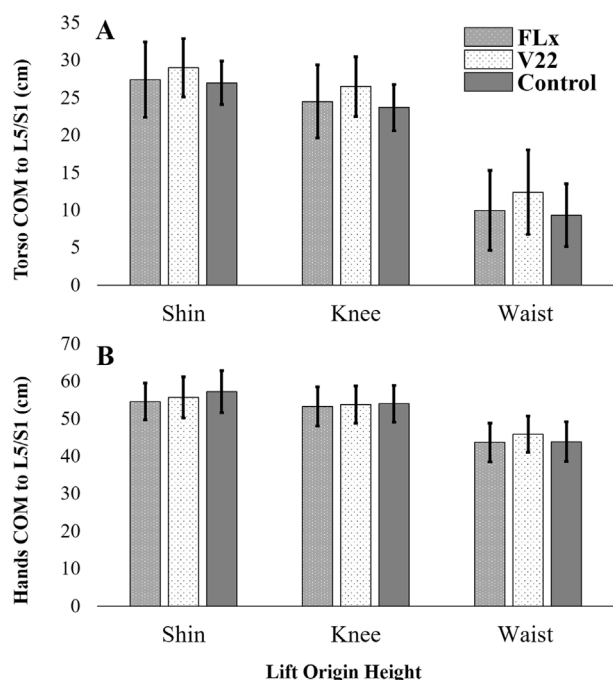


Fig. 3. Peak horizontal moment arm distances between (A) the center of mass of the torso and L5/S1 and (B) the center of mass of the hands and L5/S1 throughout each lifting exertion. These moment arm distances are indicative of external moments placed on the lumbar spine due to the mass of the torso and the load being lifted, respectively. Error bars denote standard deviation.

weight compared to the light weight ($p = 0.0024$).

The peak sagittal moment arm between the COM of the hands and L5/S1 throughout each trial was also increased for the shin lift origin relative to the knee and waist origins ($p < 0.001$) and increased when lifting the heavier load compared to the lighter load ($p < 0.001$). Though no main effect of intervention was observed for this dependent measure, a significant intervention * load weight interaction effect was observed. It is important to note, however, that this statistically significant interaction effects carries little biomechanical significance. While the interaction effect can be attributed to the fact that the effect of intervention on this moment arm distance was inconsistent across load weights, post-hoc analysis revealed that even when stratifying the data by load weight, no significant differences were observed among the devices tested and the control condition.

3.3. Spinal loads

The biomechanical spine model employed predicted the highest magnitude of spinal loading for compression and shear (A/P and lateral) at the L3/L4 and L5/S1 vertebral levels, respectively. Any differences observed between either intervention and the control condition are marginal compared to the overall magnitude of loading in both compression and A/P shear. Lateral shear values were generally low in magnitude across all three intervention levels and were subsequently deemed to pose little biomechanical significance as compared to the results for compression or A/P shear. Results are shown in Fig. 4.

As was previously mentioned, main and interaction effects involving the intervention were of most interest in this investigation. There was no statistically significant main effect of intervention observed for peak spinal loading in any dimension (compression, A/P shear, lateral shear), though there was a significant intervention * lift origin interaction effect observed for peak spinal compression and a significant intervention * asymmetry effect observed for peak lateral shear. However, like the intervention * load weight interaction observed previously for the peak hand COM to L5/S1 moment arm, post-hoc

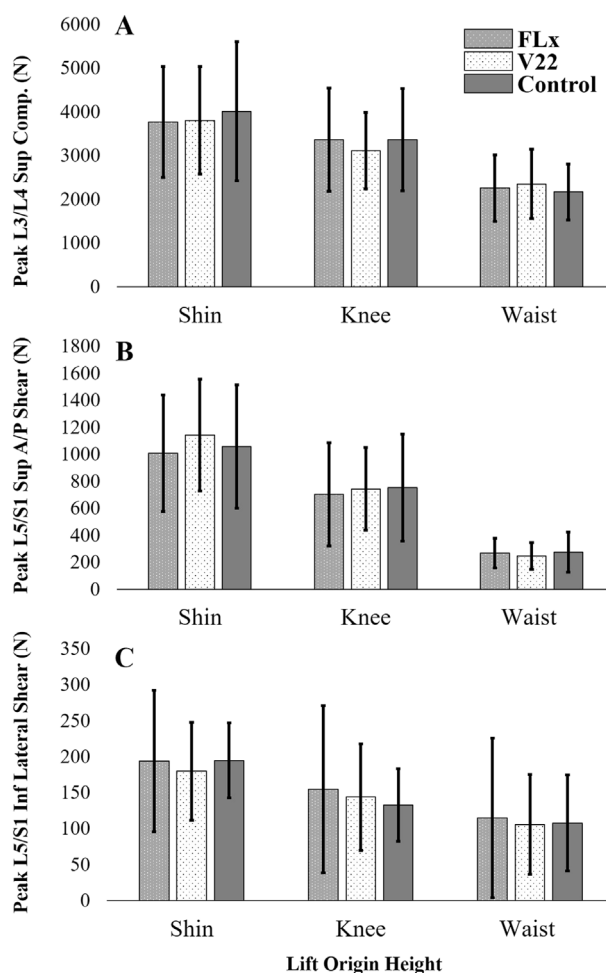


Fig. 4. Peak spinal loads in (A) compression (L3/L4 Sup endplate), (B) A/P shear (L5/S1 Sup endplate), and (C) lateral shear (L5/S1 Inf endplate) as a function of intervention level (FLx, V22, control) and lift origin (shin, knee, waist). Error bars denote standard deviation.

analysis showed no statistically significant differences in peak loads among the intervention levels at any of the individual lift origins (compression) or asymmetry levels (lateral shear).

The other independent variables assessed in this study affected peak spinal loads as might be expected. Peak spinal loads were increased in all dimensions (compression, A/P shear, lateral shear) with the heavy load compared to the light load ($p < 0.001$), though asymmetry affected only lateral shear loading, where increased peak lateral shear loads were observed for the asymmetric lift origin ($p = 0.024$). Peak spinal compression and A/P shear loads for each of the three lift origins differed significantly from one another, where peak loading increased as the lift origin moved closer to the ground ($p < 0.001$), whereas peak loads differed significantly only between shin and waist lift origins in terms of lateral shear ($p = 0.0011$).

4. Discussion

While several industrial exoskeletons aimed at reducing the risk of LBDs are commercially available and are continually being introduced into occupational environments, there remains a void in the scientific literature relative to the effectiveness of exoskeletons intended to provide postural guidance in reducing low back injury risk. The main focus of this study was to investigate the relative differences in kinematics, moment arms, and peak spinal loading among two postural assist exoskeletons and a control condition. This study found that neither

device offers significant advantages relative to the control condition from a biomechanical perspective.

Central to the design of both devices tested is that encouraging more upright torso angles will result in a lower risk of injury to the lumbar spine. By encouraging more upright torso postures, the moment arm between the lumbar spine and the center of the mass of the torso is expected to be reduced. Subsequently, the external moments placed onto the spine from the weight of the torso segment are also expected to be reduced, as are the demands on the trunk extensor muscles and biomechanical loads acting on the lumbar spine. The FLx was designed to keep its wearers more upright by providing feedback (and encouraging correction) when lifting is performed with either extensive torso flexion or trunk twisting, which are known risk factors for LBDs (Marras et al., 1993, 1995). The V22 was very similar to the FLx, though it also had cables that the manufacturer suggests were designed to transfer loads through the shoulders and the rigid back of the device itself toward the hips and legs. Among the conditions tested, mean peak sagittal torso flexion was unchanged between the FLx and control conditions at knee and waist lift origins. However, the results of this investigation showed that the FLx significantly reduced peak torso angles of its wearers when lifting from shin height, reducing mean peak sagittal torso flexion by 14.2° relative to the control condition. Peak torso flexion was not decreased with use of the V22 relative to the control condition. This difference between the devices is likely attributable to the hand effectors on the V22 exoskeleton. In order to engage the hand effectors effectively, additional torso flexion was required of the subjects.

As the mass of the torso and weights being lifted were consistent or comparable across conditions, examination of the peak moment arms of the torso or load being lifted from the lumbar spine are indicative of external moments acting on the lumbar spine. It should be noted that despite the fact that beneficial changes in torso kinematics were observed (particularly for the FLx), neither exoskeleton device offered a benefit when examining moment arms or peak spinal loads; the horizontal moment arm between the torso COM and lumbar spine effectively remained the same across conditions with use of the FLx and increased across conditions with use of the V22. Likewise, the moment arm between the COM of the hands and the lumbar spine was also unaffected by either device. This rather unexpected result could be attributed to changes in kinematics of joints other than the torso, as has been shown previously. Sparto et al. (1998) investigated a lifting belt and found tradeoffs between trunk and hip angles and velocities while wearing it. Marras et al. (2000b) obtained a similar result when examining the effects of an elastic back support. In that study, changes in torso kinematics were also accompanied by increased flexion in the hips and knees that ‘washed out’ the benefits of reduced torso flexion; neither external moments on the lumbar spine nor spinal compression decreased. Though not a statistically significant result in this study for the population as a whole, it was apparent that over half of the subjects increased hip flexion upon decreasing their torso flexion angle, just as was seen by Marras et al. (2000b). Though not assessed as dependent variables of interest in this study, kinematics in the upper extremity could have also been altered in response to wearing either device.

Several other examples denote that improved posture does not necessarily correlate to lower spinal loading. For example, studies that were conducted on lumbar back supports (back belts) in their height of popularity found that a decrease in poor posture did not always reduce the loading on the spine (Granata et al., 1997; NIOSH, 1994; Reddell et al., 1992). A review of biomechanical studies related to lifting technique and low back loading by van Dieen et al. (1999) further illustrates this point. Upon comparing squatting and stooping lifting techniques, this study found that net moments and spinal compression forces based on model estimates were about the same if not somewhat higher for squatting, suggesting that there is little evidence for advocating one lifting technique over another. Yet another review article by Hsiang et al. (1997) concludes that though lifting has been

associated causally with low back pain outcomes, no studies have directly linked lifting technique to LBDs. It is clear, then, that kinematic measures alone are just one of many factors influencing biomechanical loads placed onto the lumbar spine.

Ultimately, from a biomechanical standpoint, neither of the devices tested considerably reduced the loading on the spine. The extensor muscles still needed to counteract large moments generated from loads in front of the body from the torso and weight being lifted, which is why there was not a notable difference in compression or shear among the three conditions. These results also demonstrate that at this time, it may still be most beneficial to implement engineering interventions related to the external working environment itself, particularly interventions that raise the lift origin of the load (such as a lift table or scissor lift) or reduce the weight of the load being lifted (like an overhead crane or hoist). In this study, the largest changes observed in peak spinal loads were related to these factors. For example, peak compression and peak A/P shear loads decreased by 42% and 75% respectively between the shin and waist lift origins conditions. Moreover, prior studies have already shown reduction in injury risk to the lumbar spine with use of these types of interventions (Faber et al., 2009; Marras et al., 2000a; Plamondon et al., 2012). Given, however, that it may not always be feasible to alter the external work environment due to monetary or physical constraints, practitioners should remain hopeful that other passive or active exoskeletal interventions will be better designed from a biomechanical standpoint such that their use could offer increased versatility across a multitude of work situations.

As with any investigation, there were limitations present in this study. First, this study was performed in a controlled laboratory setting. As such, the results of this study are specific to the conditions tested, and practitioners should be careful when extrapolating the results of this study to lifting scenarios outside of the lift origin range (shin-waist), asymmetry range (0–45° relative to sagittal plane), or load weight range (9.07–18.14 kg) tested. Additionally, all of the subjects were males, were inexperienced with manual materials handling operations, and were trained on and instructed to use a squatting lifting technique for all trials. Allowing subjects to adopt a self-selected lifting technique could have led to more noticeable differences in the dependent measures, particularly in subject kinematics. That being said, the manufacturer is clear in their recommendations for use that both exoskeletons should be used in *combination with* training and what they consider to be a proper lifting technique (squatting). Thus, testing the devices without training and enforcement of a squatting technique could have artificially inflated the apparent effectiveness of the device, as neither exoskeleton was designed to replace training altogether.

Finally, in some regards, the external validity of the study is limited by the fact that the experimental task performed represents a ‘best case’ scenario for the exoskeletons tested. For example, the box holding the load was raised off of the lifting surface by 1.5” to allow for subjects to directly slide their hands under the box to pick it up. This was set up this way so that when subjects were equipped with the V22, they could instantly engage the hand effectors. In reality, this is an unrealistic positioning of how loads would be lifted in an industrial setting. Without the box being raised, the V22 would likely be even less effective. In other ways, however, the experimental task did represent a ‘worst case’ scenario. The subjects’ feet were confined on the force plate for each lift. Subjects twisted to perform asymmetric lifts as opposed to turning their body to face the load, counter to recommendations made by the manufacturer of the devices tested. However, in a typical industry setting where the workers are under time restrictions to move objects, workers may be more likely to twist and grab the object then pivot their feet back and forth depending on the space available to them.

5. Conclusion

Upon enforcing a squatting lifting technique across all conditions

consistent with manufacturer recommendations, the two postural assist exoskeletons tested provided no significant biomechanical benefit (compared to a non-intervention control condition) in regard to joint flexion angles, moment arms between a given load and the L5/S1 joint, or biomechanical loads on the tissues of the lumbar spine. While these results are specific to the two exoskeletons tested in this study and may not be applicable to all industrial exoskeletons, these results suggest that interventions that alter the external environment of the lift itself (i.e., lift tables or scissor lifts) may be more beneficial to LBD risk reduction than postural assist exoskeletons if implementation of this type of intervention is feasible.

Conflicts of interest

The authors have no competing interests to declare.

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