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Biomechanical evaluation of exoskeleton use on loading of the lumbar spine

Eric B. Weston[a,b,*], Mina Alizadeh[a,b], Gregory G. Knapik[a,b], Xueke Wang[b], William S. Marras[a,b]

[a] Spine Research Institute, The Ohio State University, Columbus, OH, United States
[b] Department of Integrated Systems Engineering, The Ohio State University, Columbus, OH, United States

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ABSTRACT

The objective of this study was to investigate biomechanical loading to the low back as a result of wearing an exoskeletal intervention designed to assist in occupational work. Twelve subjects simulated the use of two powered hand tools with and without the use of a Steadicam vest with an articulation tool support arm in a laboratory environment. Dependent measures of peak and mean muscle forces in ten trunk muscles and peak and mean spinal loads were examined utilizing a dynamic electromyography-assisted spine model. The exoskeletal device increased both peak and mean muscle forces in the torso extensor muscles ($p < 0.001$). Peak and mean compressive spinal loads were also increased up to 52.5% and 56.8%, respectively, for the exoskeleton condition relative to the control condition ($p < 0.001$). The results of this study highlight the need to design exoskeletal interventions while anticipating how mechanical loads might be shifted or transferred with their use.

1. Introduction

Work-related musculoskeletal disorders (MSDs) continue to represent a major problem in modern occupational environments. Among all MSDs reported, low back disorders (LBDs) and shoulder injuries are by far the most prevalent (Holmstrom and Engholm, 2003; Widanarko et al., 2012, 2014; Wijnhoven et al., 2006). The Bureau of Labor Statistics indicates that between 2014 and 2015, work-related MSDs resulted in a median of 12 lost work days per incident, with low back and shoulder complaints making up 40% and 15% of the total cases, respectively (BLS 2016). These MSDs represent an immense economic burden, in which the direct cost of treatment of LBDs annually in the United States totals over $50 billion (Davis et al., 2012), and the direct cost of treating shoulder injuries totals to over $7 billion (Meislin et al., 2005).

Though workers in occupational environments are exposed to a wide range of exposures, the effects of using heavy hand tools to perform tasks such as drilling, countersinking, riveting, bucking, and swaging has received considerable attention. Hand tools may need to be used in unfavorable postures such as is seen in overhead work, asymmetric exertions, or kneeling (Burdorf et al., 1991). It is no surprise, then, that workers subjected to hand tool use have noted high rates of low back and shoulder injuries (Stenlund et al., 1993; Keyserling et al., 1991).

In response to musculoskeletal complaints related to hand tool use, various interventions have been introduced into occupational environments, including cranes and other lift assist devices. Unfortunately, among their many advantages, these devices also have significant costs. Their use can be both time and space consuming, and workers tend not to use them if loads fall within their strength capacity or if extensive learning is required (Graham et al., 2009). As a result, wearable exoskeletons have recently been introduced as an alternative workplace intervention. Exoskeletons are a type of mechanical intervention that are designed to work in concert with the worker in order to provide support or enhance their capabilities, and it is conceivable that exoskeletons could make a greater impact than existing interventions. In fact, previous investigations have already shown that exoskeletal devices can be helpful in reducing the sum of joint torque in the upper arm (Sylla et al., 2014) or decreasing the effective load on the shoulder (Naito et al., 2007).

Despite the fact that numerous work-related exoskeletons are commercially available and have already been introduced into some occupational environments, there has been relatively little research examining the potential benefits, drawbacks, and trade-offs of exoskeleton use in an occupational workplace. Likewise, most of the studies that have been conducted are limited in some capacity by the methods used. A number of studies used electromyographic (EMG) data to evaluate the impact of exoskeleton use, but the EMG data oftentimes was not normalized or modulated for muscle length and velocity or was just averaged across subjects (Abdoli-Eramaki et al., 2006; Bosch et al., 2016; Graham et al., 2009; Naito et al., 2007; Kobayashi and Nozaki, 2007). Numerous studies examined kinematic measures, but these were frequently confined to just the sagittal plane or purely static assessments (Abdoli-Eramaki et al., 2006; Abdoli-Eramaki et al., 2007;

* Corresponding author. 1971 Neil Avenue Rm 512, Columbus, OH 43210, United States.
E-mail address: weston.101@osu.edu (E.B. Weston).

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Heydari et al., 2013; Ulrey and Fathallah, 2013a, 2013b; Graham et al., 2009; Bosch et al., 2016; Sadler et al., 2011). Finally, those studies that used a biomechanical modelling approach often used models that were static or were unable to account for passive muscle forces and muscle coactivity (Abdoli-Eramaki and Stevenson, 2008; Frost et al., 2009; Graham et al., 2009; Heydari et al., 2013; Naruse et al., 2003; Panizzolo et al., 2016; Ulrey and Fathallah, 2013a, 2013b; Abdoli-Eramaki and Stevenson, 2008; Wehner et al., 2009).

Moreover, to the authors' knowledge, no studies have examined biomechanical loading to other joints that the exoskeletons were not specifically designed to support. For example, several industries including shipbuilding and aerospace manufacturing have considered the use of or have already implemented exoskeletal interventions that were specifically designed to mitigate risk to the upper extremities (particularly the shoulders) during powered hand tool use. However, it remains unclear if these interventions simply sacrifice risk elsewhere, such as the low back. Back problems in particular represent the most disabling medical condition to affect mankind worldwide (Hoy et al., 2014) and are the primary reason workers under the age of 45 have activity limitations (Marras, 2008). Thus, the impact of exoskeleton use on the low back should at the very least be investigated, even if the exoskeleton was designed for applications in which the upper extremity is the main concern.

There is a significant void in the body of knowledge concerning the use of work-related exoskeletons that should be addressed before this class of intervention is safely employed in occupational environments. Thus, the objective of this study was to employ a biomechanical model to evaluate how the use of an exoskeletal intervention affects muscle force and spinal load measures in the low back during simulated hand tool use in a laboratory environment.

2. Methods

2.1. Approach

A laboratory study was conducted in an attempt to understand the biomechanical impact of using an exoskeleton during occupational work. In this case, a mechanical arm was connected to an exoskeletal vest to support a power tool during a simulated work task. Muscle forces in the power-producing muscles of the torso and lumbar spinal loads in compression, anterior/posterior (A/P) shear, and lateral shear were evaluated for two different tools with and without the exoskeletal vest and arm using an EMG-assisted dynamic biomechanical spine model. The biomechanical model employed has been described extensively in the literature with numerous publications documenting its implementation and validation (Dufour et al., 2013; Granata and Marras 1993, 1995a, 1995b, 2000; Marras and Granata, 1997; Granata et al., 1999; Knapik and Marras, 2009; Knapik et al., 2012; Marras et al., 2001, 2004, 2006; Marras et al., 2009). The model has also been recently updated to include curved muscle representations and new personalized active and passive muscle force algorithms (Hwang et al., 2016a, 2016b, 2017).

2.2. Subjects

Twelve male subjects were recruited from the local university population (age 25.3 ± 6.0 years, mass 81.9 ± 9.8 kg, and height 184.4 ± 5.2 cm). All of the subjects provided informed consent and had no reports of previous or current low back pain in the past 3 years and no prior low back surgery. This study was approved by the University's Institutional Review Board.

2.3. Study design

A repeated measures design was implemented for this investigation to evaluate independent measures of exoskeletal intervention (with and without an exoskeletal vest and arm), vertical exertion height (50%, 65%, and 100% of subject stature), and asymmetry (feet oriented at 0° and 45° away from the tool). The experimental design was carried out for each subject with two tools, a nutrunner (weight 4.54 kg) and a pneumatic impact wrench (13.61 kg). Selection of the independent variables investigated and their levels were made consistent with suggestions from an industrial partner to represent tasks common to aerospace manufacturing.

The authors note that the combination of trials (with and without the intervention) with the heavy tool at 100% of subject stature was excluded from the study design. Under these experimental conditions, the target vertical exertion height was outside of the vertical range of the mechanical arm of the exoskeletal intervention being tested. Thus, experimental conditions tested for the nutrunner (light tool) were representative of a 2 × 3 × 2 repeated measures design, while experimental conditions tested for the pneumatic impact wrench (heavy tool) were instead representative of a 2 × 2 × 2 repeated measures design.

The order in which the experimental conditions were encountered by subjects were first counterbalanced based upon exoskeletal intervention (with and without). Within each block, conditions were then randomized based upon tool weight, exertion height, and asymmetry. Two repetitions of each experimental condition were collected back to back.

2.3.1. Independent variables

Independent variables included intervention, vertical exertion height, and asymmetry. The main effects attributable to these independent variables as well as potential intervention*height and intervention*asymmetry effects were assessed separately for each of the two tools. Effects that were found to be consistent across both tool weights were determined to be of greatest importance for discussion.

2.3.2. Dependent variables

Dependent measures consisted of peak and mean muscle forces and peak and mean three-dimensional spinal loads for each trial. Muscle forces were estimated for the power-producing muscles of the torso, including the right and left erector spinae (ES), internal oblique (IO), latissimus dorsi (LD), external oblique (EO), and rectus abdominis (RA) muscles. Likewise, three-dimensional spinal loads (compression, A/P shear, lateral shear) were calculated at the superior and inferior endplates of the lumbar spine extending from T12/L1 to L5/S1.

All dependent measures were derived from simulations in the multibody dynamics solver, Adams (MSC Software, Santa Ana, CA, USA). The aforementioned biologically-driven biomechanical model utilized inputs of subject-specific anthropometry, MRI-derived muscle locations and sizes, full body kinematics, muscle activity for the power-producing muscles of the torso, and tissue material properties. Muscle forces were estimated via modulation of EMG activity with a gain ratio determined from model calibration, muscle location and cross-sectional area derived from Magnetic Resonance Imaging (MRI), and force-length and force-velocity relationships of the muscle (Jorgensen et al., 2001; Marras and Granata, 1997; Marras et al., 2001). Likewise, spinal loads were estimated via combination of the muscle force data with whole body kinetic loads, torso cross sectional area, muscle lines of action, muscle moment arms, vertebral angles, and other geometric information.

2.4. Apparatus and instrumentation

2.4.1. Tools

Two different tools were employed for testing. The light tool (4.54 kg) condition utilized a right angle nutrunner (EA34LA19-80, STANLEY Engineered Fastening, New Britain, CT, USA.) The heavy tool (13.61 kg) condition utilized a pneumatic impact wrench (1 in. Industrial Pinless Air Impact Wrench, Central Pneumatic, Camarillo, CA, USA.) These specific tools were selected since they were both at the

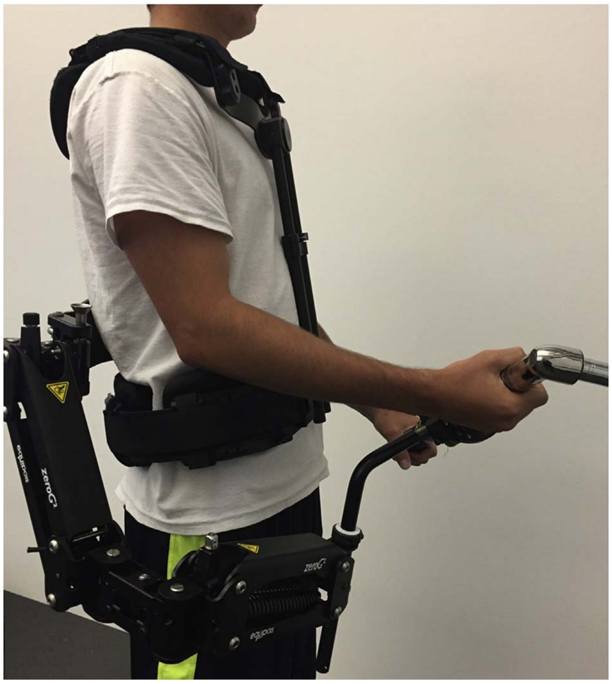

Fig. 1. Sample of the Steadicam Fawcett Exoskeletal vest attached to an Equipois mechanical arm. The vest and arm were used together to support the weight of nutrunner (shown) and impact wrench hand tools.

upper weight range of each mechanical arm's capacity and required a horizontal force exertion during operation.

2.4.2. Exoskeletal vest and mechanical arm

The exoskeletal device tested was initially designed to support camera systems but has recently been incorporated into other occupational environments including shipbuilding and aircraft construction. Subjects wore an exoskeletal vest (Steadicam Fawcett Exoskeletal vest, Tiffen, Hauppauge, NY, USA), as shown in Fig. 1. Two different models of mechanical arms (zeroG2 and zeroG4, Equipois, Manchester, NH, USA) were attached to this vest and used to support the nutrunner and impact wrench, respectfully. The tools were mounted to the mechanical arms using a Saturn series gimbal system (Equipois, Manchester, NH, USA).

2.4.3. Instrumentation

EMG activity was obtained bilaterally for the latissimus dorsi, erector spinae, rectus abdominis, internal oblique, and external oblique. The electromyographic activity of the trunk muscles was collected with surface electrodes and sampled at 1000 Hz (Motion Lab Systems MA300-XVI, Baton Rouge, Louisiana, USA). Signals were notch filtered at 60 Hz and its aliases (up to 480 Hz) and band-pass filtered at 30-450 Hz. The signals were then rectified and smoothed using a fourth-order low pass filter with a cutoff frequency of 1.59 Hz (corresponding to a time constant of 100 ms). Kinetic data were also captured at 1000 Hz. Subjects exerted against a small custom-built six-axis load cell (HT0825, Bertec, Worthington, OH, USA) mounted on a custom-made height and angle adjustable frame. Likewise, subjects stood on a separate six-axis force plate (6090-15, Bertec, Worthington, OH, USA) recording ground reaction forces. Finally, kinematic data of individual body segments were collected via a 42-camera optical motion capture system (Optitrack Prime 41, NaturalPoint, Corvallis, OR, USA) with a 120 Hz sampling rate. All signals were simultaneously gathered with customized laboratory software via a data acquisition board (USB-6225, National Instruments, Austin, TX, USA).

2.5. Procedure

After arriving at the laboratory, subjects were briefed on the study design and allowed to ask any questions they might have regarding the study. Subjects then gave informed consent per University Institutional Review Board guidelines. Anthropometric measures were collected including body mass, stature, trunk circumference, trunk breath, and trunk depth. Subjects then donned the exoskeletal vest, and it was adjusted to fit them correctly according to manufacturer recommendations (Fawcett and Hayball 2013). The vest was then removed, and EMG sensors were placed on the aforementioned muscles of the torso based upon placement guidelines from literature (Mirka and Marras, 1993). Motion capture markers were placed on 41 different locations on the body as set forth by standard placement locations from OptiTrack's motion capture software. Markers were also placed on the force plate and load cell to track their location relative to the subject.

After sensor placement, subjects went through a calibration procedure consisting of a series of dynamic concentric and eccentric lumbar motions and exertions in multiple planes while holding a 9.07 kg weight (Dufour et al., 2013.) This calibration data was used to derive personalized muscle and other model properties for the subject. Separate calibrations were performed with and without the exoskeletal vest on the subject. After calibration, the mechanical arm was attached to the right side of the exoskeletal vest, and the subject was provided training time to get comfortable with the operation of the exoskeletal vest, mechanical arm, and each tool. Adjustments were made to confirm the arm was supporting the tool correctly and to ensure the subject could correctly perform the experimental tasks.

Subjects were instructed to perform tasks naturally within the provided guidelines. For sagittally symmetric exertions (i.e., no task asymmetry), subjects faced their feet forward toward the load cell, as shown in Fig. 2. Likewise, during asymmetric exertions, subjects placed their feet at a forty-five degree angle to the left relative to the load cell toward one of the optical markers embedded within the force plate, all the while focusing their eyes on a target within the room. Thus, though foot placement on the load cell was not strictly controlled, some degree of consistency in foot placement for asymmetric conditions was maintained within and between subjects. Subjects started each trial in a relaxed, upright standing posture while holding the tool. With the beginning of each trial, subjects then placed the tool in position against the load cell, and exerted up to a target horizontal force level of 75 N. This target force level approximates that set forth by a previous study for a simulated drilling task (Alabdulkarim et al., 2017). After reaching the target force level, an audible tone sounded, indicating that subjects could remove the tool and return to a neutral posture. In order to prevent the subject from exerting too rapidly or impacting the tool on the load cell, a rate limiter ($>$ 20 N/s) sounded a separate audible tone indicating when the trial would need to be repeated. During the 100% of stature height conditions, the load cell was oriented facing 45° down from vertical. In all other conditions the load cell was aligned vertically. In each trial, the tool was supported mainly with the right hand.

2.6. Statistical analysis

Results were analyzed using JMP 11.0 software (SAS Institute Inc., Cary, NC, USA). A repeated-measures, three-way analysis of variance (ANOVA) with a significance level (α) of 0.05 was implemented to determine the effects of the independent measures and two-way interactions on dependent measures of peak and mean muscle force and spinal load. It was assumed that a heavier tool weight would result in increased muscle forces and spinal loads; thus, to account for the unbalanced study design employed (no data collected for the heavy tool at the 100% stature vertical exertion height), separate statistical analyses were performed for each tool weight. Post-hoc analyses were performed using Tukey HSD tests where appropriate.

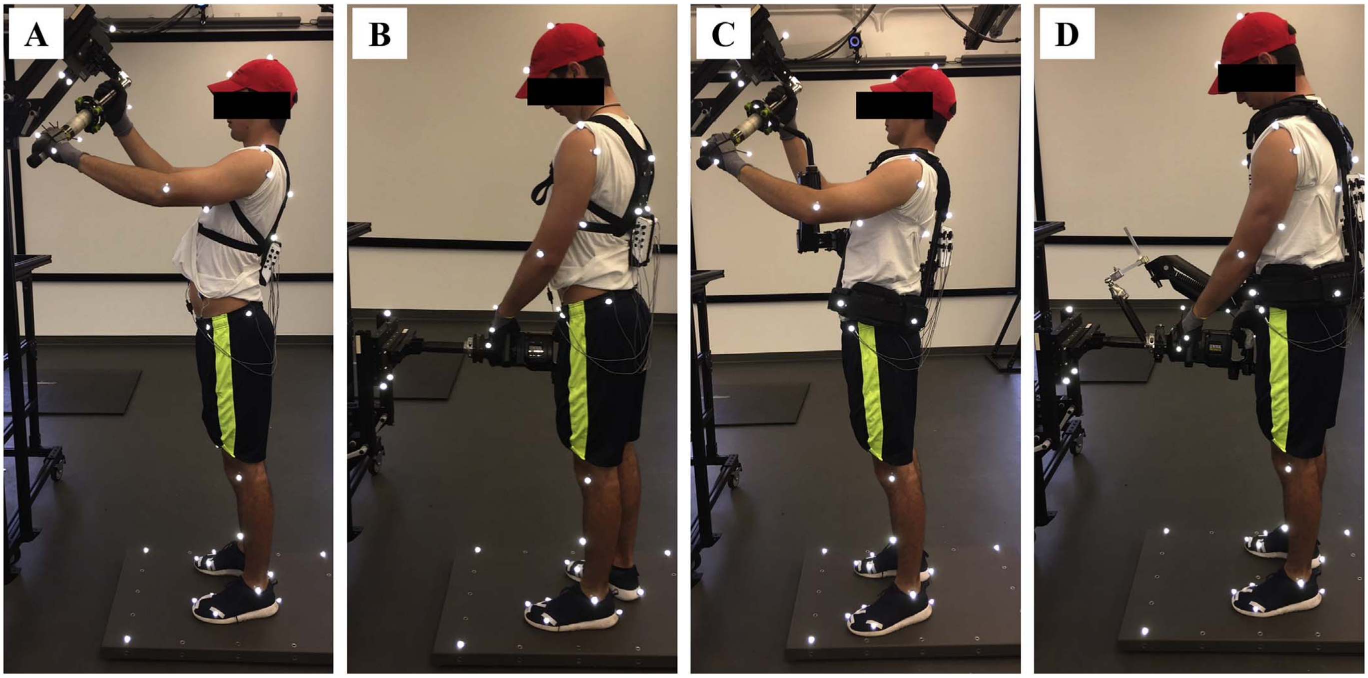

Fig. 2. Photos denoting experimental setup and conditions including **(A)** control condition with light tool at 100% of subject stature height, **(B)** control condition with heavy tool at 50% of subject stature height, **(C)** exoskeleton and arm intervention with light tool at 100% of subject stature height, and **(D)** exoskeleton and arm intervention with heavy tool at 50% of subject stature height. Vertical height and asymmetry were also investigated as independent measures.

3. Results

As was assumed prior to running the statistical analysis, the heavier tool weight generally increased peak and mean muscle forces and peak and mean spinal loads. Likewise, independent variables of exoskeleton condition, exertion height, asymmetry, and combinations of their interactions were all found to significantly affect dependent measures of peak and mean muscle force and peak and mean spinal load. Though statistically significant effects varied slightly dependent on each individual spinal endplate level, the trends observed as a result of each main and interaction effect were consistent across all spinal levels within each dimension of spinal loading. Thus, for improved clarity to the reader, spinal loading results have been presented for just the endplate levels in which the highest spinal loads were observed in each individual direction of spinal loading (compression, A/P shear, lateral shear). The factors and interactions that led to statistically significant results for each muscle and direction of spinal loading across both tool weights are shown in Table 1; effects observed for just one tool weight will subsequently be discussed but are not represented in the table.

3.1. Muscle force

Across both tool weights, peak and mean muscle forces were larger bilaterally in the torso extensor muscles (left erector spinae, right erector spinae, right internal oblique) for the exoskeleton condition relative to the control condition ($p < 0.001$). This effect was most pronounced in the erector spinae muscles. Collectively across both tool weights, the exoskeletal intervention increased peak muscle forces in the left erector spinae by 55.5%, peak muscle forces in the right erector spinae by 63.1%, mean muscle forces in the left erector spinae by 78.5%, and mean muscle forces in the right erector spinae by 120% (Fig. 3). Though the effects were inconsistent across both tool weights, the exoskeletal intervention also increased peak and mean muscle forces under light tool conditions in the left internal oblique ($p < 0.03$) by 42.9% and 66.1% respectively and in the left external oblique ($p < 0.04$) by 32.2% and 41.8% respectively. However, the opposite trend was noted under heavy tool conditions in a 50.2% reduction in peak left rectus abdominis muscle force with the exoskeletal intervention ($p = 0.036$). While subjects could generally perceive the increased muscle force demands in the torso extensors due to the exoskeletal intervention, subjects generally did not perceive the changes in flexor activity observed for the intervention.

In terms of vertical exertion height, peak and mean muscle forces were generally lowest for the 65% stature condition in all muscles in which the main effect of height was determined to be significant (Table 1). For the light tool conditions in which three vertical exertion heights were tested, peak and mean muscle forces in the torso extensors (erector spinae, internal oblique) were highest for the 50% stature condition. Conversely, peak and mean muscle forces in the torso flexors (external oblique, rectus abdominis) under these same experimental conditions were highest for the 100% stature condition with no significant differences observed between the 50% and 65% stature conditions. For the heavy tool, however, muscle forces in the left and right rectus abdominis were increased for the 65% stature condition relative to the 50% stature condition ($p < 0.007$). Upon further examination, this rather unexpected main effect was driven by changes within the control condition rather than changes in both the control and exoskeleton conditions. Without the exoskeletal vest, subjects likely required increased spinal stability to use the heavy tool (but not the light tool) at a higher vertical exertion height, explaining this observed increase in coactivity.

Peak and mean muscle forces on the left side of the body were consistently higher than muscle forces on the right side of the body ($p < 0.001$). Additionally, asymmetric experimental conditions affected mainly the muscles on the left side of the body; peak and mean muscle forces were increased in the left internal oblique, left latissimus dorsi, left external oblique, and left rectus abdominis across both tool weights for asymmetric exertions relative to symmetric ones ($p < 0.001$). Asymmetric conditions also resulted in increased peak and mean muscle forces in the left erector spinae ($p < 0.01$), increased peak and mean muscle forces in the right rectus abdominis ($p < 0.03$), and increased mean muscle force in the right external oblique ($p = 0.013$) for heavy tool conditions. However, the opposite trend was noted for peak and mean right erector spinae muscle forces

Table 1
Statistically significant (*p < 0.05, **p < 0.01, ***p < 0.001) effects that were consistent across both tool weights.

	Intervention	Height	Asymmetry	Intervention * Height	Intervention * Asymmetry
Peak Muscle Force					
Left Erector Spinae	**	*			
Right Erector Spinae	***				*
Left Internal Oblique		*	***	***	
Right Internal Oblique	**			*	
Left Latissimus Dorsi			***		
Right Latissimus Dorsi				**	
Left External Oblique			***		
Right External Oblique				**	*
Left Rectus Abdominis		**	*	***	**
Right Rectus Abdominis		**		***	
Mean Muscle Force					
Left Erector Spinae	***	*		**	**
Right Erector Spinae	***				**
Left Internal Oblique		**	***	***	***
Right Internal Oblique	**			***	
Left Latissimus Dorsi		*	***		*
Right Latissimus Dorsi				*	
Left External Oblique			***		
Right External Oblique				***	*
Left Rectus Abdominis		**	*	***	**
Right Rectus Abdominis		**		***	
Compression					
L3/L4 Superior (Peak)	***	*			
L3/L4 Superior (Mean)	***	*			
A/P Shear					
L5/S1 Superior (Peak)	***	*		***	*
L5/S1 Superior (Mean)	***	*		***	***
Lateral Shear					
L5/S1 Inferior (Peak)		*	***	**	
L5/S1 Inferior (Mean)		*	***	***	

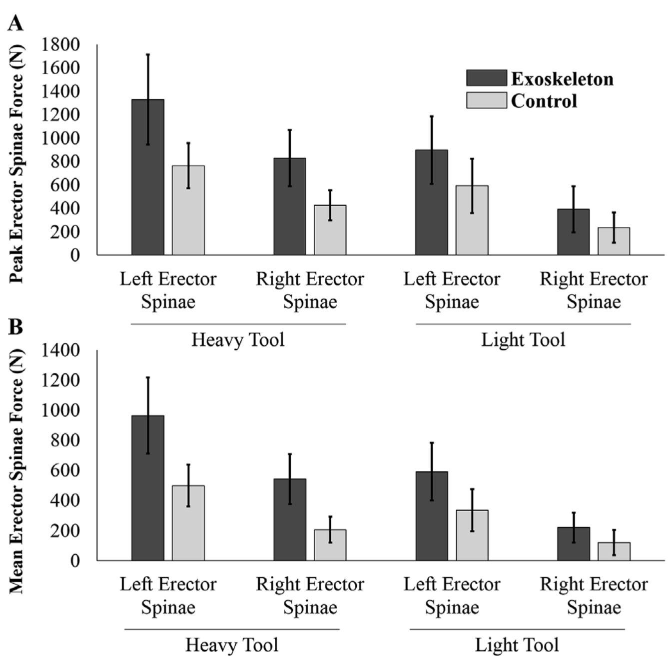

Fig. 3. Left and right extensor muscle forces predicted by the biomechanical spine model including **(A)** peak erector spinae force, **(B)** mean erector spinae force. The back extensors were most affected by the exoskeleton condition due to the added moment from the weight of the vest and arm. Error bars denote standard deviation.

($p < 0.002$) and mean right latissimus dorsi force ($p = 0.0043$) for light tool conditions.

3.2. Spinal load

Compressive spinal loads were of greatest magnitude at the L4/L5 Superior endplate, whereas the greatest A/P and lateral shear loads were observed at the Superior and Inferior endplates at L5/S1, respectively. Peak spinal loads rarely exceeded tissue tolerance limits presented within the literature (Gallagher and Marras, 2012; NIOSH, 1981). However, subjects were most likely to record peak spinal loads exceeding tissue tolerance values using the exoskeletal vest and arm with the heavy tool.

As shown in Fig. 4, peak and mean compressive spinal loads were increased for the exoskeleton condition relative to the control condition ($p < 0.001$). For the light tool conditions, peak and mean compressive spinal loads were increased by 30.9% and 38.5%, respectively, with use of the exoskeleton as compared to the control condition. The effect was more pronounced for the heavy tool, where peak and mean compressive spinal loads increased 52.5% and 56.8%, respectively, across all heavy tool conditions with use of the exoskeleton.

As shown in Fig. 5, use of the exoskeletal intervention was also noted to affect both peak and mean A/P shear loads across both tool weights ($p < 0.001$). Again, the effect was more pronounced for the heavy tool, where peak and mean A/P shear loads were increased by 26.0% and 30.0%, respectively, with use of the exoskeletal intervention. Peak and mean lateral shear loads were influenced by the exoskeleton with the light tool ($p < 0.02$) but not the heavy tool. As inconsistent effects were observed across tool weights and the main focus of this investigation was to determine differences attributable to the exoskeletal intervention, lateral shear spinal loads are not reported and will not be discussed further.

Across all experimental conditions, the 65% stature consistently recorded the lowest compressive and A/P shear spinal loads ($p < 0.004$). Post-hoc analyses showed no significant differences between the 50% and 100% stature conditions with the light tool for either peak and mean spinal compression or peak and mean A/P shear. Additionally, task asymmetry consistently increased peak and mean compressive and A/P shear spinal loads for the heavy tool ($p < 0.001$) but not the light tool. Finally, spinal load measures were subject to

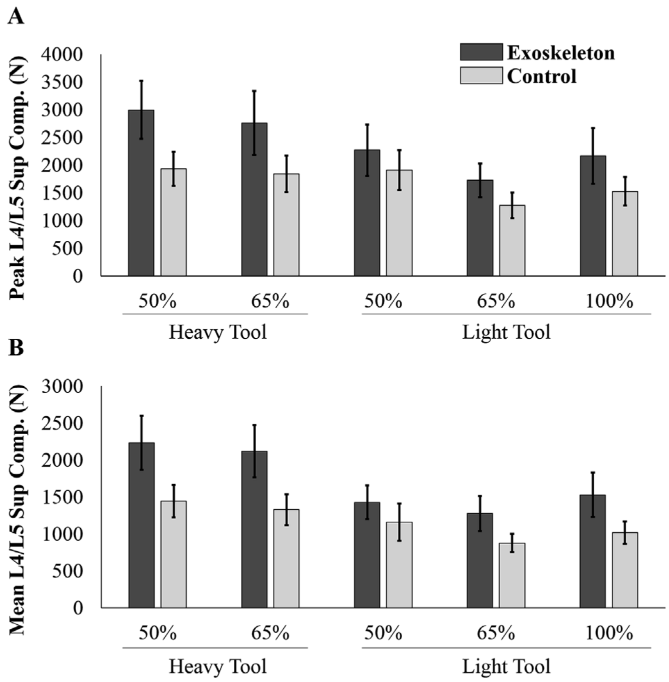

Fig. 4. **(A)** Peak and **(B)** mean L4/L5 Superior compressive spinal loads as a function of intervention level (exoskeleton or control), vertical height, and tool weight. Peak and mean compressive spinal loads were also increased with asymmetry during use of the heavy tool ($p < 0.001$). Error bars denote standard deviation.

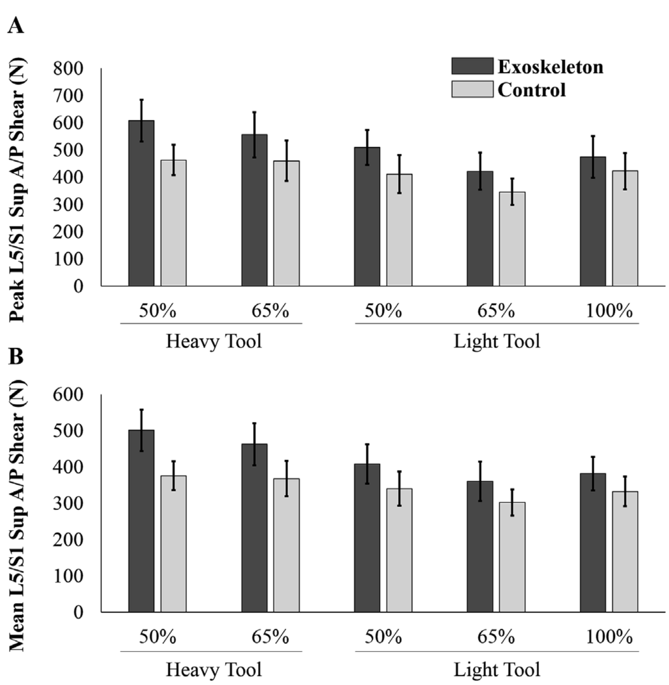

Fig. 5. **(A)** Peak and **(B)** mean L5/S1 Superior A/P shear spinal loads as a function of intervention level (exoskeleton or control), vertical height, and tool weight. As was observed for spinal compression, A/P shear loads were also increased with asymmetry ($p < 0.008$), but to a lesser extent than the independent variables included here. Error bars denote standard deviation.

several significant interaction effects worth noting. Peak and mean A/P shear loads were subject to an intervention*height interaction for both tool weights in which the effect of exertion height (namely, increased loads at 50% or 100% stature) were more pronounced with use of the exoskeleton than under control conditions ($p < 0.001$). A similar trend was noted for both tool weights (despite no main effect being observed for the light tool) in a significant intervention * asymmetry effect noted for peak and mean A/P shear; the exoskeletal intervention amplified the effects of asymmetry (increased spinal load under asymmetrical conditions) when compared to control conditions ($p < 0.04$). With the light tool but not the heavy tool, a significant intervention*height interaction was also noted for peak and mean compression, and a significant intervention *asymmetry interaction was noted for mean compression ($p = 0.022$).

4. Discussion

Though numerous exoskeletons are already commercially available and have been introduced into industry, this investigation remains one of the first to quantify biomechanical loads to joints for which the exoskeletons were not specifically designed to support. In particular, this investigation aimed to quantify muscle forces and spinal loads that result from performing a simulated work task using a hand tool with and without an exoskeletal vest and arm.

Most studies examining biomechanical risk to the low back employ load-tolerance logic, comparing predicted peak spinal loads to tissue tolerance values previously reported within the literature (Gallagher and Marras, 2012; NIOSH, 1981). It is important to note that peak spinal loads rarely exceeded these reported tissue tolerances in the present investigation. This is not to say, however, that the spinal loads predicted for the exertions tested pose no biomechanical risk. It is suspected that the effects of both fatigue resulting from constant torso muscle activation and higher cumulative loading that accompanied this exoskeleton's use could be costly to the lumbar spine.

The use of the exoskeletal vest and arm intervention significantly increased peak and mean muscle tensions in the back extensors (erector spinae, internal oblique) compared to control conditions in which the vest was not used. The increased muscle forces likely resulted from the need to counteract an increased external moment acting on the lumbar spine from the weight of the vest and tool considering its relatively large moment arm. The increased muscle loads in the back extensors that resulted from use of the exoskeletal intervention also resulted in up to 56.8% higher mean (and 52.5% higher peak) compressive spinal loads. It is important to note that the spinal loads observed here represent a 'best-case' scenario in terms of biomechanical load; after all, the exoskeleton was worn by subjects in this study for a relatively short period of time (approximately 1 h). Consistently high levels of torso muscle activation (as was observed with exoskeleton use during this study) would be expected to negatively influence muscle fatigue and alter muscle coactivation patterns over the course of a work shift, potentially leading to increased peak and mean loads placed onto the spine that surpass the tolerance of the tissue (Marras et al., 2006).

The exoskeleton intervention also caused significant right/left differences to be observed. As the hand tool was attached to a mechanical arm situated on the exoskeletal vest on the right side of the body, muscles on the left hand side of the body had to activate at higher levels to compensate for the imbalanced moment caused by the weight of the arm and the attached tool. This need for more muscle activation on the left hand side of the body was further exacerbated with the introduction of task asymmetry. The imbalanced muscle activations observed are expected to further promote muscle fatigue.

The exoskeletal intervention tested was designed to shift the load from powered hand tools in 'anatomically-appropriate ways,' delivering the majority of the weight of the tool to the pelvis and shoulder blade regions of the body. It is clear, however, that significant loads were also shifted onto the low back, as evidenced by the high muscle forces observed in the torso extensor muscles and the increase in compressive spinal loads (up to 56%) that resulted. In the opinion of the authors, a properly designed exoskeletal intervention should be capable of reducing loads to a joint of interest (in this case the shoulder) without a direct cost to another region of the body (the low back). After all, exoskeletons should work in concert with the worker. It is likely, too, that exoskeletal interventions may help under some specific conditions but provide little to no benefit to users under other conditions.

The manufacturer of the exoskeletal vest acknowledges that muscle discomfort is to be expected for novice users of the intervention, particularly in the torso extensors. However, the manufacturer also reassures users that the carrying capacity of the torso extensors will increase with increased use of the device and that discomfort will subside as a result (Fawcett "Steadicam Posture"). While it is true that human tissue becomes stronger and adapts to increased load demands (Wolff's law), this adaption process is also heavily dependent on adequate rest and recovery time in addition to the delivery of nutrients for repair. In the case of occupational work, significant rest may occur only between work shifts and on the weekends and may still be interrupted by housework or recreational activity. Inadequate cumulative rest time could decrease the tissue tolerance of spinal structures and could instead lead to a decline in low back function rather than strengthening (Marras et al., 2014).

In terms of the other workplace variables investigated, the data quantitatively demonstrated the benefit of 65% stature height condition as compared to 50% and 100% stature conditions, as this vertical height saw the lowest spinal loads in compression and A/P shear. This result is consistent with previous work investigating loading for pushing utilizing the same biomechanical model, in which pushing at a medium vertical height (about 65% stature) was found to generate the lowest A/P shear force (Knapik and Marras, 2009). The 65% stature condition required workers to neither reach above their heads (causing an excessive moment about the spine due to the upper extremity segment weight and tool weight) or bend at the torso (causing an excessive moment about the spine due to the weight of the torso). Thus, it is recommended for workers in industry to perform similar manual work at about the 65% of stature level in order to reduce spinal loads and prevent MSDs.

As is seen for any investigation, several limitations were associated with this work. First, measures were predicted under laboratory conditions. Subjects recruited for this investigation were young (25.3 ± 6.0 years) and inexperienced in manual materials handling; thus, the population employed was not perfectly representative of the working population. Additionally, subjects were not trained on the exoskeletal device tested for a significant period of time, and task performance was not measured within this study. No trials were collected before subjects had time to adjust to the intervention, but it remains unclear whether subjects would become significantly more/less comfortable or would perform significantly more/less consistently with the device with a longer duration of use. It also remains unclear how biomechanical loading would differ with repeated cycles. In this investigation, just one cycle of the experimental task constituted each trial before subjects returned to a neutral posture, but it is likely that in industrial applications like shipbuilding or airplane manufacturing that many cycles of the task and longer durations of overhead work would be more prevalent. Furthermore, though all of the subjects were given instruction on how to perform the task at hand before any data were collected, posture was not precisely controlled. However, varied postures assumed by subjects are expected to be noted in a real occupational environment.

Though the goal of this investigation was to quantify muscle forces and spinal loading specifically in the lumbar spine, quantification of shoulder joint loads should be considered in future studies in order to precisely quantify the potential trade-offs that exist between the shoulders and low back. Moreover, future studies should also seek to investigate exoskeletal interventions that direct loads directly to the ground or use a counterbalance behind the body to offset the large moment created by the power tool. It is conceivable that these design changes could make for an exoskeletal intervention that appropriately offloads the shoulders without a cost to the lumbar spine. Regardless, this study served as an appropriate baseline assessment of an exoskeletal intervention supporting the weight of a powered hand tool, as it relies on a rather simple load path directing loads to the shoulders, hips, and the low back.

Finally, subjective discomfort measures related to torso and lower back fatigue, shoulder and upper limb fatigue, and general impressions of the exoskeleton intervention as compared to control conditions were not formally recorded during this investigation. The authors do note, however, that subjects often expressed that they found the exoskeletal vest to be uncomfortable and mentioned that they would prefer not to use the intervention if they had the choice, at least for the duration of this study and the specific experimental conditions within it.

5. Conclusion

While exoskeletal interventions offer the potential to mitigate biomechanical risk to the shoulders resulting from the use of heavy hand tools in occupational environments, the results of this study suggest that use of this particular exoskeleton may come at some cost to the low back. It is important to note that the results presented within this investigation are specific to the exoskeletal intervention and specific conditions tested. As such, these results should not be extrapolated to all exoskeleton devices. However, the results of the present investigation do highlight the need for exoskeletons designed for use in occupational environments to be developed using a systems approach, considering biomechanical loading not only at a particular musculoskeletal joint of interest (i.e., the shoulders) but also how mechanical loads might be shifted or transferred with exoskeleton use.

Competing interest/conflict of interest

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

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