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RESEARCH ARTICLE

The effect of a soft active back support exosuit on trunk motion and thoracolumbar spine loading during squat and stoop lifts

Jacob J. Banks[a,b], David A. Quirk[c], Jinwon Chung[c], Jason M. Cherin[c], Conor J. Walsh[c] and Dennis E. Anderson[a,b]

[a]Center for Advanced Orthopaedic Studies, Beth Israel Deaconess Medical Center, Boston, MA, USA; [b]Department of Orthopedic Surgery, Harvard Medical School, Boston, MA, USA; [c]John A. Paulson School of Engineering and Applied Sciences, Harvard University, Boston, MA, USA

ABSTRACT
Back support exosuits aim to reduce tissue demands and thereby risk of injury and pain. However, biomechanical analyses of soft active exosuit designs have been limited. The objective of this study was to evaluate the effect of a soft active back support exosuit on trunk motion and thoracolumbar spine loading in participants performing stoop and squat lifts of 6 and 10 kg crates, using participant-specific musculoskeletal models. The exosuit did not change overall trunk motion but affected lumbo-pelvic motion slightly, and reduced peak compressive and shear vertebral loads at some levels, although shear increased slightly at others. This study indicates that soft active exosuits have limited kinematic effects during lifting, and can reduce spinal loading depending on the vertebral level. These results support the hypothesis that a soft exosuit can assist without limiting trunk movement or negatively impacting skeletal loading and have implications for future design and ergonomic intervention efforts.

PRACTITIONER SUMMARY
Back support exosuits have the potential to reduce musculoskeletal workplace injuries. We examined and modelled the impact of a soft active exosuit on spine motion and loading. The exosuit generally reduced vertebral loading and did not inhibit trunk motion. Results of this study support future research to examine the exosuit as an ergonomic intervention.

Abbreviations: ANOVA: Analysis of Variance; API: Application Programming Interface; BMI: Body Mass Index; CT: Computed Tomography; IMU: Inertial Measurement Unit; L# - Lumbar Vertebral Level (#); RoM: Range of Motion; T# - Thoracic Vertebral Level (#)

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KEYWORDS
Musculoskeletal modelling; lifting; exosuit; lift technique

1. Introduction

Back pain is the leading cause of global disability (Wu et al. 2020). While back pain is often diagnosed as idiopathic or non-specific (Deyo and Weinstein 2001; Hoy et al. 2010), work related back pain is commonly associated with *in vivo* tissue (e.g. muscle, disc, vertebral) loading from repeated or prolonged non-neutral trunk postures and high external demands (Halonen et al. 2019; Marras 2000). Work related back pain claims result in over $12 billion in annual treatment costs in the United States alone (Druss et al. 2002). Manual material handling workers (e.g. those who perform lifting and lowering tasks) are especially susceptible to back tissue injury and pain (Bureau of Labour Statistics 2016). As such, engineers and ergonomists have attempted to abate manual materials handling related back injuries via a variety of interventions with the goal of reducing internal tissue loads in the back (Gallagher and Heberger 2013; Roman-Liu, Kamińska, and Tokarski 2020).

The motivation for any lift is to move an object from one location to another. Influencing the position of an object requires physical work by an individual(s) or device. In ideal circumstances a mechanical lift device (e.g. hoists, ceiling lifts) can perform the work to dramatically reduce the risk of a back injury. Unfortunately, mechanical lifts are not always available due to financial, logistical, or spatial constraints (e.g. Dang et al. 2022). Therefore, alternative solutions are

CONTACT Dennis E. Anderson danders7@bidmc.harvard.edu Center for Advanced Orthopaedic Studies, Beth Israel Deaconess Medical Center, 330 Brookline Avenue, RN115, Boston, MA 02215 USA

often pursued. Ergonomists will often recommend redesigning the task to better fit the worker (Ayoub 1982; Marras 2000), but these too have financial and spatial constraints. Training programs aim to educate workers on best practices and optimal lifting techniques. Despite the low cost and marginal spatial constraints of training programs, they have resulted in limited efficacy in reducing back injuries (Alberto et al. 2018; Marras 2000; Martimo et al. 2008). Lumbar supports, or back belts, offer the potential of a mobile intervention. Back belts are designed to reduce lift related back tissue demands on workers by encouraging proper kinematics, increasing intra-abdominal pressure, and stabilising the spine (McGill 1993). However, controversy exists concerning their biomechanical effectiveness, physiological trade-offs, and whether back belts can ultimately prevent injury (McGill 1993; van Duijvenbode et al. 2008). Alternatively, recent technological advances have promoted a promising hybrid lift-assist approach, where a lift assist exoskeleton is comfortably worn by a worker.

Back exoskeletons, or exosuits, are devices designed to aid the trunk musculature of the user during lifting and lowering tasks. Exosuits vary in their overall design (e.g. soft vs. rigid) and force delivery mechanism (e.g. active vs. passive) (Ali et al. 2021; de Looze et al. 2016; Kermavnar et al. 2021; Theurel and Desbrosses 2019). Several research studies have demonstrated how back exosuits can reduce back extensor muscle activity (6–48%) and net internal joint moments (up to ~30%) during lifting tasks (for recent reviews see Ali et al. 2021; Kermavnar et al. 2021). Musculoskeletal modelling has further estimated the resulting *in vivo* tissue loading on lumbar joints may be reduced by 5–27% (Abdoli-Eramaki et al. 2007; Kermavnar et al. 2021; Madinei and Nussbaum 2023; Schmalz et al. 2021; Ulrey and Fathallah 2013). This evidence suggests that by reducing the *in vivo* tissue demands on the user, exosuit devices can potentially reduce risk of injury and pain.

The most dramatic biomechanical effects from back exosuits tend to be associated with rigid exosuits that deliver pure assistive extension moments (Schwartz, Theurel, and Desbrosses 2021). Despite their promise, rigid devices still face several challenges to be accepted by the workforce; due in part to their solid, stiff, and bulky nature (Ali et al. 2021; Siedl, Wolf, and Mara 2021). Less is known about the biomechanical impact of soft exosuits. Soft exosuits typically deliver tensile forces via a strap or cable that runs parallel to the user's spine, which can potentially contribute to spinal compressive forces (Abdoli-Eramaki et al. 2007; Lamers, Yang, and Zelik 2018). It is generally hypothesised that soft exosuits have net positive effect on reducing spinal loads, given the larger mechanical advantage of the cable relative to the extensor muscles of the trunk, but prior biomechanical analyses have been limited by assuming a fixed moment arm distance of the cable, and concentrating on a single lumbar level (e.g. L5/S1 intervertebral joint). Given the complex curvature of the spine, soft exosuits may have a varying impact across the levels of the spine. Therefore, for a holistic understanding of the *in vivo* impact of a soft active back exosuit during dynamic lifting tasks, a comprehensive analysis of loading across the entire thoracolumbar spine is warranted.

The aim of this study is to evaluate the biomechanical impact of a soft active exosuit (Chung 2023; Quirk et al. 2023a; 2023b). We hypothesised that, as with previous musculoskeletal evaluations of back exosuits (Ali et al. 2021; Kermavnar et al. 2021), there will be a reduction in vertebral joint forces and a limited impact on trunk kinematics during lifting. Moreover, for the first time, our analysis will use a full body thoracolumbar musculoskeletal model (Akhavanfar et al. 2023; Alemi et al. 2023; Bruno et al. 2017, Bruno, Bouxsein, and Anderson 2015; Burkhart et al. 2020) to investigate the impact of an exosuit on compressive and shear loading of the spine at all thoracolumbar vertebral levels. As lifting style and weight lifted both affect spine loading (Chaffin and Park 1973; van Dieën, Hoozemans, and Toussaint 1999; von Arx et al. 2021; Zander et al. 2015) we will also evaluate the effect of the exosuit in both stoop and squat lifting styles with different weights lifted. Thus, this study will provide a comprehensive biomechanical analysis of a soft active back exosuit, and therefore better inform ergonomists on the safety and viability of this type of exosuit design intervention for reducing back injuries and pain related to manual materials handling.

2. Materials and methods

2.1. Participant recruitment

Participants were recruited from the local population. To be eligible, participants were interviewed concerning their ability to comfortably lift and lower an object, history of musculoskeletal (including back pain in the last 6 months) or neurological disorders, age (between 18 and 55 years of age), obesity (BMI ≤ 30), and recent COVID-19 related symptoms. A priori, a sample size calculation determined that fourteen participants would be sufficient to detect exosuit differences within participants using an 80% power and $\alpha = .05$ (Quirk et al. 2023b). All participants provided written informed consent to participate in this study, which was approved by the Institutional Review Board of Harvard Medical School (IRB18-0960).

2.2. Soft active back exosuit for assisting lifting

We evaluated a soft active back exosuit that is designed to assist back and thigh extensors throughout a lift (for additional details on the exosuit design, control mechanism, and assist profile see: Chung 2023; Quirk et al. 2023a, 2023b). In brief, the self-contained and mobile exosuit weighed approximately 2.7 kg and consisted of a battery and electromechanical actuator pack located on the user's back attached via backpack-like shoulder straps (Figure 1). The actuator pulled a nylon cord which branched off at a buckle to connect to each upper thigh. The assistive force magnitude (ranging from 10 to 250 N) of the electromechanical actuator on the cord was generated by a custom controller as a function of trunk and thigh kinematics from embedded IMUs. The combined impact of the assistive force and the moment arm of the cord is designed to deliver back extension assistance of up to 30 Nm at lift initiation (Quirk et al. 2023a). The controller commands an assistive force based on the level of trunk flexion and scaled based on the movement direction (Chung 2023; Quirk et al. 2023a, 2023b). The approach was designed to provide more assistance during trunk extension to maximally offload back extensors and less assistance during trunk flexion, as to not impede the movement of the user. The assistive force on the cord was measured with an inline load cell (LSB200, FUTEK Advanced Sensor Technology Inc., CA, USA), for testing and modelling purposes.

2.3. Experimental procedure

Participants attended a single laboratory session. All participants donned form-fitting compression clothing to facilitate motion capture marker and inertial measurement unit (IMU; MTi-3 AHRS, Xsens Technologies B.V., Enschede, the Netherlands) placement. Following height and body weight measurements, participants were provided with verbal and visual instructions along with the opportunity to practice squat and stoop lifts with and without the exosuit. The eight different lift scenarios included all combinations of two different crate masses (6 or 10 kg), two lift types (stoop or squat), and two exosuit conditions (exosuit or no exosuit). These lift scenarios were assigned to investigate how the exosuit performed during two unique lift styles and workloads. Prescribing the user to specific lifting styles allowed us to investigate the impact of the exosuit with fewer concerns for kinematic adaptations which can independently impact spine loading (Khoddam-Khorasani, Arjmand, and Shirazi-Adl 2020). All practice lifts were performed until participants visually appeared to be, and verbally confirmed, their comfort with the eight lifting scenarios. Practice sessions

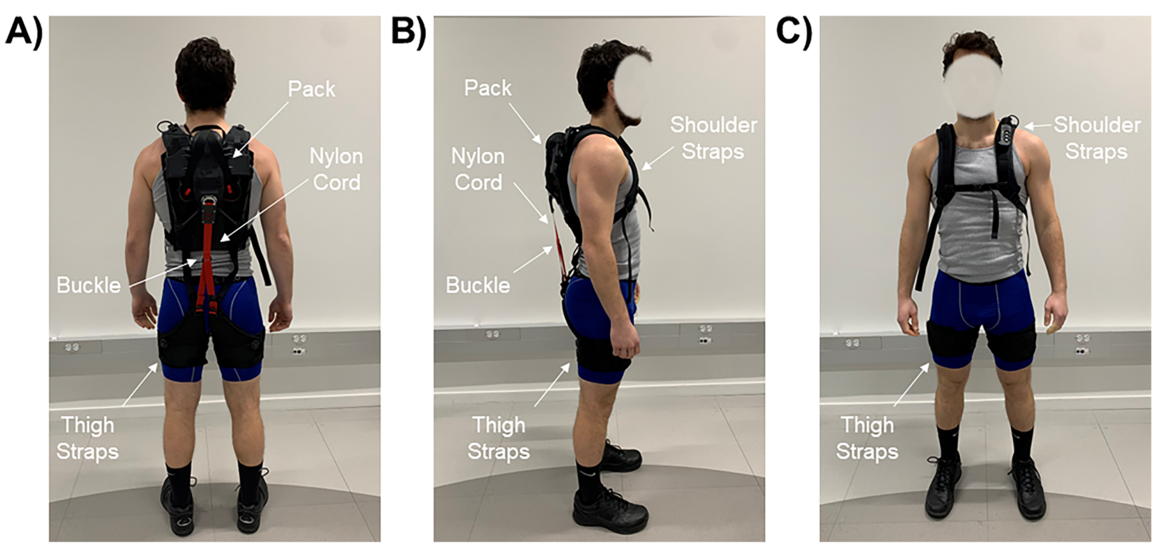

Figure 1. Soft active back exosuit for assisting lifting from the A) posterior, B) side, C) and front. Key features indicated with white labels and arrows and described in 2.2.

typically lasted between 5 and 10 minutes. Four IMUs were then placed over the T1, T8, L3, and sacral levels of the spine, and 49 individual and clustered motion capture passive-reflective markers were placed in relation to palpated anatomical landmarks and body segments.

The experimental testing session commenced with a static calibration pose in a standardised static standing T-pose, followed by eight different lift scenarios and an exosuit tracking calibration trial. All trials consisted of lifting a weighted crate in the sagittal plane, starting with the crate on a 10 cm block, lifting it to waist height, and returning it to the starting location. Lift scenarios were performed for ten repetitions evenly spaced over two-minutes, using an audio cadence from a metronome. Stoop-style lifts utilised a 43 × 28 × 32 cm (width × depth × height) crate and were described to participants as '*keeping your legs as straight as possible while lifting/lowering*'. Squat-style lifts utilised a shorter 46 × 31 × 18 cm crate and were described to the participants as, '*bending your knees and not rounding your spine while lifting/lowering*'. The shorter crate height for squat lifts was assigned to make trunk flexion more comparable across lifting conditions. Presentation order was implemented using a counter-balanced Latin-square randomised strategy that was blocked by exosuit condition to reduce the probability that markers and IMUs would be dislodged when changing between exosuit conditions and to limit any potential learning and fatigue effects. Between conditions, participants were provided a 1–2 minute break, with additional standing rest time as researchers helped don or doff the exosuit. The exosuit condition block was uniquely accompanied by an exosuit tracking calibration trial, where participants performed two unweighted 'lift' repetitions of both the squat and stoop lift to track and then model the location of the exosuit cord for each lift type. In the exosuit tracking calibration trial, additional markers were attached to the exosuit pack, cord junction (i.e. the buckle where the cord branched off to the thighs), and on the thigh straps.

Throughout the testing marker positions were recorded (200 Hz) with a sixteen-camera motion analysis system (Qualisys AB, Sweden). The four spine IMUs and the exosuit assistive force were sampled (200 Hz) using an 8-bit embedded microcontroller (PIC18F25K80, Microchip Technology, Inc., AZ, USA). The exosuit signals were recorded via custom MATLAB scripts (The MathWorks Inc., USA) writing to an onboard flash memory card (SDSQUNC-032G-AN6IA, Scandisk, CA, USA) that was synchronised to a common pulse from the motion analysis system.

2.4. Musculoskeletal model

The thoracolumbar models used in this study were created and solved in OpenSim 4.2 (Delp et al. 2007). The base full-body models consisted of 620 musculotendon actuators, 78 rigid-bodies, and 165 independent degrees of freedom (Bruno et al. 2017, Bruno, Bouxsein, and Anderson 2015; Burkhart et al. 2020). Participant-specific models were created from the appropriate male or female base model according to static calibration pose marker positions and measured body mass. Trunk musculotendon actuator sizes and moment arms were adjusted using regression-based estimates of muscle size developed from a large dataset of muscle measurements from CT scans (Johannesdottir et al. 2018). Standing sagittal spine curvature was adjusted to match estimated internal spine curvature from the spine mounted IMUs during the static calibration pose. Specifically, the sagittal-plane angles of the four IMU sensors at T1, T8, L3, and sacral levels were used to fit a quadratic polynomial estimate of skin-surface angle across the length of the spine. These polynomials were used to estimate skin-surface Cobb angles, which were then used as inputs for published regression equations to estimate internal spine Cobb angles (Furlanetto et al. 2017; Grindle et al. 2020). Spine curvatures of participant-specific models were adjusted to match the internal spine Cobb angles. All analyses were performed using participant-specific models. The crate inertia was modelled by attaching a rigid body of half the mass and estimated inertial moment properties of the actual lifted crate to each hand (Akhavanfar et al. 2022). The exosuit was modelled by attaching representative rigid-bodies to the dorsal trunk and thighs joined by path actuators representing the assistive cord and dorsal contact points along select ribs (Figure 2).

Kinematics from each lifting trial were calculated by fitting recorded motion capture passive-reflective marker positions to the corresponding markers on the participant-specific models (Figure 2) using the OpenSim Inverse Kinematics Tool. IMU orientations were not considered for the lifting trial kinematics. Coordinate coupling constraints were applied to the model during inverse kinematic calculations to reduce the fifty-one degrees of freedom associated with the thoracolumbar spine to a determinate three (Alemi et al. 2021; Banks, Umberger, and Caldwell 2022). During the exosuit conditions, the exosuit actuator path locations were calculated based on regressions that were specific to each lift type during each participant's exosuit tracking calibration trial. All resulting kinematics were filtered within OpenSim by applying a

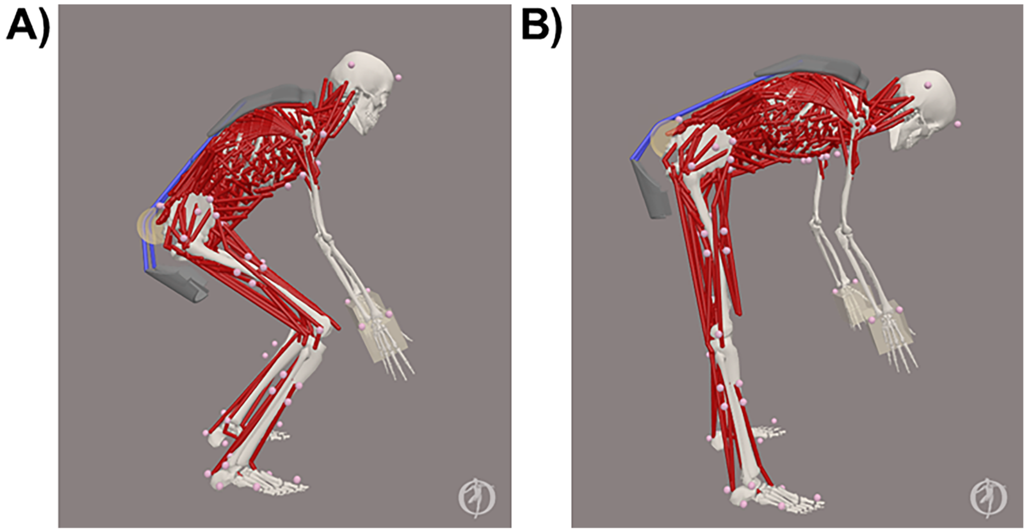

Figure 2. Full-body thoracolumbar OpenSim musculoskeletal model posed in a A) squat and B) stoop type lift. Dark grey spheres represent the location of the 49 reflective markers used for motion tracking, red cords the 620 musculotendon actuators, and blue cord(s) the nylon extension assist cord linking the exosuit pack and thigh straps.

zero-lag low-pass (4 Hz) fourth-order Butterworth filter selected from residual curves (Winter 2009).

Trial kinetics were estimated using the OpenSim Static Optimisation and Joint Reaction Analysis Tools. To properly include and exclude the crate, the modelled crate inertia properties were only considered following pick-up. During static optimisation the recorded exosuit assistive force (when applicable) was assigned as the actuator force while the remaining unknown actuator and musculotendon forces were optimised by minimising the sum of their cubed activations (Crowninshield and Brand 1981). All modelling steps were controlled via custom API scripts in MATLAB (Lee and Umberger 2016).

2.5. Analysis

Five of the ten lifting repetitions were analysed from each lift scenario. To ensure the participant had adapted to the scenario and the recording session was not prematurely cut-off during the final tenth lift, the five lifts typically comprised of repetitions four through nine. However, exceptions were made when a marker(s) was obscured, or the participant moved unnaturally (i.e. scratched their head or mishandled the crate). Each lift repetition was time-normalized from the frame immediately prior to the participant bending forward with the crate still located on the block until when the participant was standing upright holding the crate. Participant kinematics and kinetic variables were extracted as an ensemble average of the five repetitions.

Primary outcome variables of interest included: (1) sagittal plane angular ranges of motion (RoM) measurements for the spine, pelvis, and overall trunk (i.e. spine plus pelvis angles), and (2) peak vertebral compressive and resultant shear forces (in Newtons) on each of the seventeen thoracic and lumbar vertebrae during the lift. Angular RoM was calculated as the absolute difference between the maximum and minimum angle observed during each lift. Vertebral body compressive and resultant shear forces were calculated as the average of the adjacent inferior and superior intervertebral joint forces, as derived from the OpenSim Joint Reaction Tool.

Three separate repeated-measures four-way analysis of variance (ANOVA) tests were conducted to evaluate the effects of exosuit condition (with or without), vertebral level or RoM location (T1 to L5; Pelvis/Spine/Trunk), lift type (squat or stoop), crate mass (6 or 10 kg), and their interactions with RoM, vertebral compression force, and vertebral resultant shear force as dependent variables. Significant ANOVA effects were interpreted using a Bonferroni correction considering the three analyses ($\alpha < 0.017$ or .050/3). Post hoc analyses of significant exosuit effects were performed using Tukey's Honest Significant Difference pairwise comparisons, grouping by variables displaying significant interactions with exosuit as needed and focused on comparing differences between exosuit conditions. All statistical tests were performed in custom scripts utilising the statistical toolbox in MATLAB.

3. Results

3.1. Participant demographics

Fourteen ($n = 14$) participants (male 10; female 4) consented and successfully completed the entire experimental protocol. The mean ± standard deviation participant height was 1.75 ± 0.09 m in stature (male 1.79 ± 0.07 m; female 1.66 ± 0.02 m), body mass was 75.7 ± 13.4 kg (male 81.3 ± 10.9 kg; female 61.9 ± 8.2 kg), and age was 31 ± 4 years (male 32 ± 2 years; female 31 ± 6 years). With counter-balanced randomisation seven participants started in the exosuit block and seven in the no exosuit block.

3.2. Exosuit impact on pelvis, spine, and trunk kinematics

Primary ANOVA results for kinematics (Table 1) indicated main effects of lifting style and kinematic location, but not for exosuit and mass lifted. Specifically, pelvic tilt (44.9 ± 16.9 degrees) and spine flexion/extension (36.9 ± 13.4 degrees) were similar, but both had less ($p < .001$) RoM than the trunk (78.6 ± 13.0 degrees). Stoop lifting (59.1 ± 24.9 degrees) had more ($p < .001$) RoM than squat lifting (47.8 ± 19.8 degrees). There were significant two- and three-way interactions between exosuit condition and location ($p < .001$), and exosuit condition, location, and lift type ($p = .005$), respectively. Post hoc analyses of exosuit effects by location and lift type (Figure 3) indicated that during squat lifts pelvic tilt was lower in the exosuit (35.8 ± 16.3 degrees) than no exosuit (39.0 ± 14.3 degrees) condition, but spine flexion/extension was higher (38.7 ± 12.3 and 34.7 ± 11.7 degrees for exosuit and no exosuit conditions, respectively). These effects seemed to offset and result in no impact on trunk RoM (69.7 ± 10.2 and 69.0 ± 10.2 degrees for exosuit and no exosuit conditions, respectively). Similar effects were observed

Table 1. Significance summary of kinematic (range of motion; RoM) and kinetic (peak vertebral compression and resultant shear) dependent variables. *P*-values are presented for all main effects and exosuit conditions (E) interactions with vertebral or RoM location (L), lift type (T), and crate mass (M). Significant effects (α < .017) are *italicised.*

	RoM	Compression	Resultant Shear
E	.222	*<.001*	*.001*
L	*<.001*	*<.001*	*<.001*
T	*<.001*	*<.001*	*.001*
M	.494	*<.001*	*<.001*
E×L	*<.001*	*<.001*	*<.001*
E×T	.110	*.006*	.286
E×M	.274	.137	.027
E×L × T	*.005*	*<.001*	*<.001*
E×L × M	.310	.092	.041
E×T × M	.128	.175	.943

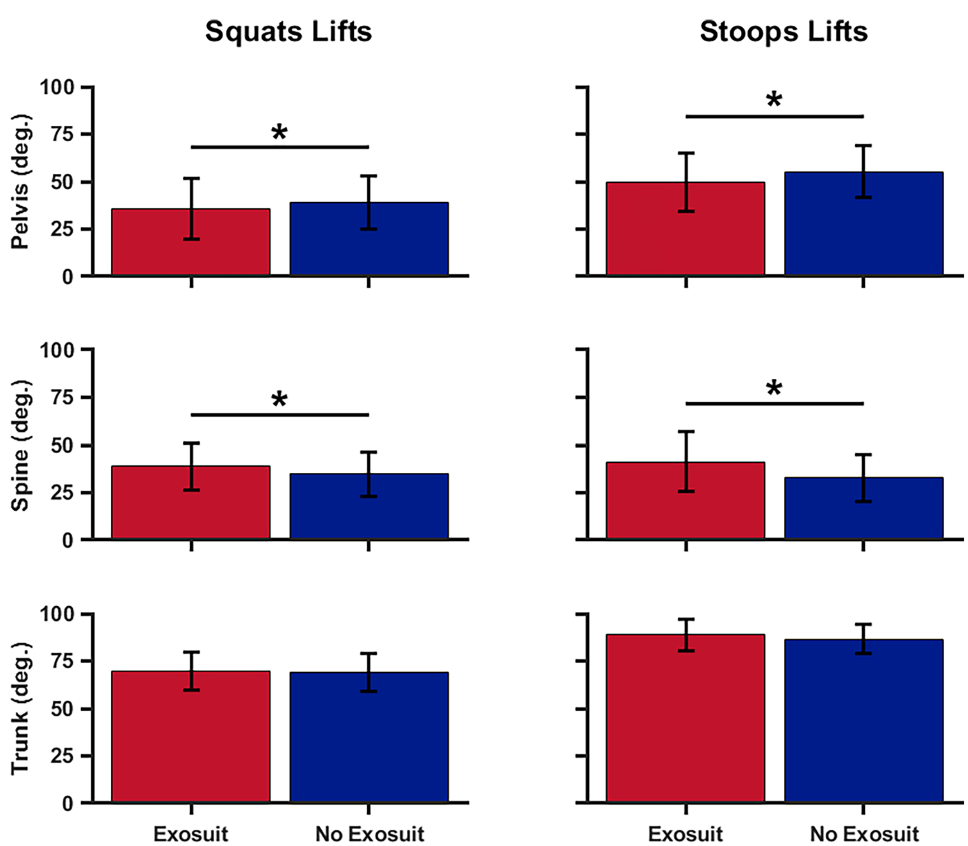

Figure 3. Pelvis, spine, and trunk range of motions (in degrees) during the two lifting scenarios from the exosuit (*red*) and no exosuit (*blue*) conditions. Asterisks indicate significant differences from post hoc analyses between exosuit conditions at equivalent joint levels. Data whiskers represent the standard deviation about the mean.

during the stoop lifts, where changes in pelvis tilt (49.6 ± 15.6 and 55.3 ± 14.2 degrees for exosuit and no exosuit conditions, respectively) and spine flexion/extension (41.3 ± 15.7 and 32.8 ± 12.6 degrees for exosuit and no exosuit conditions, respectively) during the exosuit condition resulted in no impact on trunk RoM (88.9 ± 8.4 and 86.6 ± 7.7 degrees for exosuit and no exosuit conditions, respectively). There were no other significant exosuit interactions (Table 1).

3.3. Exosuit impact on peak thoracolumbar compressive forces

Primary ANOVA results for compressive loading (Table 1) showed main effects of exosuit, lifting style, vertebral level, and mass lifted. Specifically, the exosuit reduced vertebral compression forces by about 5% relative to the no exosuit conditions (1556 ± 906 vs. 1487 ± 887 N, $p < .001$) averaged across all vertebral levels and lifts. Stoop lifting (288 ± 137 N) led to lower compressive forces than squat lifting (301 ± 144 N; $p < .001$). Compressive loading varied widely among vertebral levels ($p < .001$), from a low of 231 ± 70 N at T1 to a high of 2815 ± 450 N at L5. Box mass increased compressive forces (from 272 ± 132 N with the 6 kg crate to 316 ± 145 N with the 10 kg crate, $p < .001$), however no mass with exosuit interactions occurred (Table 1). Two- and three-way interactions were observed for exosuit condition and vertebral level ($p < .001$), exosuit condition and lift type ($p = .006$), and exosuit condition, vertebral level, and lift type ($p < .001$). Post hoc analyses of exosuit effects for each vertebral level across both lift types indicated that during the squat lifts the T7 through L1 vertebral peak compression force was reduced by 3 to 7% (Table 2; Figure 4). During the stoop lifts T6 through L2 vertebral peak compression was reduced by 5 to 14%. At all other vertebral levels peak compression loads were not significantly changed by wearing the exosuit during either the stoop or squat lifts. There were no other exosuit condition two-, three-, or four-way interactions of significance for peak compression loading (Table 1).

3.4. Exosuit impact on peak thoracolumbar resultant shear forces

Primary ANOVA results for shear loading (Table 1) showed main effects of exosuit, lifting style, vertebral level, and mass lifted. Specifically, the exosuit reduced peak resultant shear loading by over 5% relative to the no exosuit conditions (302 ± 144 vs. 285 ± 136 N, $p < .001$) averaged across all vertebral levels and lifts. Stoop lifting (288 ± 137 N) led to lower shear forces than squat lifting (301 ± 144 N; $p < .001$). Shear loading varied widely among vertebral levels ($p < .001$), from a low of 133 ± 29 N at L2 to a high of 648 ± 204 N at L5. Crate mass had a significant ($p < .001$) effect on peak resultant shear forces (316 ± 145 N and 272 ± 132 N for 10 and 6 kg crate, respectively). There were two- and three-way interactions of exosuit condition with vertebral level ($p < .001$) and vertebral level and lift type ($p < .001$), respectively. From post hoc analyses, during the squat lifts the exosuit significantly reduced peak resultant shear forces by 3 to 26% in eight of the seventeen vertebrae (Table 2; Figure 4). However, exosuit also increased peak resultant shear forces by 6 to 19% in five vertebral levels. A similar phenomenon was observed during stoop lifts, where peak shear forces were decreased by 5 to 29% in eight levels and increased by 6 to 22% in six levels with the exosuit. These effects were not uniform at all vertebral body levels and were dependent on exosuit condition (Table 2). There were no other exosuit condition two-, three-, or four-way interactions of significance for peak resultant shear forces (Table 1).

4. Discussion

This study evaluated the impact of using a soft active back support exosuit on the kinematics and spinal loading during four different sagittal-plane lifting scenarios. Our results demonstrated that while the exosuit may slightly alter spine and pelvis kinematics, the exosuit had no impact on the overall trunk RoM. This partially supported the initial hypothesis, that soft exosuit design would not alter the user's kinematics. Kinetic results indicated that both peak compressive and resultant shear loading were reduced overall while wearing the exosuit. The reduction of compressive and shear loading was largely concentrated on the lower thoracic and upper lumbar vertebrae; however, peak shear loading did increase in a few vertebrae. Once again, the hypotheses were partially supported, as not all vertebral levels benefitted from the exosuit. Regardless, these results indicate that generally, the exosuit does not alter lifting kinematics and does not pose a risk of increasing internal vertebral compressive loading.

The finding that overall trunk motion was unchanged is an important outcome, likely related to the soft active exosuit design examined here. Maintaining trunk kinematics is important, as they can independently modify compressive forces (Khoddam-Khorasani, Arjmand, and Shirazi-Adl 2020). By using a

Table 2. Peak vertebral compressive and resultant shear forces (N) of the seventeen thoracolumbar vertebrae across the exosuit conditions and two lift types. Significant differences from post hoc analyses between exosuit conditions at the same vertebral level are indicated by greater and less than signs.

	Compression (N)						Resultant Shear (N)					
	Squat Lifts			Stoop Lifts			Squat Lifts			Stoop Lifts		
	Exosuit		No Exosuit	Exosuit		No Exosuit	Exosuit		No Exosuit	Exosuit		No Exosuit
T1	234.0 ± 71.2		223.6 ± 72.5	231.9 ± 65.4		234.8 ± 73.4	246.5 ± 46.8		234.1 ± 42.5	233.4 ± 39.8		224.9 ± 35.3
T2	432.2 ± 94.0		407.5 ± 92.9	389.4 ± 83.1		386.3 ± 82.6	331.4 ± 44.4	>	311.6 ± 45.0	302.4 ± 39.7	>	285.0 ± 41.2
T3	579.2 ± 102.8		551.7 ± 98.0	512.0 ± 90.1		508.3 ± 82.5	337.0 ± 37.3	>	315.3 ± 42.2	305.5 ± 35.9	>	284.9 ± 39.9
T4	695.8 ± 114.0		675.1 ± 106.5	606.1 ± 97.3		613.0 ± 88.6	315.5 ± 42.4		314.6 ± 48.3	282.9 ± 41.4		286.0 ± 47.0
T5	765.1 ± 125.1		766.1 ± 122.5	653.0 ± 105.5		689.9 ± 100.9	284.2 ± 42.9	<	324.9 ± 42.3	245.5 ± 43.8	<	298.8 ± 41.6
T6	846.9 ± 149.6		863.8 ± 154.7	708.1 ± 128.0	<	769.6 ± 124.1	273.5 ± 52.8	<	340.8 ± 60.7	231.0 ± 50.5	<	316.7 ± 54.0
T7	1081.0 ± 204.8	<	1129.2 ± 207.0	885.8 ± 166.0	<	993.5 ± 157.1	283.6 ± 52.2	<	352.7 ± 60.0	246.2 ± 48.7	<	334.3 ± 51.9
T8	1366.2 ± 259.9	<	1461.0 ± 261.3	1113.4 ± 206.4	<	1286.9 ± 198.7	327.7 ± 55.0	<	384.4 ± 57.8	294.8 ± 52.0	<	369.7 ± 50.7
T9	1545.5 ± 280.5	<	1655.5 ± 285.5	1269.4 ± 227.5	<	1471.1 ± 225.2	363.7 ± 57.1	<	399.0 ± 58.4	332.3 ± 53.4	<	384.7 ± 51.6
T10	1824.7 ± 322.1	<	1959.0 ± 326.9	1513.3 ± 263.2	<	1754.3 ± 261.3	435.7 ± 68.9		440.5 ± 67.9	406.9 ± 65.4	<	426.8 ± 57.6
T11	2157.1 ± 379.5	<	2306.7 ± 381.9	1812.1 ± 312.6	<	2079.3 ± 305.2	370.6 ± 62.1	>	344.6 ± 62.2	371.7 ± 63.6	>	349.9 ± 52.9
T12	2318.5 ± 394.7	<	2437.1 ± 397.6	1984.0 ± 333.4	<	2217.3 ± 321.7	216.1 ± 51.5	>	182.3 ± 53.6	257.4 ± 51.6	>	210.8 ± 42.1
L1	2445.4 ± 396.6	<	2523.2 ± 396.3	2145.1 ± 346.8	<	2324.6 ± 325.8	157.6 ± 28.9		161.9 ± 28.6	177.8 ± 39.9	>	162.4 ± 22.7
L2	2549.4 ± 406.0		2584.2 ± 397.5	2288.6 ± 372.7	<	2402.5 ± 329.2	125.0 ± 25.2	<	140.7 ± 24.9	133.4 ± 38.5		133.4 ± 22.4
L3	2571.0 ± 410.9		2564.7 ± 397.3	2351.4 ± 395.7		2400.2 ± 331.6	138.2 ± 41.0	<	186.0 ± 50.9	114.4 ± 30.9	<	161.0 ± 44.9
L4	2629.1 ± 420.1		2603.1 ± 409.6	2431.1 ± 415.7		2450.3 ± 350.4	150.1 ± 50.9	>	136.5 ± 47.9	177.3 ± 47.5	>	155.2 ± 48.9
L5	2919.1 ± 458.1		2889.3 ± 458.0	2716.5 ± 461.5		2735.0 ± 406.2	642.4 ± 197.9	<	663.7 ± 221.8	624.2 ± 196.1	<	662.6 ± 209.5
Total	1585.9 ± 921.4		1623.6 ± 939.2	1388.9 ± 840.6		1489.2 ± 867.2	294.0 ± 140		307.9 ± 146.7	278.7 ± 132.1		296.9 ± 141.2

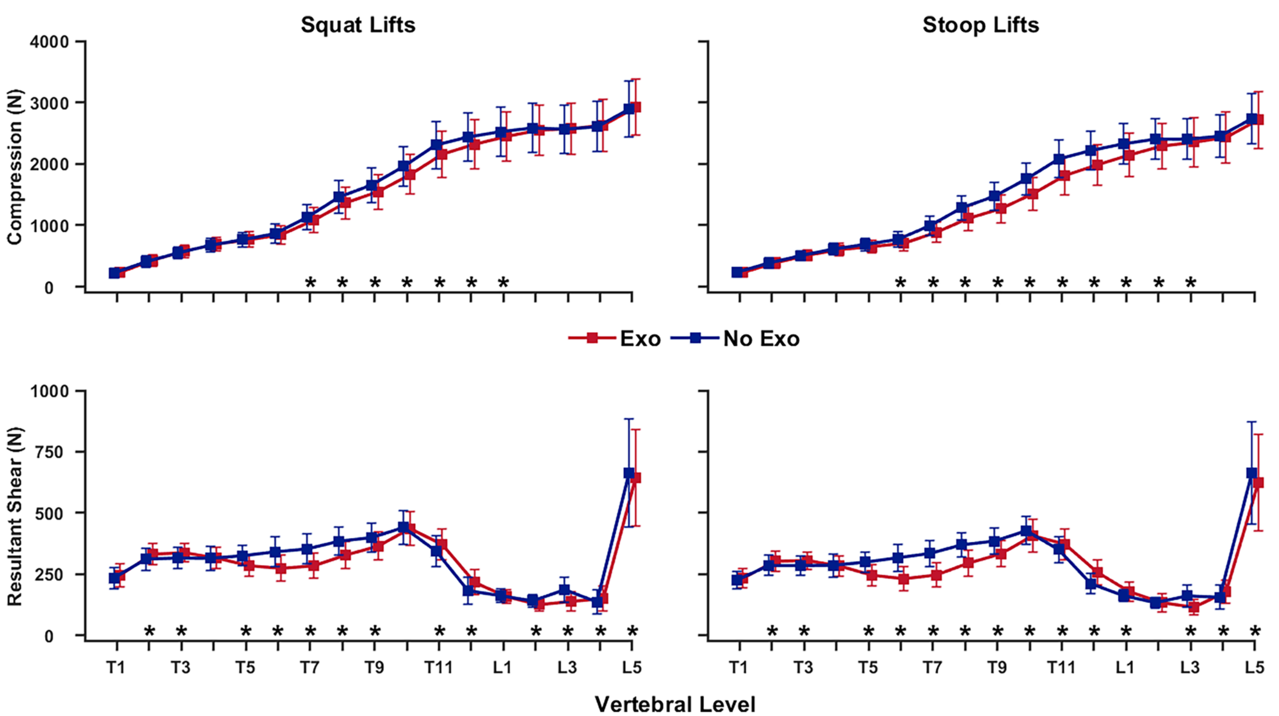

Figure 4. Peak vertebral compressive (*top row*) and resultant shear (*bottom row*) forces (N) across the seventeen thoracolumbar vertebrae during the exosuit (*red*) and no exosuit (*blue*) conditions and the two lift types (*left column*: squat lifts, *right column*: stoop lifts). Asterisks immediate to the tick marks above the abscissas indicate significant differences from post hoc analyses between exosuit conditions at those vertebral levels. Data whiskers represent standard deviations about the mean.

soft platform design and delivering lower forces when an individual is flexing, this exosuit was designed to minimise movement restriction while providing comparable lifting assistance (Chung 2023; Quirk et al. 2023a; 2023b). Previous work on exosuits, both active and passive, have reported reductions in trunk flexion and/or velocity, which can hinder adoption in the field (Ali et al. 2021; de Looze et al. 2016; Kermavnar et al. 2021; Siedl, Wolf, and Mara 2021). While this study demonstrated the exosuit was successful in preserving trunk motion, it did present trade-offs in lumbo-pelvic motion. Across the four lifting tasks, the exosuit generally increased spinal flexion/extension RoM, with a corresponding reduction in movement at the pelvis. A study of a similar cable-driven exosuit also reported increased lumbar RoM during exosuit use but, like most all previous work, did not report pelvic or overall trunk movements (Li et al. 2021). While the reason for the observed trade-off effect presented here is unclear, the exosuit force applies a moment simultaneously to both the hip and lumbar spine. Hypothetically this may be more restrictive to hip flexion than to lumbar flexion, resulting in the user favouring the latter. Alternatively, local kinematic changes could reflect systematic motion artefacts from taut actuating cables inducing movement on clothing-mounted sensors (anterior pelvic tilt to shorts and/or posterior torso tilt to the shirt). Additional research including transparent exosuit conditions and skin mounted sensors to separate lumbo-pelvic motion is necessary to test whether this effect changes as the user becomes more familiar with the exosuit, and if the magnitude of the reported kinematic adaptations are biomechanically meaningful.

Average compressive force while wearing the exosuit was reduced by roughly 8% at the lower thoracic and upper lumbar levels. This magnitude of reduction in spinal compression is in line with the 5–27% reduction in spinal compressive forces reported for other exosuits and exoskeletons at the L4/L5 or L5/S1 intervertebral joints (Abdoli-Eramaki et al. 2007; Kermavnar et al. 2021; Madinei and Nussbaum 2023; Schmalz et al. 2021; Ulrey and Fathallah 2013). However, our study identified no reduction in spinal compression at the L4 or L5 vertebrae. The biomechanical efficacy of a back exosuit is dependent upon design features including device mass and the magnitude, direction, and location of the assistive force. Regarding device design, rigid devices typically apply moments about a hinge aligned with the lower lumbar joints/vertebrae, thereby delivering more perpendicular forces acting on the thoracic spine and upper trunk (Koopman et al. 2020; Madinei et al. 2020; Schmalz et al. 2021). Soft devices applying parallel assistance, like the one presented in the current study, generate moments about the back in a different manner as compared to rigid hinged devices. They work by delivering external assistance via a cable(s) with a moment arm that is larger than the *in vivo* extensor muscle moment arms of the trunk (Figures 1 and 2; Abdoli-Eramaki et al. 2007; Lamers, Yang, and Zelik 2018). Further, the moment arm length of soft parallel devices varies at each vertebral level as individuals move throughout the lift cycle (see Figure 5 for a post hoc comparison of the moment arm of the exosuit cord relative to the intervertebral joints of the musculoskeletal model for both lift types). The variation in moment arm length is particularly evident at the lower lumbar levels where the exosuit assistive cord moves closer to the skin during peak flexion/lordosis at lift off.

The exosuit generally reduced peak vertebral resultant shear force, by an average of 16% in the mid

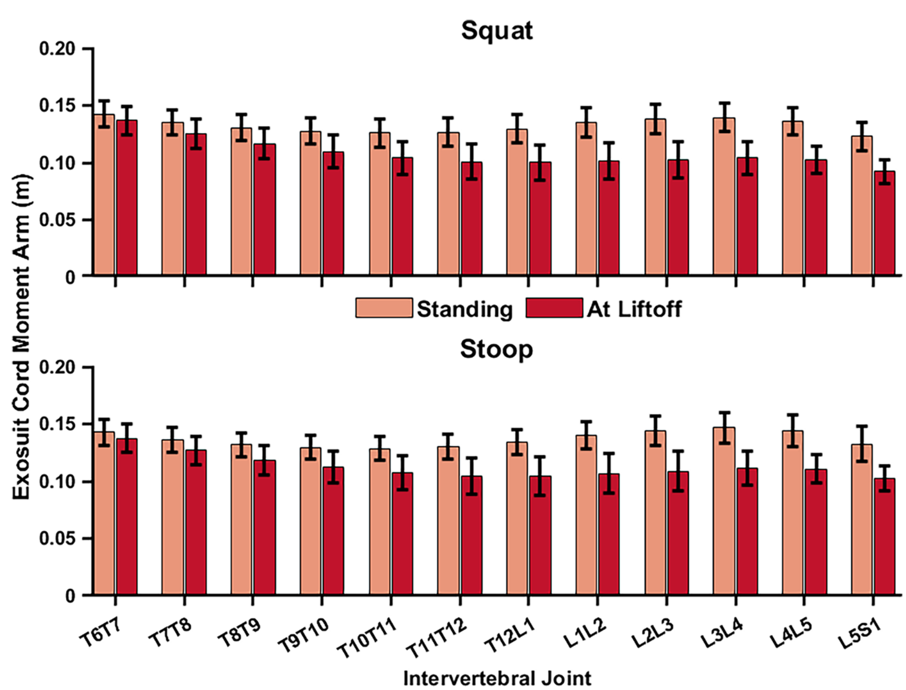

Figure 5. Exosuit cord moment arm (in metres) about the intervertebral joints at standing (*salmon*) and lift-off (*crimson*). The effective moment arm was calculated within OpenSim (Sherman, Seth, and Delp 2013). Note: the exosuit cord did not extend across the entire thoracolumbar spine.

thoracic and lower lumbar vertebrae. This level of overall reduction is greater than the 8–12% reduction reported in the limited number of studies that have estimated shearing at a single lower lumbar level (Abdoli-Eramaki et al. 2007; Picchiotti et al. 2019; Ulrey and Fathallah 2013). We observed a ~7% reduction in L5 shear that is comparable to these previous reports. In addition to reductions in shear at many vertebral levels, we also observed small increases in shear loading at the T3, T11, and T12 vertebrae. While any increase in spinal force is undesirable, it is important to note the peak shear loading only significantly increased by about 25 (15–45) N, and these resulting shear magnitudes of only about 260 (180–370) N (Figure 4 and Table 2) were still far below the 700 N threshold for shear-related chronic injury (Gallagher and Marras 2012). In the one vertebral level (L5) that did approach and exceeded shear thresholds in the control condition the exosuit condition successfully reduced loading (Table 2). It is difficult to state for certain how the shearing can increase at a few vertebral levels while decreasing in others both cranially and caudally. This might be an artefact of the complexities of estimating shear loading using a musculoskeletal model (Kingma, Faber, and van Dieën 2016). Because of occlusion and friction from the exosuit, this study relied on a limited number of motion capture markers and impose kinematic constraints to estimate thoracolumbar kinematics (Alemi et al. 2021). Model assumptions and imposed kinematic constraints could also lead to errors in single level intervertebral shear force estimates. Future work should look to better understand this phenomenon with either finite element models or a complementary modelling approach (Naoum et al. 2021).

This study is the first, to our knowledge, to report on the impact of an exosuit on the entire thoracolumbar spine while also incorporating the exosuit as part of the musculoskeletal model. By modelling the assistive cord path, rather than assuming it maintained a constant moment arm from each intervertebral joint, we were better able to account for variations in the curvature of the spine and assistive force path and is a potential reason why a significant reduction in the lower lumbar compressive forces was not found. If the goal of an exosuit is to reduce spinal loading to a specific level, one approach would be to increase the moment arm between the cable and the spine (Abdoli-Eramaki et al. 2007; Zelik, Lamers, and Scherpereel 2020). Further reductions in spinal loading might also be possible through increasing the exosuit assistive force and/or improvements in exosuit force timing.

It is unknown whether the magnitude of reductions (and increases) we observed in peak loading on the vertebrae are practically meaningful in a biomechanical, physiological, or ergonomic sense. This work was not intended to establish what level or kinetic metric has an impact during occupational tasks. Likewise, it is unknown if any non-significant changes in peak force are influential. Worth noting, average vertebral forces over the duration of the lift displayed similar trends to the peak forces (though for brevity they were not presented here). Further, despite being commonly associated with lower back pain (Deyo and Weinstein 2001), we did not report soft tissue (i.e. muscle, tendon, ligaments) force outcomes. However, with repetitive use, any exosuit related reductions in spinal forces could potentially reduce cumulative damage (Zelik et al. 2022). Future work will quantify how this exosuit can impact soft tissue loading, muscular effort and fatigue, and injury rates.

Four different sagittal plane lifting tasks, comprising of two crate masses and two lifting techniques, were performed. Prior studies have established that lifting style and weight lifted can affect spine loading (Chaffin and Park 1973; van Dieën, Hoozemans, and Toussaint 1999; von Arx et al. 2021; Zander et al. 2015), and these findings were reaffirmed by main effects in our primary analyses. In our study, stoop lifting had lower spinal loading than squat lifting. However, it is important to note that stoop lifting was performed with a taller box than squat lifting in attempts to normalise task-related trunk flexion. We also found that spine loading was affected by the exosuit, varying by vertebral level and lifting style, but not with weight lifted. Focusing our analysis on exosuit interactions within a lifting style, we found the exosuit reduced compressive loads across more vertebral levels in stoop versus squat lifting. Although these task-related differences were not large, they could be explained by two factors. First, despite attempting to make trunk flexion comparable between lifting types, trunk flexion was lower in the squat than the stoop lift (Figure 3). Given the exosuit delivers assistance relative to trunk flexion angle (see Chung 2023; Quirk et al. 2023a; 2023b), such kinematic differences could lead to distinct assistance levels (Quirk et al. 2023a). Secondly, there was a reduced effective moment arm of the exosuit cord relative to the intervertebral joints at the point of peak spinal loads during squat lifts (Figure 5).

Some limitations in this methodology should be acknowledged. An established thoracolumbar musculoskeletal model (Alemi et al. 2023; Bruno et al. 2017, Bruno, Bouxsein, and Anderson 2015; Burkhart et al. 2020) was utilised to evaluate pelvis, spine, and trunk sagittal kinematics and vertebral loading. The models were individualised to each participant based on gender, height, weight, and measurements collected in the static calibration trials. While participant-specific modelling based on medical imaging can affect spine loading outcomes (Bruno et al. 2017; Mokhtarzadeh et al. 2021), we did not have this available. Nonetheless, we have shown that model creation based on marker data alone is reliable and accurate (Burkhart et al. 2020). Kinematic constraints were applied to limit the spine kinematics to three independent degrees of freedom. We have shown that more degrees of freedom may be allowed with markers overlying the spine, but the exosuit prevented this. This can increase the error in estimating spine motion slightly and likely over-constrains thoracic motion (Alemi et al. 2021).

There are other musculoskeletal modelling limitations (i.e. muscle recruitment optimisation, inertial properties, joint degrees of freedom) that should also be considered when interpreting these results (Banks, Umberger, and Caldwell 2022). Specifically, static optimisation is known to under-predict antagonist muscle activity which can reduce the magnitude of estimated compressive forces (Banks, Umberger, and Caldwell 2022; Cholewicki, McGill, and Norman 1995). While this model has been validated for dynamic lifting using static optimisation (Akhavanfar et al. 2023; Alemi et al. 2023), estimates of spinal loading could potentially be different with the use of electromyography (EMG) driven models or other approaches to systematically force agonist and antagonist activity. Further, the spinal forces reported here do not account for passive tissue forces that also balance spinal moments. Passive tissues have different moment arm lengths and orientations than muscle fascicles, which may be of particular importance during stoop lifts (Bazrgari, Shirazi-Adl, and Arjmand 2007). However, as the primary outcomes were comparisons of exosuit effects within each subject, we would not expect these limitations in our modelling approach to change the reported outcomes. Given that the exosuit analysed in the current study does not significantly increase antagonist activity in individuals with (Quirk et al. 2023b) and without (Quirk et al. 2023a) low back pain, this limitation is systematic across the conditions tested and would allow for valid comparisons between them. Similarly, passive tissue forces are driven by kinematics, and while there were differences between exosuit conditions for torso and pelvis angle, overall trunk kinematics were similar. Finally, for this first biomechanical analysis of this exosuit design we recruited and analysed data from young healthy inexperienced participants performing restricted sagittal plane lifts over a few hours, which may not reflect the real-world usage. Additional research is needed to see how experienced workers in real-world environments are influenced by this soft active back support exosuit, including more complex tasks and longer-duration usage.

In conclusion, this study examined the kinematic and kinetic impact of a soft active back exosuit during sagittal lifting. For the first time compressive and resultant shear forces of the entire thoracolumbar spine were presented along with pelvis, spine, and trunk motion. Results indicated that this soft exosuit design did not inhibit overall trunk flexion/extension. However, there were apparent trade-offs between spine and pelvis movement. The exosuit reduced compressive loading in the lower thoracic and upper lumbar vertebrae, but not in the upper thoracic and lower lumbar vertebrae. Shear loads were generally reduced in the mid thoracic and lower lumbar levels, but the model estimated small increases in the lower thoracic vertebrae; however, assumptions in the modelling approach may have contributed to these findings. The observed overall reduction in vertebral loading is promising and would suggest that using this exosuit design as an intervention could help reduce fatigue and manual material handling-related musculoskeletal disorders in the spine, especially given its ease of use and known benefits on muscular effort (Quirk et al. 2023a), although further research is needed. This study highlights the importance of accurately modelling the exosuit and supports future research to examine the exosuit as an ergonomic intervention.

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Disclosure statement

C.J.W. and J.C. are inventors of at least one patent application describing the exosuit components described in the paper that have been filed with the U.S. Patent Office by Harvard University. Harvard University has entered into a licencing agreement with Verve Inc., in which C.J.W., and J.C. have an equity interest and C.J.W. has a board position. All other authors report no conflict of interest.

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