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Evaluating adaptiveness of an active back exosuit for dynamic lifting and maximum range of motion

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ABSTRACT

Back exosuits deliver mechanical assistance to reduce the risk of back injury, however, minimising restriction is critical for adoption. We developed the adaptive impedance controller to minimise restriction while maintaining assistance by modulating impedance based on the user's movement direction and nonlinear sine curves. The objective of this study was to compare active assistance, delivered by a back exosuit via our adaptive impedance controller, to three levels of assistance from passive elastics. Fifteen participants completed five experimental blocks (4 exosuits and 1 no-suit) consisting of a maximum flexion and a constrained lifting task. While a higher stiffness elastic reduced back extensor muscle activity by 13%, it restricted maximum range of motion (RoM) by 13°. The adaptive impedance approach did not restrict RoM while reducing back extensor muscle activity by 15%, when lifting. This study highlights an adaptive impedance approach might improve usability by circumventing the assistance-restriction trade-off inherent to passive approaches.

Practitioner summary: This study demonstrates a soft active exosuit that delivers assistance with an adaptive impedance approach can provide reductions in overall back muscle activity without the impacts of restricted range of motion or perception of restriction and discomfort.

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Low back pain; electromyography; biomechanics; exosuit; wearable robotics

1. Introduction

According to the National Institute of Occupational Safety and Health (NIOSH), back injuries are the most common (20%) workplace injury, representing a large economic ($100 billion per year) and personal burden in the US (Katz 2006; Ferguson et al. 2019). Back injuries are most common in manual material handling tasks, such as repetitive lifting tasks in warehouses, which place high peak and cumulative loads on the spine (Granata, Marras, and Davis 1997; Norman et al. 1998). Ergonomic interventions, such as installing rollers or electric hoists, have successfully reduced back injuries (Marras 2000). However, they are often expensive, and their usage is limited to certain locations. Back belts have been developed to mitigate the risk of back injury without this restriction to a specific location. However, there hasn't been clear evidence of back belts reducing back muscle activity or back injuries (van Poppel et al. 2000; van Duijvenbode et al. 2008).

In recent years, more advanced technology has led to the development of back exos, including exoskeletons and exosuits. Back exos generate assistive moments to the back of the wearer. A number of biomechanical studies have shown that back exos successfully reduce lumbar moments and the back extensor muscle activity, measured as electromyography (EMG) amplitudes, during repetitive lifting tasks (de Looze et al. 2016; Kermavnar et al. 2021; Ali et al. 2021). Thus, it has been postulated that back exos could mitigate the risk of back injury in the workplace by offloading back exertion if they can be worn throughout the shift (Zelik et al. 2022).

Depending on the actuation mechanism and the power source, back exos are categorised into passive and active devices (Toxiri et al. 2019). Passive exos use elastic components such as springs and elastic bands to apply assistive forces to the user. While the lightness and simplicity of passive systems could be well

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suited for assisting static bending tasks, their fixed impedance property poses a challenge in maximising the level of assistance for dynamic tasks (Poliero et al. 2022). Passive exos generate back extensor moments proportional to trunk flexion and the device's stiffness. While increasing a device's stiffness results in higher assistance for the same movement, it has potential trade-offs including increased restriction and discomfort for some dynamic tasks (Näf et al. 2018; Frost, Abdoli-E, and Stevenson 2009; Kim et al. 2020; Baltrusch et al. 2019; Yandell et al. 2020). For example, passive exos have been characterised as being more uncomfortable when a user engages in maximal trunk flexion movements such as picking up a pencil from the floor (Kim et al. 2020). Hence, given that device fit and comfort are key determinants for an individual's intent to use a device, there is a growing need in the literature to test these devices during these provocative tasks (Baldassarre et al. 2022; Babič et al. 2021).

Active exos use powered actuators such as electric motors and thus can offer greater adaptability for tasks by controlling assistive force profiles based on sensor data (Koopman et al. 2019). While this elevated adaptability may improve intent to use during more dynamic tasks, active exos still have burdens that need to be addressed. Active devices are often heavier and bulkier and have additional complexity (ie maintenance or charging) compared to their passive counterparts (Kermavnar et al. 2021; Ali et al. 2021; Baldassarre et al. 2022; Siedl, Wolf, and Mara 2021; Schwerha et al. 2022). Additionally, the higher adaptability of active controllers could be perceived as less predictable if it were used for a task that it was not specifically designed for (Baldassarre et al. 2022; Babič et al. 2021; Crea et al. 2021).

Previous research comparing passive and active systems has shown that each device type can offer different benefits and burdens depending on the specific task analysed (Poliero et al. 2022). However, current work in this area is limited by comparing back exos with different hardware architecture and maximum assistance characteristics (Schwartz, Theurel, and Desbrosses 2022). Given that device architecture and assistive force affect both biomechanical and perceptual efficacy, there is a need to understand both the benefits and drawbacks of active and passive assistance when confounding factors such as device weight, body interface and rigidity are carefully controlled (Kim et al. 2020; Siedl, Wolf, and Mara 2021; Schwartz, Theurel, and Desbrosses 2022; Schwartz et al. 2021).

To leverage the strengths of an active system we have developed a lightweight soft active back exosuit that uses an adaptive impedance controller. This controller applies higher impedance during trunk extension to maximise assistance while providing reduced impedance during the trunk flexion so as not to hinder forward bending. Furthermore, by employing non-linear sine impedance curves rather than linear lines, the adaptive impedance controller effectively reduces force commands for extensive trunk flexion angles, ultimately decreasing hindrances during deep bending. The purpose of this study is to compare this adaptive impedance approach to a fixed impedance (stiffness) approach, using an identical human interface and system architecture. To achieve this purpose, we compared active assistance from an active back exosuit, delivered via the adaptive impedance controller, to three levels of assistance from an exosuit with passive elastic elements. Three passive elastic bands were carefully selected with low, medium and high stiffness to match the lowering force, average force and lifting force delivered by the active system respectively. Comparing an adaptive impedance controller to these passive springs, we hypothesised the adaptive impedance controller could mitigate the measurements and perception of restriction associated with a high stiffness elastic when a participant engages in a maximal flexion task, while retaining the assistive properties (peak back extensor electromyography amplitude and moment reductions) that a high stiffness elastic affords when an individual performs a constrained lifting task.

2. Materials and methods

2.1. Participants

Fifteen participants, eleven men and four women (31 ± 4 years old, 73 ± 12 kg, 172 ± 13 cm, with a BMI of 25 ± 5 kg/m^2) volunteered for this study. Participants were screened ensuring they were 18–65 years old, generally healthy and engaged in moderate physical activity more than 3 h a week. Participants were excluded if they reported a recent (<6 months) history of activity limiting low back pain or any musculoskeletal or neurological conditions that could interfere with their ability to perform the experiments. Consistent with Helsinki's guidelines, all participants provided informed consent to a study approved by Harvard Medical School's Internal Review Board (IRB18-0960).

2.2. Back exosuit and experimental conditions

In this study, we developed an active back exosuit, which is capable of applying forces up to 250 N through ribbon cable driven actuation (Figure 1(A)).

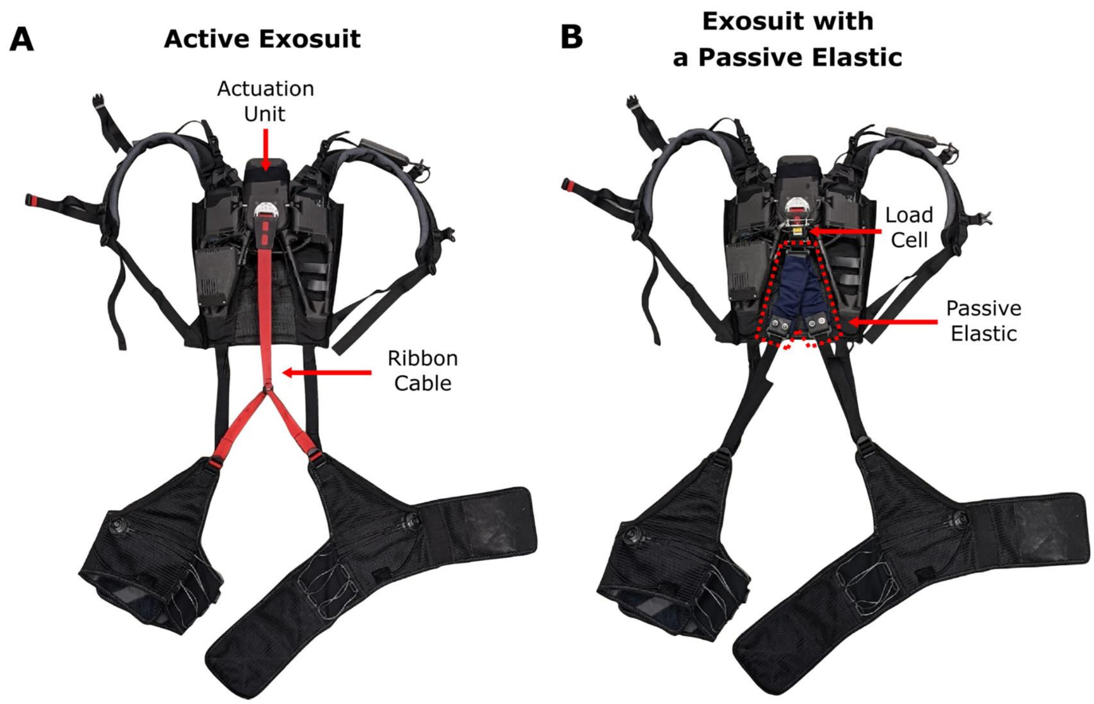

Figure 1. Visual representation of device setup for the active and passive exosuit conditions with identical human interface components. In the active condition (A), a ribbon cable applied forces delivered by the actuation unit whereas the passive condition (B) used elastics that spanned the back panel and thigh wrap.

The exosuit incorporated inertial motion sensors (IMUs) to capture movements of the back and thighs at a frequency of 100 Hz, allowing for the application of forces based on the user's movements. In this study, a generic adaptive impedance controller is used for all the participants without tuning or optimisation to each individual. This controller is a type of impedance controller that generates force commands as a function of trunk flexion angle and angular velocity. To calibrate 0° trunk flexion angle as a reference, participants were instructed to stand upright at the initialisation of the device.

The adaptive impedance controller provides assistance based on two principles. First, the shape of the force profile is described by nonlinear sine impedance curves as shown in Figure 2(A) (blue line). We employed a sine impedance approach to reduce sagittal plane restriction even further by yielding assistance after 90° of trunk flexion, as opposed to the ever-increasing force provided by linear impedance lines (Figure 2(A); dotted lines). An additional benefit of using sine impedance curves was that the exosuit applied 80% of its maximal force over a larger range of trunk flexion angles compared to a linear passive elastic (Figure 2(A); shaded area). Second, the adaptive impedance controller provides asymmetric assistance based on trunk angular velocity, delivering about one-third of the force during the lowering phase compared to the lifting phase (Figure 2(A); blue line). This asymmetric assistance aimed to maximise the assistance during lifting while not restricting a range of motion when reaching down to the ground. During rapid trunk flexion with an angular velocity exceeding 120°/s, the controller provides mild lowering assistance, as indicated by the lower blue curve in Figure 2(A). Conversely, during rapid trunk extension with an angular velocity lower than −120°/s, the controller provides significant lifting assistance, following the higher blue curve in Figure 2(A). For trunk angular velocities falling within the range of ±120°/s, the exosuit assistance is interpolated between the high lifting force and mild lowering force, depending on the trunk flexion velocity.

Five conditions were tested during the experiment. During the no-suit condition (NS), participants performed tasks without an exosuit to understand their natural biomechanical demands. Participants wore the back exosuit for the other four conditions. During the active suit condition (AS; Figure 1(A)), the force profile is generated based on the adaptive impedance controller described above. For the other three conditions, passive elastic (PE) bands of varying stiffness (Low, Medium and High) were installed (Figure 1(B)) to apply forces to the users. To ensure consistent force transmission mechanism between active and passive conditions, the elastic bands were anchored at the

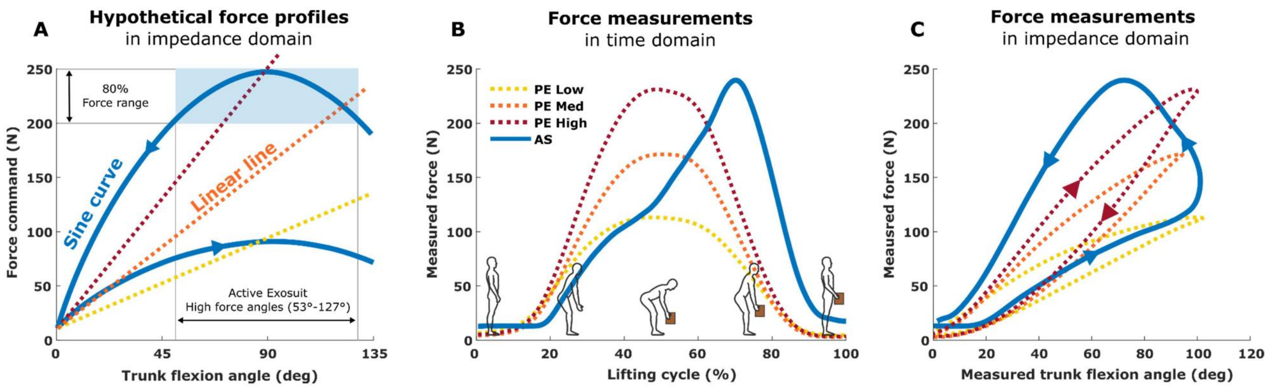

Figure 2. Hypothetical force profiles of the active exosuit and the passive elastics (A) and force measurements during stoop lifting (B–C). (A) The hypothetical force-profile of the adaptive impedance controller (blue line) depicts how the controller switches between a low stiffness elastic (yellow line) to a high stiffness elastic (red line) when lowering (forward arrow) and lifting (backward arrow) respectively. Additional to asymmetry is the sinusoidal shape of the adaptive impedance controller designed to apex at 90° of trunk flexion to minimise sources of restriction from increasing linear impedance (dashed lines) and to deliver 80% of assistance over a generous flexion range (53–127°). (B) In the time domain, the adaptive impedance controller switches between the fixed impedance states of passive elastics (PE, dotted lines), enabling it to produce low assistance during lowering (0–50%) while quickly injecting high assistive forces, similar to PE High, during lifting (50–100%). Distortion of the time-domain shape is a consequence of differences in active and passive force profiles in the impedance domain that capture (C) passive elastics apply lower forces during the trunk extension phase compared to the forces during the trunk flexion phase due to friction between the elastic bands and the user. In contrast, the active exosuit applies higher forces during the trunk extension phase.

point where the red ribbon cable of the active exosuit exits while the motor is turned off. These passive elastics were carefully tensioned, after multiple lifts and bends to remove system slack and thigh wrap slipping, to deliver a peak force designed to emulate the AS during a stoop task as illustrated in Figure 2(A) (Goršič et al. 2021). During stoop lifting, PE Low produced a peak force of 113 ± 8 N comparable to the active exosuit (AS) during the trunk flexion. PE High produced a peak force of 233 ± 29 N, comparable to the 230 ± 14 N AS during the trunk extension. PE Med produced a peak force of 174 ± 55 N representing the average force of the AS produced throughout the lifting cycle (Figure 2(B)).

2.3. Experimental protocol

All participants attended an informal familiarisation session and an experimental session. The order of experimental conditions was randomised using a counterbalanced Latin square. For the experimental session day, participants were prepared for data collection including the placement of IMUs and EMG sensors.

To normalise electromyography signals, participants performed four maximum voluntary isometric contractions (MVIC) (Burden 2010). For all normalisation exercises, participants were secured to a rigid structure and asked to practice 1–2 warm-up contractions. Afterwards, the participants performed 3 repetitions of a maximum contraction where they were instructed to pull against non-elastic straps ‘as hard as possible, for 3 seconds’, with 1–2 min of rest between trials to minimise the effects of fatigue.

To normalise the trunk flexors, participants sat comfortably on a chair with their trunk at a 90° angle. Participants were secured to the chair using a chest harness and strap and performed a maximal effort crunch into the chest harness. Normalisation of the hip and knee flexors were also performed seated. Using a padded cuff around their ankle, a strap was tensioned so the participant’s knee would be in 70–80° flexion, as the participant attempted to maximally extend their right and left knee. To normalise the trunk and hip extensors, participants were positioned on a roman chair. Participants assumed a prone position on the chair with their shanks under the posterior bar and their pelvis (anterior superior iliac spine) aligned to the edge of the chair. The torso harness secured the participants to the base of the chair using a strap designed to keep their trunk parallel to their thigh upon maximal back extension (Figure S1). Following MVIC, passive reflective markers were placed for motion capture and participants completed a static calibration.

Participants performed a series of tasks progressing between conditions (AS, NS and PE Low, Med and High) according to their Latin square order. Participants started with a maximal flexion task. Standing on an aerobic step, participants were instructed to stand comfortably and ‘keep their legs straight as comfortable’ throughout this task. From standing, participants flexed their trunk timed to a metronome (50 beats per minute (bpm)). For 3 repetitions, the participants were instructed to flex down towards the floor as far as comfortable for 2 beats, to squeeze into a maximal flexion for 2 beats (Figure 3(A)), to slowly return to a comfortable standing position for 2 beats and to hold their comfortable standing position for 2 beats until the next rep. At the end of 3 repetitions, the participants completed two ten-point numerical rating scales (NRS) asking them their level of discomfort and restriction.

Participants then performed a series of repetitive lifting tasks. To mitigate the confounding influence of movement variability between conditions, these repetitive lifting conditions were constrained spatially and temporally. Constrained lifting was performed using a 10 kg box using both squat and stoop lifting styles for all five conditions. As a tertiary aim, participants performed squat and stoop lifts in no-suit condition using a 6 kg box (NS6) to contextualise the effect of the exosuit conditions. The order of lifting conditions was randomised. Squat-style lifts were performed using a 46 × 31 × 18 cm (width, depth, height) box where participants would 'bend with your knees and not round your spine' (Figure 3(B)). Stoop-style lifts used a 43 × 28 x 32 cm box with participants instructed to 'keep their legs as straight as possible' (Figure 3(C)). Different box heights were selected to increase the torso angle between tasks and to accommodate participants with low hamstring flexibility. All lifts were constrained to a 50-bpm metronome and verbal cues. A lifting cycle lasted 4.8 s (4 beats (B1–B4)). Lifting with their elbows 'as straight as possible' on B1 participants would flex 'down' to the mass, B2 lift the mass 'up', B3 lower the mass 'down', B4 extend 'up' to standing. Participants completed 10 repetitions with 7.2 s of rest between each lift. At least 1 min of rest was provided before a participant moved onto another mass lifting style. Participants' comfortable foot placement was marked over 2 force plates (AM6800, Bertec TM, Columbus, OH). Mass origin was aligned vertically on a block to ensure a 32 cm high mass was aligned to tibial tuberosity and horizontally to what the participant deemed natural which was marked with masking tape.

2.4. Data collection setup and processing

2.4.1. IMU and suit data

Three IMUs (IMUs, MTi-3 AHRS, Xsens Technologies B.V., Enschede, the Netherlands) were positioned on the 8th thoracic spine and right and left posterior thigh, approximately in line with the middle of the glutaeal fold and popliteal fossa. These sensors were secured to the participant's skin using adhesives. Suit force (Load cell) and IMU sensor data were directly sampled at 200 Hz using an eight-bit microprocessing unit (PIC18F25K80, Microchip Technology, Inc., AZ, USA) and an onboard flash memory card (SDSQUNC-032G-AN6IA, Scandisk, CA, USA). IMU angular data were post-processed using custom Matlab code and corrected using a zero-lag 4th order 2 Hz low pass filter.

2.4.2. Electromyography (EMG)

Following standard skin preparation, bar surface electrodes (10 mm interelectrode distance) were positioned over 8 muscle sites from 4 muscle groups (back extensors, trunk flexors, hip extensors and the rectus femoris) using standardised guidelines and minor adjustments based on palpation (Figure S2). Muscle sites included for the back extensors: the thoracic (T95) and lumbar

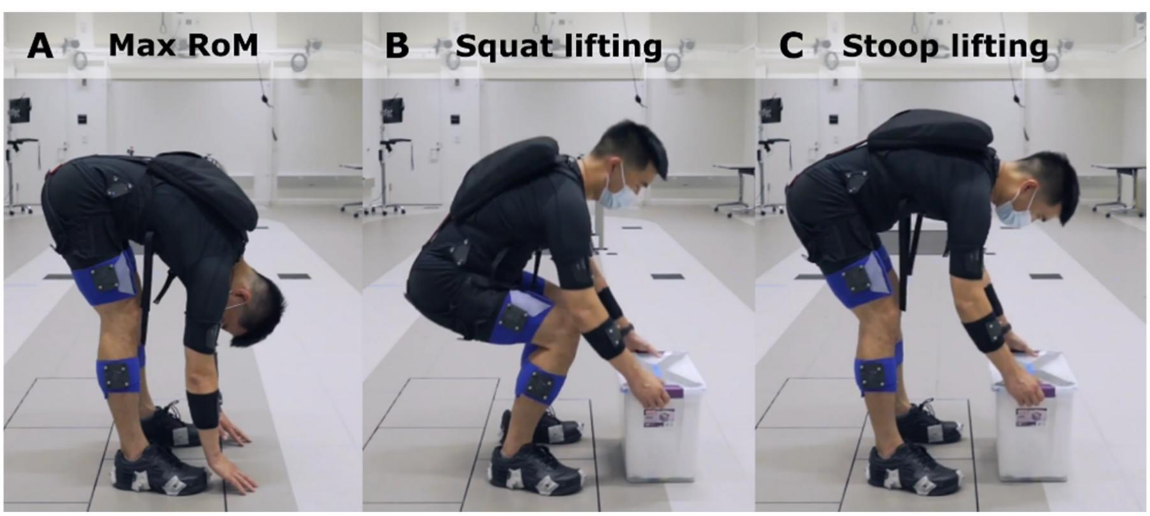

Figure 3. Demonstration of the range of motion (A), squat lifting (B) and stoop lifting (C) tasks.

erector spinae iliocostalis (L16) (5 cm lateral to the 9th thoracic and 1st lumbar spinous process, respectively) and the erector spinae lumbar longissimus (L33) (3 cm lateral to the 3rd lumbar spinous process). For the trunk flexors, signals were monitored from the upper rectus abdominis (URA) (3 cm lateral to the Linea alba) and middle external obliques (EO) (15 cm lateral to the umbilicus oriented 45° to the linea alba). SENIAM guidelines were used to position the glutaeus maximus (GM), biceps femoris (BF) and rectus femoris (RF). EMG signals were amplified, digitised (2148 Hz) and filtered (Hardware band-pass 20–450 Hz) using a series of Duo wireless bioamplifiers and EMGWorks Software (Delsys Inc., Natick, MA).

EMG signals were band-pass filtered (50–450 Hz), rectified and converted to a 6 Hz low-pass linear envelope (Drake and Callaghan 2006). Signals were amplitude normalised to the maximum 500 ms moving average signal from the MVIC trials and time normalised from 0 to 100% over 4800 points across the entire lift cycle using the IMU events (Burden 2010). Ensemble average signals were generated for each trial, condition and muscle. The primary outcome measure *peak EMG amplitude* was calculated from the ensemble average waveform. Peak EMG was calculated for the lifting (0–50%) and lowering (50–100%) phase of the task. These peak measures were then categorised according to the primary muscle group (back extensor, hip extensor, trunk flexor and rectus femoris) for statistical analysis. For completeness single muscle sites were evaluated in a tertiary statistical analysis (see Supplementary Material).

2.4.3. Kinematic motion capture data

Standard laboratory-based motion capture analysis was conducted. Passive reflective markers were positioned bilaterally on the: radial and ulnar styloid, the medial and lateral malleolus, the medial and lateral femoral epicondyles, the greater trochanter, the acromion, the sternoclavicular joint, the anterior and posterior-anterior iliac spine. An individual passive reflective marker was positioned on the 7th cervical spinous process. Four marker rigid body clusters were positioned bilaterally on the iliac crest, thigh, shank, and 6 markers were placed bilaterally on the participant's shoe at the toe, heel, 1st and 5th metatarsal and a medial and lateral location between the metatarsal and the heel (Figure S2). Following setup, a standing calibration (Inverted Y-Pose) captured the three-dimensional marker position relative to the rigid bodies using sixteen infra-red emitting cameras (Oqus 700, Qualisys). For the constrained lifting tasks, ground reaction forces were measured separately from the participant's left and right foot positioned on a force plate (Bertec^{TM}). All kinematic and kinetic data were sampled at 200 Hz at ±5 V using a 16-bit analog-to-digital board (230599, Qualisys^{TM}) in Qualisys Track Manager.

Kinematic and kinetic (force plate) data were post-processed in Visual3D (CMotion Inc., Kingston, ON). All kinematic and kinetic data were 4th-order low-pass filtered at 6 Hz. Coordinate systems axes of the foot, shank, thigh, pelvis and torso were defined using anterior-posterior (Y), medial-lateral (X) and axial (Z). Three-dimensional relative Euler angular kinematics were calculated around the ankle, knee, hip and trunk using a flexion-extension, ab-adduction and axial-rotation rotation sequence. For the constrained lifting tasks, a bottom-up inverse-dynamics approach was used to calculate overall moments acting around the ankle, knee, hip and trunk using a series of Newton-Euler equations. Using this bottom-up approach the mass of the exosuit was measured by the force plate, and inertial properties of the torso were not relevant to calculate the proximal (L5/S1) pelvis or lumbar moment. Given the minimal mass (150 g) of each thigh wrap, inertial properties of the thigh wraps were not modelled. All relative angular kinematics and inverse dynamic moments in the sagittal plane were time normalised from 0 to 100% over 480 data points using a quadratic spline interpolation algorithm defined by relevant IMU trunk motion events.

2.4.3. Suit forces and biological moment

The inverse dynamics represents the overall or net moment of a joint, without consideration of the external moments produced by an exosuit. Tensile forces delivered by both the active and passive exosuit were directly measured via load cells (LSB200, FUTEK Advanced Sensor Technology, Inc., CA, USA) placed in the actuation unit, or in parallel with the elastic (Figure 1) to measure tensile forces on the ribbon and were sampled at 200 Hz (see IMU and Suit Data).

Measured tensile forces were corrected using a zero-lag 4th order low pass filter at 2 Hz in Custom Matlab code. Corrected tensile force data were assumed to be pure acting directly at the torso anchor point load cell in the sagittal plane only. Tensile forces were converted to an extensor moment around the lumbar L5/S1 joint centre by assuming a constant moment arm length of 0.12 m when considering the flesh margin between PSIS and L5/S1 (Reed, Manary, and Schneider 1999). Around the hip; tensile forces were converted to a moment around the hip joint centre, first assuming there was a 15% reduction in tensile force when applied to the thigh wrap attributed to friction, and second assuming the hip joint centre had a constant moment arm length of 0.15 m (Lamers and Zelik 2021; Kim et al. 2019). These lumbar and hip joint moments are referred to as suit moments, which were time normalised from 0 to 100% over 480 data points.

These suit moments were utilised in future calculations to measure the biological trunk and hip moment by directly subtracting these suit moments from the respective trunk and hip overall moment. For this study we consider the product of the overall moment corrected for the exosuit moment to represent the 'biological moment' to which biological tissues must respond. The peak suit and biological lumbar moment were considered primary outcome measures. Peak moments were calculated for the lifting (0–50%) and lowering (50–100%) phases of the task.

2.4.4. Movement phase segmentation

All the biomechanical data were synchronised using a common signal logged by all equipment and segmented into two phases using IMU data for in-depth analysis. The lifting phase includes the unweighted trunk flexion and the weighted trunk extension, ie bending down without a weight and lifting a box from the ground. The lowering phase includes the weighted trunk flexion and the unweighted trunk extension, ie lowering a box to the ground and coming straight back up without a weight (Figure 6(A)). Movement events were determined using relative trunk (T8-Right thigh) angular velocity in the sagittal plane. The beginning of the trunk flexion was defined as the time at which trunk flexion velocity exceeds a threshold of 5°/s for at least 20 ms (T0). The end of the trunk extension was defined as the point at which extension velocity was below 5°/s for at least 20 ms (T100). Events were used to time-normalise all subsequent kinematic, kinetic (exosuit load cell and force plate) and EMG data.

2.5. Statistical analysis

Statistical analysis was performed using Linear Mixed Model (LMM) ANOVAs to test all hypotheses. To correct for family-wise error significance was Bonferroni corrected for four co-primary outcome measures ($\alpha = 0.05/4$ or 0.0125) to prevent type one error. Secondary outcome measures, often sought to determine no difference between conditions. To confirm a null-hypothesis it is recommended to implement a more liberal adjustment to minimise the risk of type two error thus α was set to 0.10/12 or 0.008 to correct for multiple comparisons (Schuirmann 1987). Within ANOVA significant interaction & main effects were post-hoc tested using Tukey's HSD. Violations of data normality and linearity were remedied using transformations suggested by the Johnson's test in Minitab 19 (Minitab LLC, State College, PA). Within the manuscript, only significant conditions (exosuit vs no exosuit) main effects or interactions were expressed.

Prior to this study sample sizes were calculated using a paired t-test approach to determine fourteen participants would be needed to detect significant reductions in peak back extensor EMG amplitudes (effect size: $d = 1.08$) comparing between exosuit and no-exosuit conditions (Graham, Agnew, and Stevenson 2009), or to detect reductions in peak back extensor moments between varying elastic stiffness (effect size: $d = 0.83$) (Frost, Abdoli-E, and Stevenson 2009), with 80% power and an $\alpha = 0.05$.

For the maximal flexion task, the primary outcome measures are: trunk angular displacement and NRS perceptual restriction. They were analysed using a LMM ANOVA performed across the 5 Conditions (AS, NS and PE Low, Med and High). A secondary outcome measure, NRS perceptual discomfort, was modelled similarly.

For the repetitive lifting experiment, the primary outcome measures are peak back extensors EMG amplitudes and peak biological lumbar extensor moment. A three-factor LMM ANOVA included the following factors: (i) Condition (6 - AS, NS (10 kg), NS (6 kg) and PE Low, Med and High), (ii) Phase (two-lifting and lowering) and (iii) Style (two-squat and stoop).

A secondary analysis was included to understand whether conditions would modify: peak overall (net) external lumbar and hip moments, peak EMG amplitudes of the hip extensors, knee extensors, abdominals and peak biological hip extensor moment during the lifting task using the same 3 factor LMM listed above. The suit moment was compared using a modified 3 factor LMM that included 4 conditions (AS, PE Low, Med and High) as a secondary outcome measure. Finally, an additional secondary analysis was performed to understand whether conditions would result in kinematic trade-offs at the ankle, knee, hip and trunk. For this analysis, maximum sagittal plane angular displacement was analysed in a two-factor LMM including i) Condition (6) and ii) Style (2).

3. Results

During the maximum RoM task, trunk angular displacement was not significantly reduced in AS compared to

NS (condition main: F(4,56)=12, $p < .001$, η^2_p =0.463). For the passive conditions, RoM of PE Low was not reduced relative to NS. However, RoM was reduced for PE Med (9°, 5%) and PE High (13°, 9%) (Figure 4(A)). Perceived restriction had magnified results compared to the trunk flexion angle (Figure 4(B)). The perceived restriction was not statistically different for AS compared to NS, however, all passive conditions significantly increased perceived restriction compared to NS (condition main: F(4,56)=22, $p < .001$, η^2_p =0.609). The magnitude of perceived restriction increased with higher elastic stiffness; PE High was the most restrictive, and PE Low was the least (Figure 4(B)). Similar findings were identified for the secondary outcome measure perceived discomfort (Condition main: F(4,56)=6, $p = .001$, η^2_p =0.310, Table S1).

For the repetitive lifting tasks, both peak EMG and biological lumbar moments responded similarly to the exosuits regardless of lifting technique as shown in (Tables S2 and S3) (no condition x style interactions $p > .05$). Therefore, the remainder of the results section will present data as an ensemble average of both lifting styles when highlighting the differences between lifting with and without an exosuit.

During the repetitive lifting tasks, the AS reduced peak back extensor EMG by 15% compared to the NS condition (Figure 5(B)). All passive conditions reduced back extensor EMG compared to the NS condition. The magnitude of reduction depended on elastic stiffness. PE Low reduced peak EMG the least (7%) and PE High decreased EMG the most (13%) (Condition main: F(5,318)=15.5, $p < .001$, η^2_p =0.196). There were no significant changes in peak EMG amplitudes from synergist (hip and knee extensors) and antagonist (abdominals) muscle groups (Table S2) comparing the NS condition with any exosuit condition (Table S2).

Regarding lumbar moments, AS reduced the biological peak lumbar moments by 11% compared to NS during the repetitive lifting task (Figure 6(A)) (Condition main: F(5,295)=78, $p < .001$, η^2_p=0.570; Table S3). Passive conditions reduced biological lumbar moments dependent on elastic stiffness. PE Low resulted in the smallest reduction (7%) while PE High reduced moments by 14%. Compared to passive conditions, AS reduced biological lumbar moments similar to PE Med (11%) (Table S3). The ability of PE High to reduce biological lumbar moments more than AS is partially explained by PE High delivering higher peak suit lumbar extensor moments than AS during the squat lifting task (Condition*Style: F(3,194)=67.4, $p < .001$, η^2_p=0.510; Table S3).

Biological lumbar moments were not uniformly reduced throughout all phases of lifting (Figure 6(B)) (Condition x Phase F(5,295)=4.6, $p < .001$, η^2_p=0.072; Table S3). For the adaptive AS, biological moment reductions depended on the movement phase. During the lifting phase (Figure 6(A)), AS reduced the peak biological lumbar moments by 17 Nm, similar to PE High. However, during the lowering phase, AS reduced lumbar moments by 10 Nm, similar to PE Low (Table S3). Similar results were identified when comparing our secondary outcome measure biological hip extensor moments (Condition main: F(5,295)=27, $p < .001$, η^2_p=0.312; Table S3).

A secondary aim of the repetitive lifting tasks was to determine whether the active or passive assistance would lead to unexpected kinematic or kinetic trade-offs during lifting. Comparing overall peak lumbar

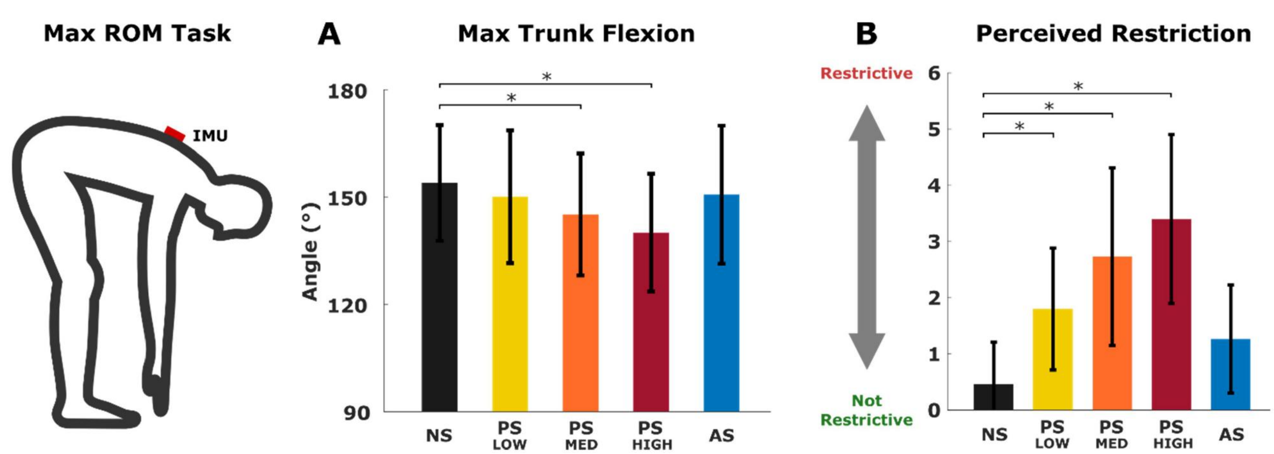

Figure 4. Maximum range of motion task results. Subplot A shows the peak trunk flexion angle participants reached during the maximum RoM task, and subplot B presents the magnitude of restriction participants perceived in numerical rating scale (NRS) upon the completion of the tasks in all five conditions. Significant condition differences between each exosuit conditions and the NS condition are noted (*). Error bars represent standard error.

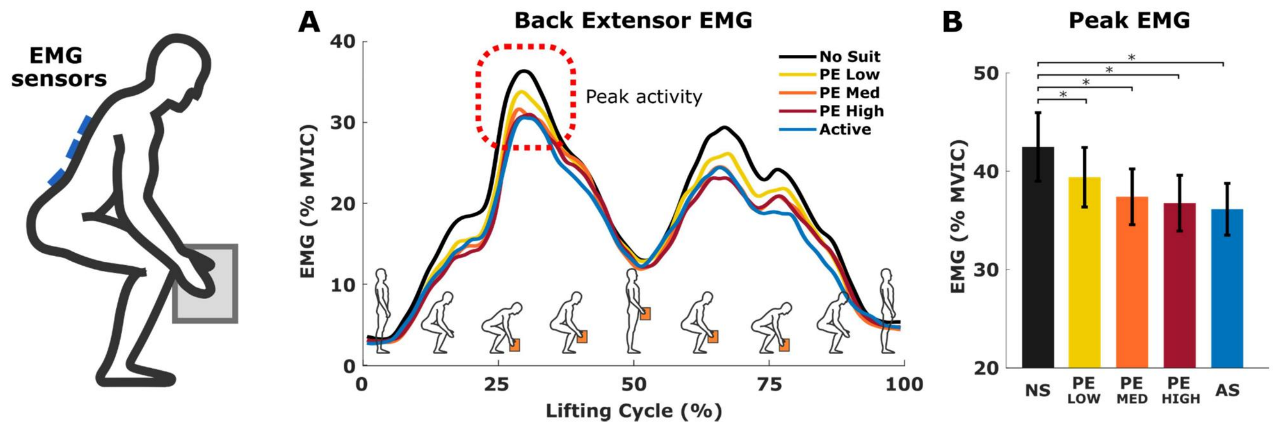

Figure 5. Back extensor EMG ensemble average across all participants and lifting styles. Subplot A shows time-series back extensor EMG data averaged across squat and stoop lifting tasks throughout the lifting cycle. Subplot B demonstrates that all exosuit conditions reduced peak back extensor EMG amplitudes compared to the NS condition. Significant condition differences between any exosuit condition and the NS condition are noted (*) and error bars represent standard error.

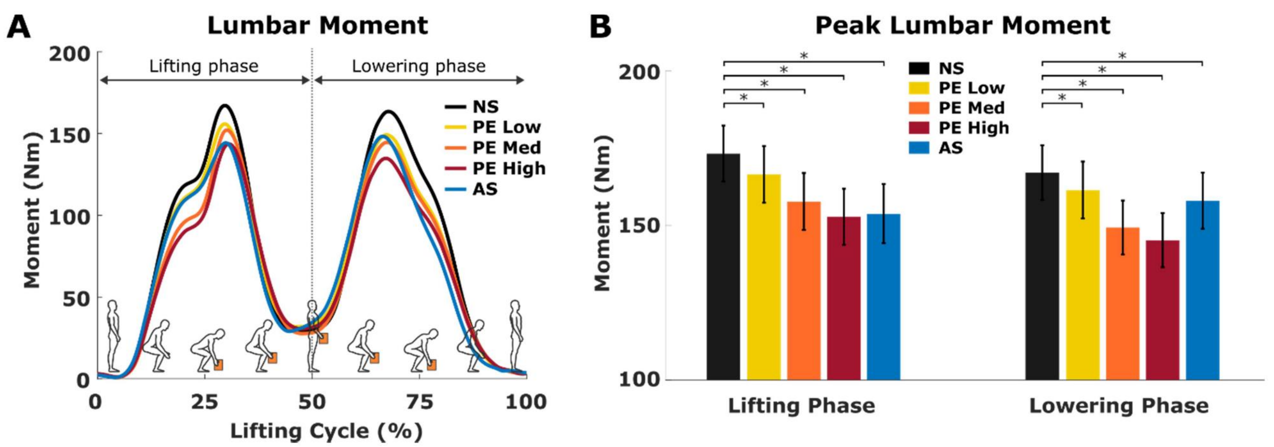

Figure 6. Lumbar moment ensemble average across all participants and lifting styles. Subplot A shows the time-series ensemble waveform comparing lumbar moments of all five conditions throughout the lifting cycle. Subplot B presents peak lumbar moments of all conditions during lifting and lowering phases. For the AS condition, the magnitude of peak lumbar moment reduction compared to the NS condition varied between the first (lifting) and second (lowering) phase of the lifting cycle. Significant condition differences between any exosuit condition and the NS condition are noted (*) and error bars represent standard error.

moments between conditions, without accounting for exosuit applied moments, we identified no significant differences (main effect or interaction) in lifting kinetics between exosuit conditions (Table S3). To understand whether there were kinematic differences, we compared maximum angular displacement around the hip, knee and ankle and determined no significant differences between conditions (Table S4). Around the trunk, PE High (124 ± 14°) significantly reduced trunk flexion angle when compared to no suit (129 ± 12°) (Condition main: $F(5,318) = 5.8$, $p < .001$, $\eta^2_p = 0.319$, Table S4).

As a tertiary aim, to contextualise the effect of peak EMG amplitude reductions, all 10 kg lifting conditions were also compared to a 6 kg lifting task performed without an exosuit (NS 6 kg). A condition main effect captured that both the PE High and AS condition effectively reduced peak back extensor EMG amplitudes to a magnitude comparable to lifting a 6 kg mass (Table S2). See Supplementary Results for more details.

4. Discussion

By comparing assistance from an exosuit delivered by active and passive approaches during the maximum RoM task and the repetitive lifting tasks, this study highlights that the adaptive impedance approach enabled by an active device can circumvent the assistance-restriction trade-off inherent to passive devices. Previous literature has alluded that while passive systems with higher assistive forces result in larger reductions in back extensor muscle activity, they generate biomechanical trade-offs such as increased antagonist muscle co-activation and potentially unfavourable movement patterns at unassisted joints (Näf et al. 2018; Kim et al. 2020; Fick 2012). Consistent with this previous work, our study provides evidence of an assistance-restriction trade-off in passive devices. While a higher elastic stiffness provides the benefit of larger back extensor moment and muscle activity reductions, it comes at the cost of increased restriction (Figure 4) (Frost, Abdoli-E, and Stevenson 2009). By actively modulating the impedance based on the user's movement direction, this adaptive impedance controller did not restrict participants compared to the no-suit condition while reducing the back extensor muscle activity as much as PE High when lifting.

By comparing passive elastic bands of varying stiffnesses our study demonstrates that the ability of the active system to minimise movement restriction was achieved by delivering asymmetric assistance. Similar to PE Low, AS did not restrict RoM during a maximal flexion task when compared to wearing no exosuit. However, there were discrepancies between these assistance methods. Survey results captured only AS was perceived to be non-significantly restrictive compared to no exosuit. This discrepancy might be explained because the adaptive impedance controller delivered non-linear assistance apexing at 90° of trunk flexion (Figure 2(A)). Given that maximum trunk flexion exceeded 150°, assistance delivered by the active system would ultimately yield, whereas assistance provided by the passive elastic increases proportionally to spring strain (Figure 2(A)) (Koopman et al. 2019). Previous work has demonstrated passive back exosuits result in increasing discomfort in tasks that require deeper trunk flexion (Kim et al. 2020), whereas our active approach would be robust to this phenomenon improving the adaptability of our device to this task.

Addressing movement restriction is a critical step towards the adoption of back exo technology since an increasing number of studies suggest that perceptual burdens with back exos, including restriction and discomfort, negatively impact an individual's intent to use a device (Hensel and Keil 2019). At the same time, applying high peak assistance is crucial for maximising the reduction in back extensor muscle activity and moments which is postulated to reduce the likelihood of injury related to spinal tissue damage (Zelik et al. 2022). In this study we demonstrated an active approach can reduce back extensor muscle activity and moments by 15% and 11% respectively. These data are in line with a previous study conducted using the same active back exosuit on 15 participants with low back pain (Quirk et al. 2023). Similar findings have been reported with other back exos reducing back extensor muscle activity and moments by 10–40% (Kermavnar et al. 2021). When comparing back extensor EMG amplitudes AS reduced peak activity similar to PE High (Table S2), with both devices capable of reducing muscle activity of lifting a 10 kg mass such that it is comparable to lifting a 6 kg mass without the suit (Figure 7). This finding was surprising considering that PE High delivered higher peak assistive forces than the AS when lifting (Table S3). Similar back extensor activity reductions achieved by AS and PE High might be explained by the timing of force delivery. Unlike a passive system that provides less assistance when lifting due to hysteresis effects (Figure 2(C)), our adaptive impedance controller delivered the highest assistance when an individual is lifting (Figure 2(B)). A period known to correspond to highest peak back extensor EMG activity (Granata, Marras, and Davis 1997).

Although this paper highlights some benefits of an active approach, there are positive attributes that favor a passive system. While AS and PE High achieved similar reductions to peak biological back extensor moments as much as PE High during the lifting phase, the reduction of peak moment achieved by AS condition was inferior to PE High during the lowering phase (Figure 6(B)). This can be explained by the lower assistance delivered by the adaptive impedance controller during the trunk flexion regardless of the external load (Figure 2(B)). For some scenarios, it could be beneficial to assist more during the weighted flexion phase, in particular when a user lowers a weight to

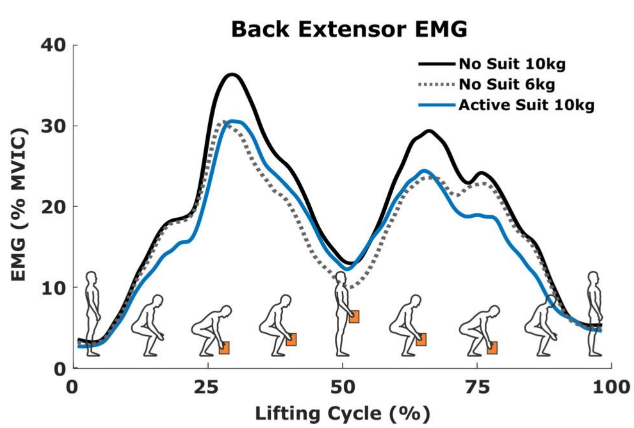

Figure 7. Back extensor EMG timeseries waveform: active condition vs. no suit condition. Back extensor EMG data demonstrate a clear difference in muscle activity when lifting a 10 kg box (solid black) and a 6 kg box (dotted black) with the no suit condition. Lifting a 10 kg box with the active exosuit (blue) achieved similar peak EMG activity compared to lifting a 6 kg box with no suit (dotted black).

the ground (Willoughby, VanEnk, and Taylor 2003; de Looze et al. 1993). In the future, active devices could merge the benefits of this asymmetric controller approach while incorporating controllers that automatically detect an external load and adaptively increase the assistance during the trunk flexion when a user is holding a weight. However, taking this approach will also have to carefully balance how much assistance can be applied during weighted lowering before it becomes a perceptual burden.

Using an adaptive impedance controller could be an important milestone to maximise the cumulative benefit of wearing a back exosuit (Zelik et al. 2022). Compared to the passive systems, where an individual must swap out elastic bands, or change modes, to achieve a stiffness that will maximise assistance or reduce restriction, an active systems is not bound by these fixed constraints (Goršič et al. 2021). In this paper we demonstrate a single controller can allow an individual to tolerate a higher level of assistance without the perceptual burdens of discomfort and restriction that arise when lowering during an unweighted range of motion task. Ultimately, this approach might yield improvements in device usability without the need to sacrifice the benefits of high assistance necessary for injury mitigation (Poliero et al. 2022; Kim et al. 2020).

Beyond differences in the back extensor activity, our study analysed synergistic muscles that might benefit from exosuit assistance. Despite the exosuit delivering hip extensor moments (Table S3), there was no significant reduction in peak muscle activity for the hip extensors (Table S2). This finding was interesting as back exosuits that span the thigh can reduce hip extensor EMG activity (Kermavnar et al. 2021). This discrepancy could be due to our approach to the calculation of biological hip moment. In our model, we assumed a 15% reduction in exosuit forces due to friction, however, this value could be considerably higher (Lamers and Zelik 2021). Future work should validate assumptions from this hip model through direct load cell measurements. A second reason we observed no change in hip extensor activity is we did not provide a sufficient adaptation period to exosuit forces. A follow-up experiment with the same participants using the same active exosuit significantly reduced both hip extensor (11%) and knee extensor (22%) muscle activity during prolonged (1-h) usage (Chung et al. 2023). Compared to the 20 lifts during this experiment, participants lifted 320 times throughout the prolonged lifting experiment allowing participants to explore and adapt to the exosuit assistance via implicit exposure (Tassignon et al. 2021). These findings align with gait studies showing exosuit efficacy improves as an individual learns to exploit this human-robot interface with repeated utilisation (Jacobs et al. 2018; Haufe et al. 2021).

Additionally, we analysed whether the exosuit would produce undesired biomechanical side-effects including antagonist co-activation and changes in joint kinematics (Kermavnar et al. 2021; Näf et al. 2018; Yun et al. 2021). The repetitive lifting task results identified no increased antagonist activity for any exosuit condition (AS and PE High-Low). Also, no differences in joint kinematics and kinetics were found for the repetitive lifting tasks between the NS and the AS conditions. However, we captured decreased trunk angular displacement when lifting with PE High compared to lifting without (Table S4). This finding is consistent with observations from the maximum range of motion task indicating that increased elastic stiffness can restrict trunk flexion (Table S1).

Minimal disruption to an individual's natural movement is desirable to achieve exosuit usability (Ali et al. 2021; Baldassarre et al. 2022; Babič et al. 2021). Despite a historical bias towards squat being a safer lifting style, both squat and stoop lifts have unique task-dependent benefits. Therefore, it is important the exosuit does not change a user's movement pattern (Baldassarre et al. 2022; van Dieën, Hoozemans, and Toussaint 1999). Given both biological moments and back extensor muscle activity had no condition by lifting style interactions, it could be inferred that our adaptive impedance controller performs similarly between squat and stoop lifts and would not bias individuals to adopt a squat lifting style as previous studies identified with passive devices (Yun et al. 2021; Sadler, Graham, and Stevenson 2011). However, it is important to acknowledge an interaction could be observed if the participant performed squat and stoop tasks to a box of the same height. Furthermore, the lack of lifting style by condition interactions might be explained by the constraints placed on participants when performing the lifting task. While constraints offer the advantage of reducing the possibility of confounding differences between conditions through altered kinematics, there is a distinct advantage in enabling participants to attempt natural movements while being exposed to exosuit assistance. Future work should explore adding unconstrained freestyle lifting patterns to bolster these claims and better learn human interactions with exosuit technology.

This study highlights the biomechanical and perceptual impacts of the amount and timing of assistance provided by an exosuit. However, there are certain limitations to our study design. Firstly, our study compared active and passive systems of similar weight and body interface to remove confounding factors. With that said, passive systems are typically lighter than active systems, which is shown to decrease discomfort (Toxiri et al. 2019; Schwerha et al. 2022). Hence, while this study demonstrates passive assistance can be more restrictive and uncomfortable compared to our active adaptive impedance controller, it does not imply all passive devices are restrictive and uncomfortable. Secondly, the benefits and burdens of exosuit controllers are task dependent (Poliero et al. 2022). While this experiment highlights an adaptive impedance approach is beneficial for a dynamic lifting task at delivering higher assistance without the burden of perceived restriction, a passive device with the higher overall assistance might be more suitable for static assembly tasks that seldomly require maximum range of motion.

Our primary motivation was to evaluate whether an active adaptive impedance approach can increase exosuit usability without compromising assistance to improve real-world utilisation (Babič et al. 2021; Hensel and Keil 2019). While, this study was conducted over a short period of time in a strictly controlled laboratory environment is a promising first step to suggest an active approach can minimise restriction and discomfort associated with wearing a back exosuit further improving usability future studies should be developed to demonstrate this adaptive impedance approach can work robustly and provide similar biomechanical advantages in the field (Babič et al. 2021; Crea et al. 2021), as maximum exosuit assistance impacts an back exo's injury mitigation potential (Zelik et al. 2022). Additionally future studies should attempt to compare passive and active devices in a long-term real-world setting, in a working population, to demonstrate whether an adaptive impedance approach translates into increased wear time. This step is important since modelling studies suggest, the number of lifts performed with a back exosuit has a profound impact on the ability of a back exosuit to reduce cumulative spinal damage and the potential of developing LBP (Zelik et al. 2022).

5. Conclusion

Our study compared assistance from an exosuit delivered by active and passive approaches during the range of motion measurement and repetitive lifting tasks. We found that an active adaptive impedance approach overcame the assistance-restriction trade-off of passive systems by modulating the impedance based on movement direction. For the back extensor muscle activities, it was shown that lifting a 10 kg box with the active back exosuit was similar to lifting a 6 kg box without wearing the exosuit. Considering its biomechanical efficacy, minimal movement restriction and no biomechanical side effects, an active back exosuit approach demonstrates promise for use in workplace environments dominated by dynamic movement to mitigate the risk of back injury.

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Author contributions

Conceptualisation: JC, MA, CJW; Development: JC; Protocol design: JC, DAQ, MA, DD, LA, CJW; Data collection: JC, DAQ, MA; Data analysis: DAQ; Funding acquisition: JC, MA, DD, LA, CJW; Writing – original draft: JC, DAQ, JMC; Writing – review & editing: JC, DAQ, MA, JMC, DD, LA, CJW

Disclosure statement

CJW and JC 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 CJW, JC and MA have an equity interest and CJW has a board position. The other authors report no conflict of interest.

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Data availability statement

All data needed to support the conclusions of this manuscript are included in the main text or Supplementary Materials. Derived data supporting the findings of this study are available from the corresponding author C.J.W on request.

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