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Influence of varied assistance levels provided by a dual-joint active back-support exoskeleton on spinal musculoskeletal loading and kinematics during lifting

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ABSTRACT

An active dual-joint back-support exoskeleton with motors at both lumbar and hip level was designed to reduce spinal musculoskeletal loading and preserve lumbar flexibility during lifting. A subject-specific controller estimated the moment actively generated by back muscles to counteract gravitational forces on the upper body, minimising a counter-productive abdominal muscle contraction. Eight subjects lifted a 15 kg load using free technique with four assistance levels, i.e. 0%, 30%, 50%, and 70% of the active moment. Time-averaged L5S1 compressive force and back muscle active moment estimated by an EMG-driven biomechanical model, decreased by 5.5–9.3% and 14.9–28.6%, respectively, with non-zero assistance. Higher assistance did not yield larger L5S1 compression reduction but did gain further reduction in the time-averaged back muscles active moment. No significant changes in abdominal muscle activity and minor changes in lumbar flexion were observed suggesting the controller and dual-joint design achieved their objectives.

Practitioner Summary: Spinal load is a risk factor for low-back pain. An active dual-joint back-support exoskeleton, with high-torque-capacity motors and a subject-specific controller, was investigated in lifting tasks. Higher assistance level did not achieve further spinal compression reduction. The dual-joint design preserved lumbar flexibility, and the controller avoided redundant exoskeleton assistance.

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KEYWORDS
Low-back pain; spine loading; active exoskeletons; lifting

1. Introduction

Low-back pain (LBP) is one of the most prevalent health problems, causing a large worldwide healthcare burden (Hoy et al. 2012; Wu et al. 2020). Sixty percent of patients who report in primary care with LBP will still have pain one year after onset or experience at least one recurrence within a year (Itz et al. 2013; Knezevic et al. 2021). Spinal loading is an important risk factor contributing to LBP (Bakker et al. 2009; Coenen et al. 2013; Griffith et al. 2012; Magnusson et al. 1996).

While applying some ergonomic interventions in working environments, such as using a lifting device or lifting an object from hip instead of floor height, may have large effects on spine compression, implementing these interventions is not always feasible or results in excessive time loss (Koopman et al. 2019b). Changes in lifting style are another option to reduce spinal compression force during lifting to some extent, but the reductions are insufficient (Kingma, Faber, and van Dieën 2010, 2016; Kingma et al. 2004). Therefore, back-support exoskeletons have been suggested as a more versatile solution to decrease the risk of LBP by providing an extension moment to support the upper-body mass and external loads during lifting (Crea et al. 2021; de Looze et al. 2016; Kermavnar et al. 2021).

Back-support exoskeletons are categorised as passive and active (Crea et al. 2021; de Looze et al. 2016). In passive exoskeletons, the extension moment is generated by spring deformation induced by a change in joint angle. Active exoskeletons have motors that can

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generate support based on the user's intent and the interaction with the environment through a control model, thereby potentially providing more versatility in the occupational environment (Koopman et al. 2019b; Poliero et al. 2020; Toxiri et al. 2018). Previous studies reported that using passive and active exoskeletons during lifting reduced the back muscles' electromyography (EMG) amplitude (Crea et al. 2021; de Looze et al. 2016; Kermavnar et al. 2021; Koopman et al. 2019a). However, a reduction in EMG amplitude may be caused by an increase in lumbar flexion (Kingma et al. 2022; Koopman et al. 2020; Madinei and Nussbaum 2023), due to a load shift from back active contractile to passive tissues, indicating that a decrease in back muscle EMG amplitude does not necessarily coincide with a reduction in muscle and spinal compression forces.

In lifting, the total moment required comprises the moment generated by active back muscles (Mactive), by passive tissues in the back (Mpassive), such as muscle parallel elastic issues, ligaments, and fascia, and by any assistive device if used, such as an exoskeleton (Crea et al. 2021; de Looze et al. 2016, 1993; Tabasi et al. 2020). Therefore, both Mactive and Mpassive, or associated compressive forces, must be estimated, to investigate the effect of exoskeleton support on spinal loads (Kingma et al. 2022; Koopman et al. 2019a, 2020; Madinei and Nussbaum 2023). Mpassive is only related to lumbar flexion (Holleran et al. 1995; Toussaint et al. 1995). Consequently, it cannot be reduced by the support of exoskeletons (Tabasi et al. 2020, 2022a). If the support exceeds the Mactive that would be needed for the same lift without an active exoskeleton, antagonistic (i.e. abdominal) muscle forces will be required to counteract the redundant exoskeleton support (Tabasi et al. 2022b) resulting in a higher abdominal EMG amplitude and thus a higher compression forces. A control model that distinguishes between Mactive and Mpassive could prevent this from happening.

Although two previous studies reported a reduction in peak compression force at the lumbosacral joint (L5S1) while wearing an active exoskeleton during lifting with a 15 kg object (Koopman et al. 2019b; Lazzaroni et al. 2019), the peak compression force still exceeded 4000 N, which was suggested to be the threshold of low-back overload risk in males over 40 years and females over 20 years old (Jäger 2018). Additionally, the exoskeleton's rigid connection between the pelvis and the trunk (Koopman et al. 2019b; Lazzaroni et al. 2019) caused decreased lumbar flexion, and this is perceived as a hindrance (Näf et al. 2018).

To establish a further reduction in spinal compression force and to prevent restricting lumbar kinematics and redundant abdominal muscles contraction, a novel active exoskeleton (EXO) has been developed. The characteristics of the EXO include (1) two pairs of bilateral actuated joints approximately aligned with the hip and L3 joint centre in the sagittal plane to allow separate lumbar and hip support and mobility, (2) motors with a large torque-generation capacity (100 Nm per motor, resulting in 200 Nm for both the hip and lumbar joints), (3) an EXO control model designed to generate support proportional to Mactive caused by gravity acting on the upper body (Mactive_ub). Note that, currently, the control does not account for the active moment caused by the load lifted.

This study aimed to investigate the effect of different levels of EXO support on L5S1 compression forces, lumbar kinematics, trunk muscle EMG, and the loads on the back active contractile and passive tissues during lifting a 15 kg load using a free technique. The levels of support included 30% (LOW), 50% (HIGH), and 70% (EXTREME) Mactive_ub. A minimum impedance mode (MINIMP), which commanded the EXO to track zero torque (i.e. offset inertial and gravitational effects) was regarded as the reference condition. Our first hypothesis was that a higher level of support would lead to a higher reduction in the peak and time-averaged compression force at L5S1. Additionally, we hypothesised that, the Mactive-based control would avoid counter-productive abdominal muscle contraction during the whole lifting procedure, and the dual-joint design of the EXO would allow for a larger range of motion and prevent the kinematics restriction on lumbar kinematics.

2. Methods

2.1. Exoskeleton

2.1.1. Structure

The EXO (designed and developed by the University of Twente and Delft University of Technology; mass: approx. 16.5 kg) contains four motors (Bacchus V3, TU Delft: mass: 1.5 kg; max torque: 100Nm; max velocity: 60 rpm). Two of the motors are installed at the level of the right/left hip joint and another two are installed approximately at the level of L3. Their output torque and joint angle are internally measured using torque sensors and encoders at 1000 Hz, respectively. The EXO contains an adjustable thorax vest (Laevo FLEX, Laevo, the Netherlands), an adjustable pelvis brace, and two bilateral thigh wraps (Figure 1).

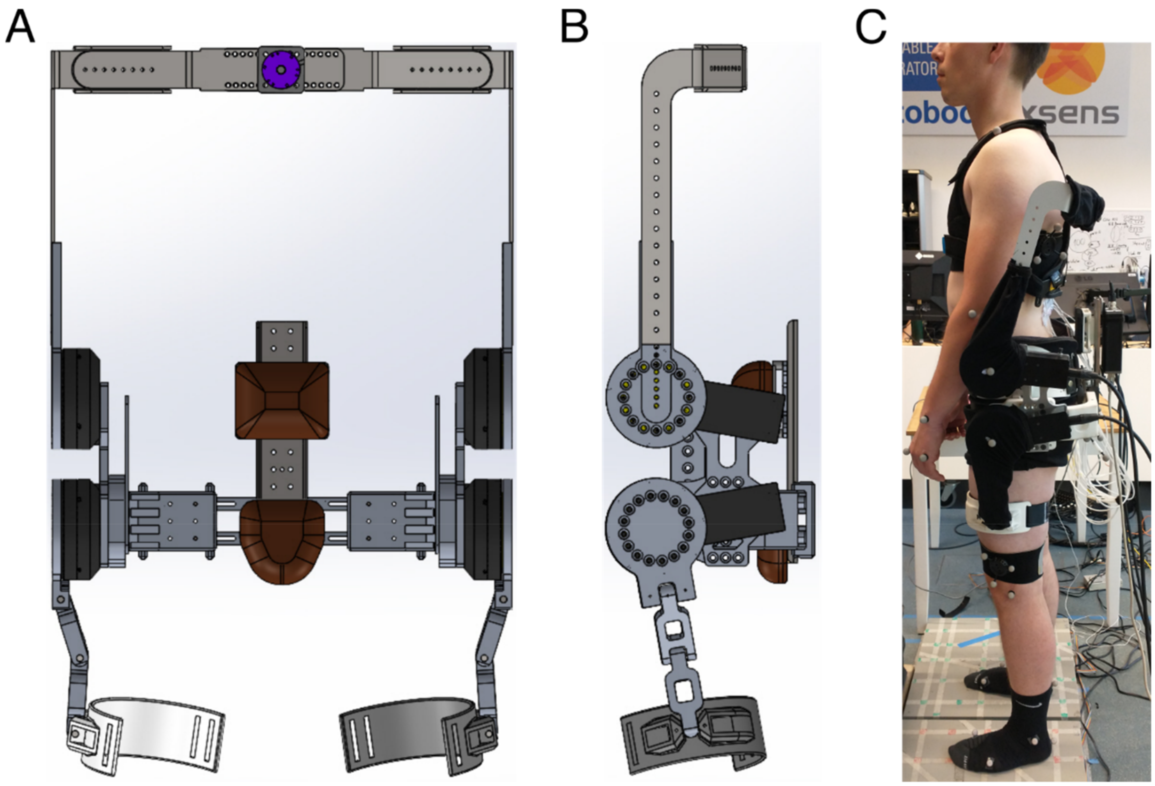

Figure 1. (A) Frontal and (B) side view of exoskeleton structure. The thorax vest, pelvis brace and padding, and thigh straps and padding are not displayed here. (C) Side view of a participant wearing the exoskeleton.

2.1.2. Control model

The support from the hip and lumbar motors was based on a percentage, depending on the support level condition (0% (MINIMP), 30% (LOW), 50% (HIGH), 70% (EXTREME)) of a real-time estimation of Mactive_ub around L5S1, estimated by subtracting Mpassive from the net moment at L5S1 due to the upper body mass (Mnet_ub). Consequently, the moment caused by an external load, such as a box lifted, was not considered. The lumbar-to-hip support ratio of the EXO was set to a typical lumbar-to-hip net moment ratio during lifting, i.e. 0.8 (Brouwer et al. 2024; Toussaint et al. 1992).

The net moment at L5S1 due to the upper body mass (Mnet_ub) was estimated using the participant's body mass, trunk length (mean posterior superior iliac spine process (MPSIS) to the C7 spinous process), and trunk inclination angle in the sagittal plane, measured by an inertial measurement unit (IMU) attached to the thorax at the position of 27.5% from MPSIS to C7, approximately at the height of the T12 spinous process (Faber et al., 2013).

To estimate Mpassive, the lumbar flexion angle was obtained by the orientation of the trunk IMU relative to an IMU attached over the sacrum. A lumbar flexion range-of-motion trial was used to calibrate an Mpassive-lumbar flexion angle relationship. The lumbar flexion angle signifying the onset of Mpassive was set to 20 degrees (Brouwer et al. 2024; van Dieën and Kingma 2005). We assumed that the net moment at L5S1 due to the upper body mass (Mnet_ub) at full flexion during the range-of-motion trial was fully generated by Mpassive. Afterwards, a fourth order polynomial was fitted to obtain the Mpassive-lumbar flexion angle relationship, and the Mactive_ub can be calculated by Equation (1).

$$Mactive\_ub = Mnet\_ub - Mpassive \tag{1}$$

The command torque from the EXO's controller was described in Equation (2). Since the EXO's lumbar and hip motors were placed in series and the EXO's support ratio was set to the typical lifting lumbar-to-hip support ratio (i.e. 0.8), the supportive moment provided by lumbar motors was shown in Equation (3).

$$Command\,torque = Mactive\_ub * Level \tag{2}$$

$$\begin{aligned} &Lumbar\,support \\ &\quad = Command\,torque * (Lumbar\ to\ hip\ support\ ratio) \end{aligned} \tag{3}$$

2.2. Participants

Eight male participants with no history of low-back pain participated in this study (mean±SD age: 27±3 years; height: 180±5 cm; weight: 73.4±6.4 kg). The study was approved by the ethical committee of the University of Twente (reference number: 230181). The participants were all male due to the tight fixation and potentially large forces at the thorax. The participants signed informed consent before starting the experiment.

2.3. Preparation procedure

2.3.1. Anthropometry, EXO fitting, EMG preparation

Anthropometric data of each participant was measured, comprising circumference of body segments, body height and mass, and trunk length. Based on the measured anthropometry, the width of the EXO pelvic brace, the size of the EXO thigh braces, and the length of the support arms attached between the lumbar motors and trunk vest were adjusted to fit the EXO to the user. Bipolar EMG electrodes were bilaterally attached over the longissimus thoracis pars lumborum, longissimus thoracis pars thoracis, iliocostalis lumborum, rectus abdominis, external oblique, and internal oblique muscles (Figure 2) (Kingma, Faber, and van Dieën 2010). The participant performed symmetric and asymmetric maximum isometric contractions of back and abdominal muscles to obtain the maximum voluntary contraction for each muscle (McGill, 1991).

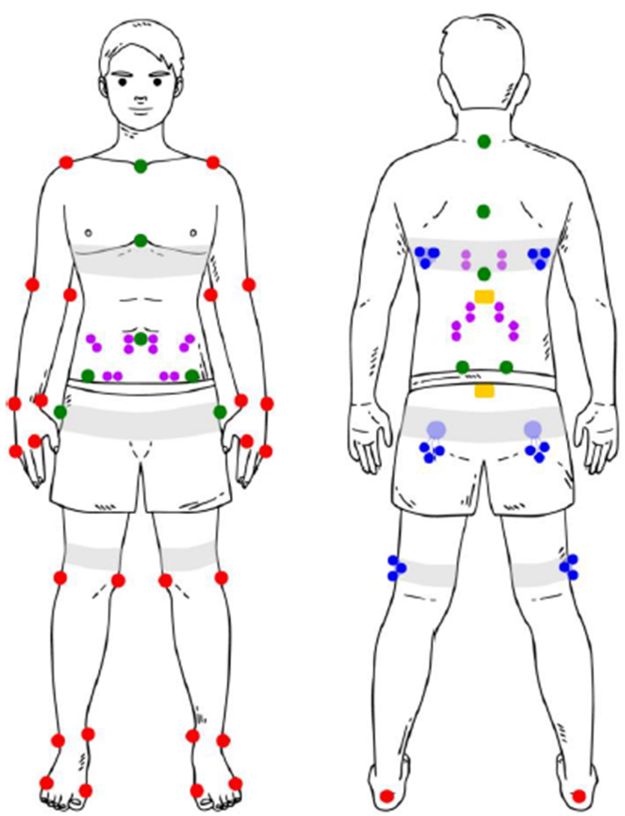

Figure 2. Schematic overview of IMU (yellow), EMG (purple) and reflective marker placement (red, blue, green). In Red: single markers recorded during all trials; Blue: cluster markers recorded during all trials; Green: single markers removed for exoskeleton trials (due to occlusion/interference with exoskeleton) and re-calculated during post-processing based on cluster markers or pelvis IMU data; Grey: elastic bands (Brouwer et al. 2024).

2.3.2. IMU calibration

A static anatomical neutral reference posture trial, a 6 m straight walking trial, and a lumbar flexion range-of-motion trial (passively hanging down with extended knees while standing) were recorded to define the neutral orientation for both IMUs, align each of the IMUs with the respective segment (Rispens et al., 2014), and calibrate the EXO control model (see section 2.1), all while wearing the EXO set to MINIMP mode.

2.3.3. Motion capture system setting

Six reflective cluster markers were attached to the thorax (two, bilateral below the scapulae), pelvis (two, bilateral at the height of the sacrum), and left and right thighs using elastic bands. Forty single reflective markers were attached to the pelvis, abdomen, thorax, and left and right feet, shanks, thighs, upper arms, forearms, and hands to measure 3D full-body kinematics (Figure 2). After recording the static anatomical neutral reference posture, the single markers placed on the trunk and pelvis and on both sides of the greater trochanter (Figure 2) were removed, since wearing the EXO would cause marker occlusion. The removed trunk and greater trochanter markers' trajectory were re-calculated based on the trajectories of the cluster markers using a transformation procedure (Cappozzo et al. 1995). The removed pelvis markers' trajectories were re-calculated based on the pelvis IMU data, because the pelvis cluster markers were contaminated by soft tissue artefacts related to glutaei muscles. Additionally, four single markers were attached laterally to the centres of the exoskeleton motors to obtain the 3D exoskeleton joint centre position.

2.4. Support level conditions

The participants performed five lifting trials without EXO and 27 with EXO. The without EXO (NOEXO) trials, including free technique lifts (0, 7.5, 15 kg) as well as stoop and squat technique lifts (15 kg), were only used in fitting an EMG-driven muscle model for each participant (model training trials; see section 2.6.1). Four of the 27 with EXO trials, i.e. free technique 15 kg box lifting tasks with the MINIMP, LOW, HIGH, and EXTREME support were selected to be analysed in this study. The trials of free technique in MINIMP and HIGH support with an empty box were used as familiarisation trials, which always preceded the investigation trials, but they were not used in analysis.

In the investigated trials, a 15 kg box was placed in front of the force plates and at the same height of force plates. The handles of the box were roughly at the same height as the middle of the shank. The participant lifted and lowered the box (releasing it after lowering) three times within a single trial. The lifting pace was controlled to be between 25 and 30 seconds for each complete trial. Within the investigated trials, the levels of support were randomly ordered.

2.5. Instrumentation

Full-body, EXO joint centres, and box kinematics were collected at 100 Hz by a twelve-camera motion capture system (Qualisys Medical AB, Sweden). Ground reaction forces and moments were recorded at 1600 Hz using two force plates (AMTI, USA). Surface EMG was recorded at 2048 Hz (Porti, TMSi, The Netherlands) and IMU (Xsens DOT, Movella, The Netherlands) orientation data were streamed at 60 Hz in real-time via a Bluetooth bridge to the EXO computer.

2.6. Data analysis

Kinematic and kinetic data were low-pass filtered using a bi-directional second-order Butterworth filter with a 5 Hz cut-off frequency. The angle of all joints was obtained by Euler decomposition of the distal segment relative to the proximal segment's anatomical axes. The order of the decomposition was Y-X-Z, i.e. flexion-extension, axial rotation, and lateral-flexion. The net reaction force at the L5S1 joint (Fnet) and the net moment at the L5S1 joint (Mnet) were estimated using a dynamic 3D linked segment model with bottom-up inverse dynamics (Kingma et al. 1996). The EXO mass was modelled at the pelvis centre of mass in the lifting trials with the EXO, since the bulk of the EXO mass was transferred to the user's pelvis segment. In trials with EXO, the flexion-extension moment around L5S1 generated by the participant (Mhuman) was calculated by subtracting the EXO lumbar supportive moment (Mexo) from Mnet.

The EMG data was band-pass filtered using a bi-directional second-order band-pass Butterworth filter using cut-off frequencies at 30–400Hz to remove motion artefacts and electrocardiographic (ECG) contamination (Redfern et al. 1993). Then, the filtered data was full-wave rectified, normalised to MVC, and low-pass filtered using a one-way 2.5 Hz second-order Butterworth filter to obtain the linear envelope while compensating for the electromechanical delay (Potvin et al. 1996).

2.6.1. EMG-driven muscle model

An EMG-driven trunk muscle model, including 164 muscle slips crossing L5S1, was used to compute the compression force and active, passive, and total muscle moments at the L5S1 joint (van Dieën and Kingma 2005; van Dieën 1997). The EMG model includes EMG-muscle force relationships and musculoskeletal anatomy, as well as active force-length, force-velocity, and passive force-length relationships. The model was driven by abdominal and back muscles' EMG linear envelopes and lumbar kinematics. The model was personalised for each participant by a calibration procedure, using anthropometric data and seven parameters defining the individual muscle contractile properties, including (1) a gain factor for the EMG-force relationship, (2,3) a scaling factor and an offset factor of the active force-length relationship, (4,5) a scaling factor and an offset factor for the passive force-length relationship, and (6,7) two scaling factors for eccentric and concentric contractions in the active force-velocity relationship, respectively (van Dieën and Kingma 2005). An optimisation function minimising the difference between Mnet and the total muscle moment (Mmuscular) estimated by the EMG-driven model was used for determining these seven parameters. All NOEXO trials were used for the EMG-driven model calibration. The RMSE between Mnet and Mmuscular was calculated over all the calibration trials (NOEXO trials) to evaluate the model fit. The R-squared and RMSE between Mhuman and Mmuscular were calculated over all with EXO trials, to evaluate the performance of the model.

2.6.2. Selected biomechanical outcomes

Since peak compression of the spine during lifting was indicated as an important risk factor for developing LBP (Coenen et al. 2013, 2014; da Costa and Vieira 2010), the peak compression force was determined using the EMG-driven muscle model. Other outcome parameters were calculated at the instant of the peak compression. Each trial included three lifting cycles considering the magnitude of the box plus exoskeleton weight and their potential effect on muscle fatigue. Each cycle contained a descending phase without the box, ascending phase with the box, descending phase with the box, and ascending phase without the box. The peak compression force was identified for the first half of each cycle (Figure S3). Consequently, three instants of peak compression force were identified for each trial. All the outcome parameters were averaged over these three instants within trials, including peak compression force (Fcompression) and other outcome parameters at peak Fcompression, such as Mactive, Mpassive, Mmuscular, mean back muscle EMG amplitude (backEMG), mean abdominal muscle EMG amplitude (abdominalEMG), lumbar flexion angle, Mexo, Mnet, and Mhuman (Mnet minus Mexo).

Since cumulative exposure to low-back loading has also been identified as an important LBP risk factor (Coenen et al. 2013), time-averaged Fcompression, Mactive, Mpassive, Mmuscular, backEMG, abdominalEMG, lumbar flexion angle, Mexo, Mnet, and Mhuman across each lifting trial (3 lifting and lowering cycles, Figure S3) were also calculated.

2.7. Statistics

A one-way repeated ANOVA was used to evaluate the influence of the levels of EXO support (MINIMP, LOW, HIGH, EXTREME) on all outcomes at the instant of peak Fcompression and all time-averaged outcomes, excluding Mexo. In case of a significant main effect ($p < 0.05$), Tukey post-hoc tests were performed.

3. Results

The RMSE between Mnet, calculated by the inverse dynamics, and Mmuscular, calculated by the EMG-driven muscle model, ranged from 9.0 to 19.4 Nm over participants for the calibration trials (NOEXO trials). Across all trials, the RMSE between Mhuman and Mmuscular ranged from 18.8 to 26.3Nm, while R-squared ranged from 0.75 to 0.93.

Although no significant effect of support level on the peak Fcompression was found (Table 1), a significant effect of support level on the Mmuscular, Mactive, lumbar flexion angle, and Mhuman at the instant of peak Fcompression was found (Table 1). Also, the support level had no significant effect on abdominalEMG at the instant of peak Fcompression (Table 1). Post-hoc tests only showed significant differences in Mhuman at the peak Fcompression instant between MINIMP and all non-zero support levels (Figure 3).

Most of the time-averaged outcomes were significantly influenced by support levels (Table 1). However, the time-averaged abdominalEMG was not significantly influenced by support levels (Table 1). While the pattern of results over conditions was very similar between outcomes at peak Fcompression (Figure 3) and time-averaged outcomes (Figure 4), the latter did, in contrast to outcomes at peak Fcompression, show more significant post-hoc differences. Specifically, compared to MINIMP, time-averaged Mactive significantly decreased 9.5% ($p = 0.002$), 19.0% ($p = 0.002$), and 28.6% ($p < 0.001$) in LOW, HIGH, and EXTREME, respectively (Figure 4B). Compared to LOW, time-averaged Mactive was reduced by 10.5% in HIGH ($p = 0.038$), and compared to HIGH, it was further reduced by 11.8% in EXTREME ($p = 0.001$) (Figure 4B). Compared to MINIMP, time-averaged backEMG reduced 12.0% ($p = 0.005$), 20.0% ($p < 0.001$), and 28.0% ($p < 0.001$) in LOW, HIGH, and EXTREME, respectively (Figure 4E). Compared to MINIMP, time-averaged Mnet increased 6.3% ($p = 0.010$) and 12.5% ($p < 0.001$) in LOW and EXTREME, respectively (Figure 4G). Compared to MINIMP, time-averaged Mhuman decreased 9.5% ($p = 0.002$), 20.0% ($p = 0.004$), and 23.2% ($p < 0.001$) in LOW, HIGH, and EXTREME, respectively (Figure 4I). Compared to LOW, the time-averaged Mhuman was reduced by 15.1% ($p = 0.002$) in EXTREME (Figure 4I).

4. Discussion

This study investigated the effect of different levels of EXO support on spinal loading, lumbar kinematics, and trunk muscle activity and loads. EXO control was subject-specific, proportional to the moment generated by the active contractile part of the back muscles to counteract gravitational forces on the upper body, to prevent counter-productive support, causing

Table 1. P-values, F-values, degree of freedoms, and effect sizes of one-way repeated ANOVA with levels of support (MINIMP, LOW, HIGH, EXTREME).

	Peak instant				Average			
	p	F	df	η2	*p*	F	df	η2
L5S1 compression force (Fcompression)	0.061	2.86	3	0.12	**0.013**	4.59	3	0.07
Total muscle moment (Mmuscular)	**0.031**	3.59	3	0.17	**0.007**	5.39	3	0.07
Active moment (Mactive)	**0.015**	4.36	3	0.17	**<0.001**	39.3	3	0.24
Passive moment (Mpassive)	0.057	2.94	3	0.06	0.052	3.03	3	0.11
Back muscle EMG (backEMG)	0.388	1.06	3	0.01	**<0.001**	40.7	3	0.09
Abdominal muscle EMG (abdominalEMG)	0.625	6.32	3	0.02	0.281	1.36	3	0.01
Lumbar flexion angle	**0.044**	3.2	3	0.02	**0.045**	3.19	3	0.06
Net L5S1 moment (Mnet)	0.322	1.24	3	0.07	**<0.001**	9.12	3	0.14
Net L5S1 moment minus Mexo (Mhuman)	**<0.001**	19.9	3	0.4	**<0.001**	29.9	3	0.34

Peak instant indicates the output parameters at the instant of peak compression, and Average indicates the time-averaged output parameters across the whole lifting trial. Significant *p*-values < 0.05 are presented in bold.

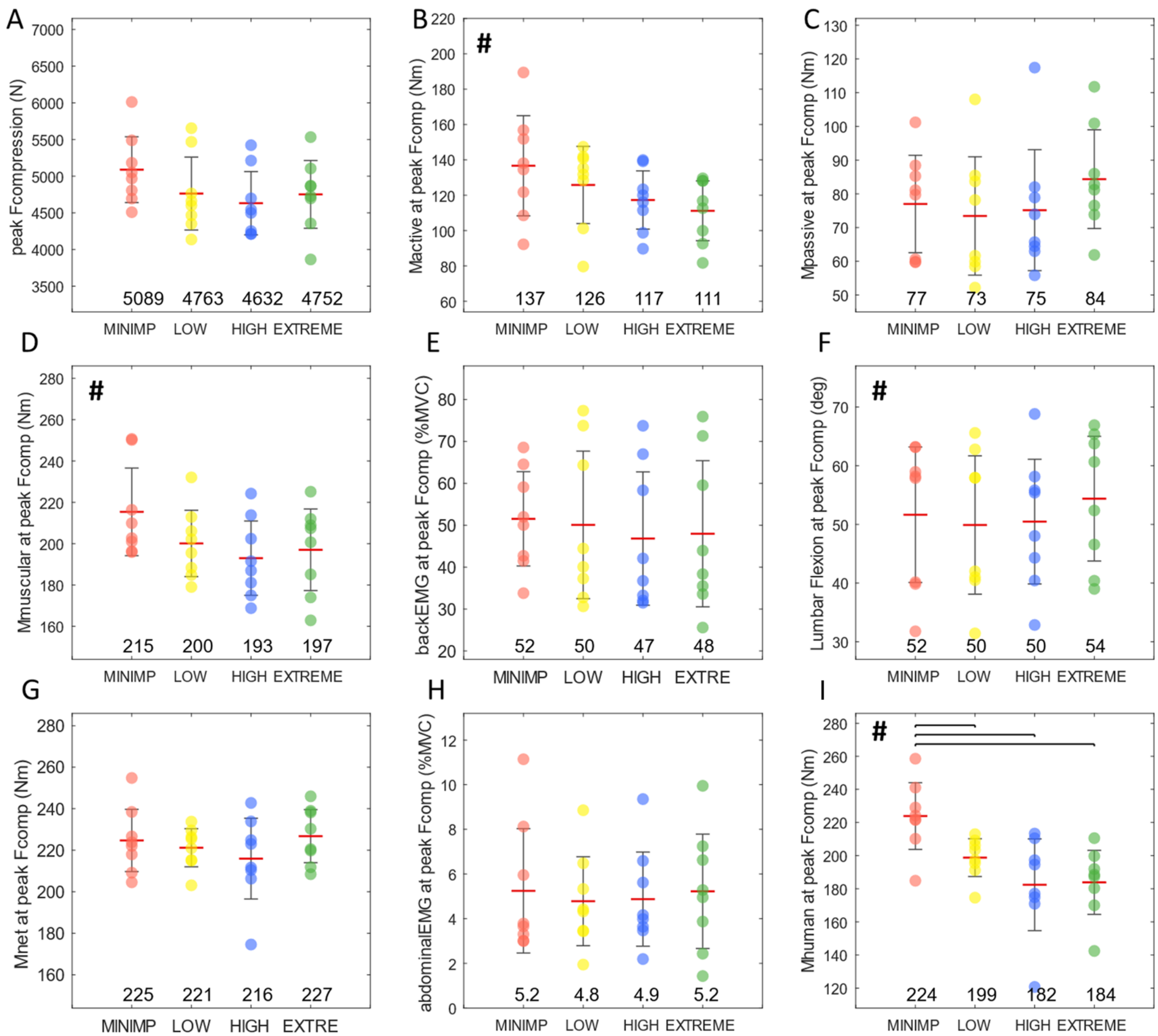

Figure 3. (A) Peak compression force (Peak Fcompression), (B) active muscle moment (Mactive), (C) passive muscle moment (Mpassive), (D) total muscle moment (Mmuscular), (E) back muscle EMG (backEMG), (F) lumbar flexion angle (Lumbar Flexion), (G) net moment (Mnet), (H) abdominal muscles EMG (abdominal EMG), and (I) moment generated by human body (Mhuman) at the instant of peak compression force (at peak Fcomp). The red horizontal lines and vertical black lines indicate the mean and standard deviation, respectively. Per condition, the average value is depicted at the bottom. A significant main effect of support level was indicated with #. The horizontal black brackets indicate a significant post-hoc difference ($p < 0.05$) between two conditions. Note that most of the y-axes of the plots do not start at zero.

abdominal muscle contraction, when reaching full lumbar flexion.

The effects of EXO support level on peak Fcompression, Mmuscular, Mactive, Mpassive, lumbar flexion angle, and Mhuman at the instant of peak Fcompression were either significant or approached significance, consistent with the time-averaged outcomes. However, for most of the peak outcomes and some of the time-averaged outcomes, which showed significant ANOVA results, we could not locate post-hoc between-condition differences, most likely due to a lack of statistical power. Thereby, a higher support level could not be linked to greater benefit in either peak Fcompression or time-averaged Fcompression, but it led to a greater reduction in Mhuman at the instant of peak Fcompression, time-averaged Mhuman, time-averaged Mactive, and time-averaged backEMG.

The support level did not show a significant influence ($p = 0.061$) on peak Fcompression, but it exhibited a significant influence ($p = 0.031$) on Mmuscular at the instant of peak Fcompression. The EXO support decreased the Mmuscular at the instant of peak Fcompression ranging from 8.0 to 19.0%, relative to MINIMP, without post-hoc tests' significance.

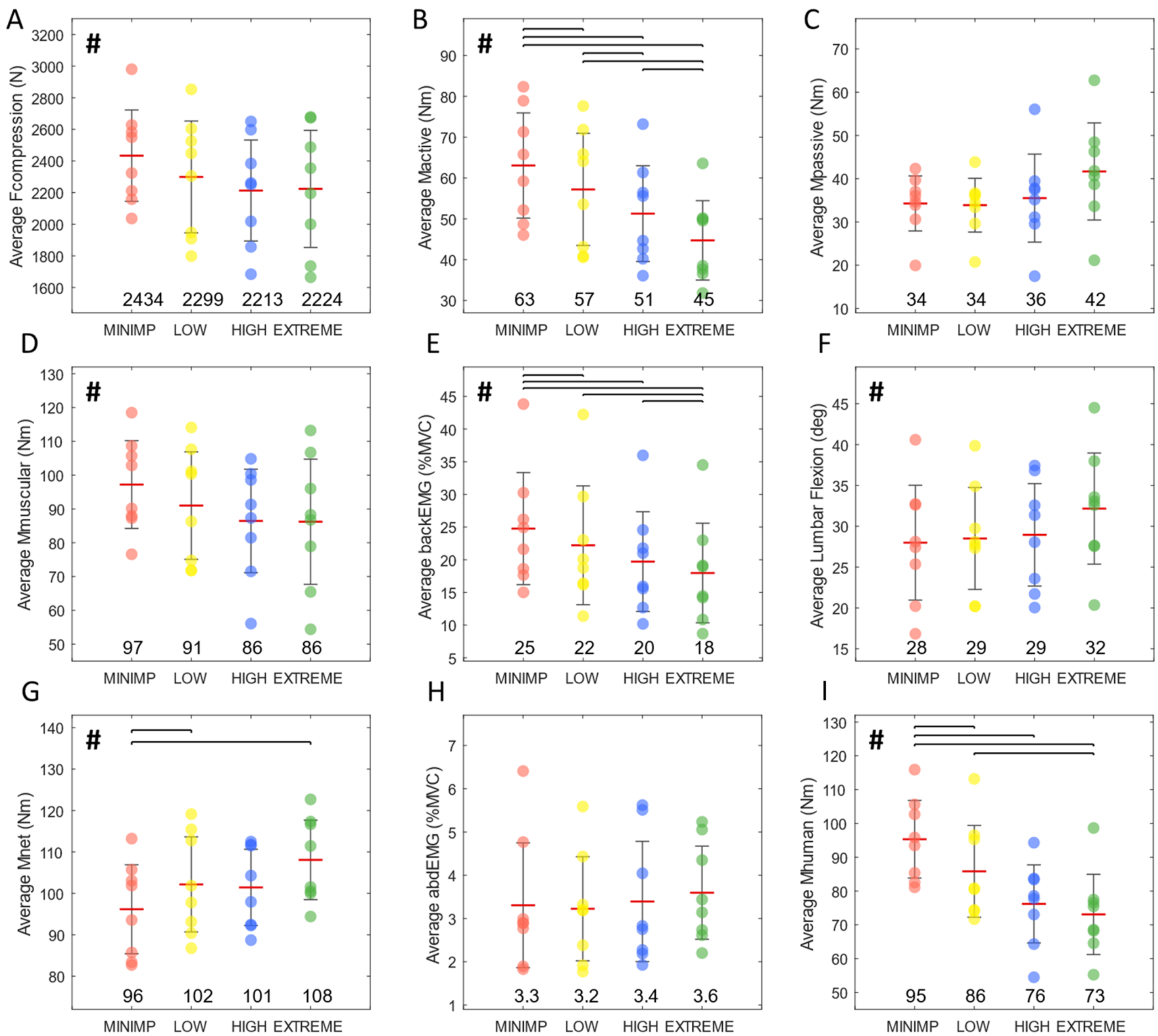

Figure 4. Time-averaged (A) compression force (Fcompression), (B) active muscle moment (Mactive), (C) passive muscle moment (Mpassive), (D) total muscle moment (Mmuscular), (E) back muscles EMG (backEMG), (F) lumbar flexion angle (Lumbar Flexion), (G) net moment (Mnet), (H) abdominal muscles EMG (abdominalEMG), and (I) moment generated by human (Mhuman). The red horizontal lines and vertical black lines indicate the mean and standard deviation, respectively. Per condition, the average value is depicted at the bottom. A significant effect of support level was indicated with #. The horizontal black brackets indicate a significant post-hoc difference ($p < 0.05$) between two conditions. Note that the y-axes of the plots do not start at zero.

Time-averaged Fcompression and Mmuscular were reduced by EXO support from 5.5 to 9.1% and from 6.2 to 11.3%, respectively, compared to MINIMP, but the post-hoc tests did not yield significant results for both parameters. This is likely mainly attributable to a lack of statistical power, as time-averaged Fcompression and Mmuscular in each non-zero support level were significantly different from MINIMP in paired T-tests without correction (time-averaged Fcompression: $p = 0.017 \sim 0.025$; time-averaged Mmuscular: $p = 0.015 \sim 0.024$). Furthermore, the increase of time-averaged Mnet with support indicated that lifting technique slightly changed with EXO support. Thereby, the benefit of EXO support in terms of time-averaged Fcompression and Mmuscular was counteracted by an increase of time-averaged Mnet, which may also partially explain the insignificant results in the post-hoc tests of average Fcompression and Mmuscular.

However, both Mhuman at the instant of peak Fcompression and time-averaged Mhuman significantly decreased from 11.2 to 18.8% ($p = 0.039$ to $p < 0.001$) and 10.4 to 22.9% ($p = 0.004$ to $p < 0.001$) by EXO support, respectively (Figures 3I and 4I). Mhuman is the required moment generated by human body during lifting, estimated by subtracting the moment measured in the EXO lumbar motors from Mnet. The slight difference between the results of Mhuman and Mmuscular could be due to several potential factors: an incomplete transfer of EXO support to the body, the inability of surface EMG electrodes to capture all the muscles around L5S1 resulting in bias in the EMG-driven muscle model, and the inaccurate detection of muscle excitation by bipolar EMG electrodes (Vieira and Botter 2021).

The EXO controller provided support proportional to Mactive_ub, resulting in a gradual decrease in the time-averaged Mactive as the support level increased (Figure 4B). The measured Mexo was close to the desired support moment estimated (and commanded to the motors) by the controller. For example, at the instant of peak Fcompression, Mnet was 222Nm in MINIMP (Figure 3G). For a lift with 15 kg loads, roughly 1/3 of the net moment (74Nm) is caused by the external 15 kg load (Kingma, Faber, and van Dieën 2016), and the moment caused by the external load was not supported by the EXO, because it was controlled based on Mactive_ub. The other 2/3 of the net moment (148Nm = 222-74) was distributed to Mpassive (77Nm, Figure 3C) and Mactive_ub (71Nm = 148-77). According to Equation (3), the desired support moment generated by the EXO lumbar motor should be 17.0, 28.4, and 39.8Nm in LOW, HIGH, and EXTREME, respectively, which was similar to the actual generated Mexo (Figure S1A). If the moment caused by the 15 kg box would have been taken into account by the control model, the Mactive at the instant of peak Fcompression would gain a further reduction, probably resulting in a further reduction in peak Fcompression.

The EXO support had a significant effect on lumbar flexion at the instant of peak Fcompression and on time-averaged lumbar flexion. Both were slightly increased in EXTREME compared to other support levels (Figures 3F and 4F), although post-hoc tests did not show significance. In line, and in accordance with literature (Holleran et al. 1995; Toussaint et al. 1995), Mpassive showed similar results (Figures 3C and 4C), even though ANOVA results were slightly above significance level. The reason for this slight increase is unclear. Observations in some participants with EXTREME support revealed that the support rapidly declined when they almost reached the box, probably due to the non-linear nature of the lumbar flexion-passive force relation. Slight deviations in actual passive forces, or in measurement of the lumbar flexion angle, could have caused a mismatch between actual passive moment increase and support decline, potentially resulting in an overshoot of the downward trunk motion. As elongation of and increasing stress on the passive tissues may cause damage to these tissues (Solomonow et al. 2003), EXTREME support might slightly increase the risk of injury. A previous study for a passive back-support exoskeleton also reported an increase in lumbar flexion while applying the maximum support, which was due to the restriction of pelvis inclination during 15 kg lifting (Arauz et al. 2024). However, this restriction was not found in the current study (Figure S1D and S2D), most likely because the current EXO has joints not only at hip but also at lumbar level, thereby providing an extra degree of freedom compared to other back-support exoskeletons.

Abdominal muscle EMG at the instant of peak Fcompression and time-averaged abdominal muscle EMG showed no difference (Figures 3H and 4H) across support levels. These results suggest that, in accordance with our second hypothesis, the control of the EXO adhered to the objectives of the subject-specific model, as proposed by Tabasi et al. (2020), which only assists active force production by back muscles and prevents extra abdominal muscle contraction needed to counteract redundant support while approaching full lumbar flexion.

Some other active back-support exoskeletons studies have considered the external load in their control model using forearm muscle EMG (Koopman et al. 2019b) or low-back muscle EMG (Hara and Sankai 2010). Koopman et al. (2019b) reported an 18% reduction in spinal compression force during 15 kg free lifting. However, their control did not take into account the distribution between Mactive and Mpassive, and their exoskeleton included a rigid hip-trunk structure with only one actuated hip joint. Therefore, it only had one degree of freedom and constrained the movement of lumbar joints resulting in a smaller lumbar flexion during lifting (Koopman et al. 2019b) and causing perceived hinderance (Näf et al. 2018). Furthermore, the reduction in spine compression by their exoskeleton was partially attributed to a reduction in lifting velocity (Koopman et al. 2019b), while there was no difference in lifting velocity while wearing the EXO in the current study (Figures S1B and S2B).

EXO support at levels found here may have limited effect on the risk of compressive damage to the spine (Jäger 2018). With a decrease of up to 9.3% in time-averaged Fcompression during a whole lifting trial, the support would provide a benefit in reducing cumulative spine compression load over a workday. Furthermore, the time-averaged backEMG substantially and significantly decreased, which may reduce the rate of development of muscle fatigue. Similar to other back-support exoskeleton studies (Alemi et al. 2019;

Bosch et al. 2016; Koopman et al. 2019b, 2020; Ulrey and Fathallah 2013a, 2013b), the EXO in the current study decreased the time-averaged backEMG by 12 to 28% (Figure 4e) compared to MINIMP. However, a reduction in backEMG does not necessarily imply a reduction in spinal loading because a change in lumbar flexion between conditions could imply that active back muscle loads can be transferred to passive back tissues. Consequently, we calculated and presented both Mactive and Mpassive implemented in this study. Our findings regarding the EXTREME condition suggest a shift from Mactive to Mpassive, which highlights the relevance of this separation.

The performance of the fitted EMG-driven muscle model was acceptable, considering the RMSE (18.8–26.3Nm) and R-squared (0.75–0.93) between Mhuman and Mmuscular, and comparable to previous studies (Koopman et al. 2019b, 2020; Tabasi et al. 2022a). The estimated peak Fcompression in current study was similar to previous assisted lifting studies (Johns et al. 2024; Kingma et al. 2022; Koopman et al. 2019b, 2020; Madinei and Nussbaum 2023).

A major limitation of this study is that the sample size limited statistical power. This may have resulted in the lack of significance in the outcomes at the instant of peak Fcompression and the time-averaged outcomes. Moreover, for significant ANOVA findings, the location of differences could only be identified with post-hoc tests for a few variables. We could not find solid evidence supporting our first hypothesis that more support leads to lower back loading. So, the number of repeated lifts, or the number of participants, would preferably have been larger in the present study. Note also that our cumulative load measure cannot be extrapolated to represent cumulative loading over a full work shift. A second important limitation was the fact that the external load was not taken into account by the control model, resulting in a much lower support than the EXO can generate, therefore the effect of the EXO could be larger if the external load was included. The EXO control model in current study did not consider the external load in view of requirements regarding equipment, calibration, and real-time calculation (Moya-Esteban et al. 2022, 2023; Tabasi et al. 2020, 2022a). Considering the torque-generation capacity of the actuators on the EXO, including the external load in the control model is possible and can probably lead to stronger EXO effects.

Another limitation is that during the control model calibration, we assumed that Mnet was fully generated by passive tissues in the fully flexed position. However, this assumption may not hold for participants who have a large range of motion in their hip joints and lumbar spine, and do not reach flexion relaxation (Laird, Keating, and Kent 2018). The familiarisation was limited, and the familiarisation trials did not include lifts with 15 kg external loads and other support levels including LOW and EXTREME. Even though some participants indicated that they found it easy to adapt to the HIGH support, a longer and more complete familiarisation procedure, including all support levels and lifting with the 15 kg box, could have been helpful for users to better utilise the EXO support, which might result in greater benefits from the EXO and improve the consistency of effects of using the exoskeleton across the users.

Finally, considering the motors' large capacity, a robust structure was needed for the current EXO prototype. Consequently, the EXO in its current state is too heavy for occupational application. Future work should also include female participants after improving thorax fixation, because the lifting kinematics are different between two genders (Lindbeck and Kjellberg 2001; Marras, Davis, and Jorgensen 2002, 2003; Plamondon et al. 2017).

5. Conclusion

We tested a back-support exoskeleton including actuated hip and lumbar joints and control providing support proportional to the lumbar moment generated actively by the back muscles to counteract the moment induced by gravity acting on the upper body. Exoskeleton support reduced the time-averaged lumbar compression force by 5.5–9.1% during lifting a 15 kg load using a free technique, suggesting that such a device can reduce the risk of low-back pain caused by repetitive lifting work. However, increasing the support level could not be linked to a further reduction in both peak and time-averaged lumbar compression forces, but yielded a reduction in time-averaged active moment generated by back muscles and time-averaged lumbar moment generated by the participant. The peak support provided by the exoskeleton was substantially lower than its actual capacity. Including the moment actively generated due to the external load in exoskeleton control would be needed to yield further spinal load reductions. Nonetheless, the present exoskeleton control model based on a subject-specific model avoided counter-productive abdominal muscle contraction, and the dual-joint design prevented a kinematic restriction of lumbar flexion.

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Author contributions

Conceptualisation: F.H., N.B., A.T., I.K., H.K., J.D. Methodology: F.H., N.B., A.T., I.K., W.D., J.D. Data Curation: F.H., N.B., A.T., I.K., M.I.M.R. Formal Analysis: F.H., N.B., I.K, W.D., J.D. Visualisation: F.H., N.B., I.K. Writing – Original Draft: F.H. Writing – Review and Editing: All authors. Funding Acquisition: H.K., I.K., J.D. Supervision: N.B., I.K, W.D., J.D.

Ethics statement

The authors assert that all procedures contributing to this work comply with the ethical standards of the relevant national and institutional committees on human experimentation and with the Helsinki Declaration of 1975, as revised in 2008. The experimental procedure was approved by the local ethical committee associated with the Department of Biomechanical Engineering, University of Twente (reference number: 230181).

Disclosure statement

No potential conflict of interest was reported by the author(s).

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Data availability statement

The data that support the findings of this study are available from the corresponding author upon reasonable request.

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