SPEXOR passive spinal exoskeleton decreases metabolic cost during symmetric repetitive lifting

Baltrusch, S J; van Dieën, J H; Koopman, A S; Näf, M B; Rodriguez-Guerrero, C; Babič, J; Houdijk, H

published in
European Journal of Applied Physiology
2020

DOI (link to publisher)
10.1007/s00421-019-04284-6

document version
Peer reviewed version

citation for published version (APA)
Baltrusch, S. J., van Dieën, J. H., Koopman, A. S., Näf, M. B., Rodriguez-Guerrero, C., Babič, J., & Houdijk, H. (2020). SPEXOR passive spinal exoskeleton decreases metabolic cost during symmetric repetitive lifting. *European Journal of Applied Physiology*, *120*(2), 401-412. https://doi.org/10.1007/s00421-019-04284-6

European Journal of Applied Physiology
https://doi.org/10.1007/s00421-019-04284-6

ORIGINAL ARTICLE

SPEXOR passive spinal exoskeleton decreases metabolic cost during symmetric repetitive lifting

S. J. Baltrusch[1,2] · J. H. van Dieën[2] · A. S. Koopman[2] · M. B. Näf[3] · C. Rodriguez-Guerrero[3] · J. Babič[4] · H. Houdijk[1,2]

Received: 14 August 2019 / Accepted: 4 December 2019

Abstract

Purpose Besides mechanical loading of the back, physiological strain is an important risk factor for low-back pain. Recently a passive exoskeleton (SPEXOR) has been developed to reduce loading on the low back. We aimed to assess the effect of this device on metabolic cost of repetitive lifting. To explain potential effects, we assessed kinematics, mechanical joint work, and back muscle activity.

Methods We recruited ten male employees, working in the luggage handling department of an airline company and having ample experience with lifting tasks at work. Metabolic cost, kinematics, mechanical joint work and muscle activity were measured during a 5-min repetitive lifting task. Participants had to lift and lower a box of 10 kg from ankle height with and without the exoskeleton.

Results Metabolic cost was significantly reduced by 18% when wearing the exoskeleton. Kinematics did not change significantly, while muscle activity decreased by up to 16%. The exoskeleton took over 18–25% of joint work at the hip and L5S1 joints. However, due to large variation in individual responses, we did not find a significant reduction of joint work around the individual joints.

Conclusion Wearing the SPEXOR exoskeleton decreased metabolic cost and might, therefore, reduce fatigue development and contribute to prevention of low-back pain during repetitive lifting tasks. Reduced metabolic cost can be explained by the exoskeleton substituting part of muscle work at the hip and L5S1 joints and consequently decreasing required back muscle activity.

Keywords Lifting device · Low-back pain · Energy expenditure · Mechanical work · Muscle activity · Movement behaviour

Communicated by Bénédicte Schepens.

Electronic supplementary material The online version of this article (https://doi.org/10.1007/s00421-019-04284-6) contains supplementary material, which is available to authorized users.

✉ S. J. Baltrusch
s.baltrusch@heliomare.nl

1 Department of Research and Development, Rehabilitation Center Heliomare, Wijk aan Zee, The Netherlands

2 Department of Human Movement Sciences, Faculty of Behavioural and Movement Sciences, Vrije Universiteit Amsterdam, Amsterdam Movement Sciences, Amsterdam, The Netherlands

3 Robotics and Multibody Mechanics Research Group, Department of Mechanical Engineering, Vrije Universiteit Brussel and Flanders Make, Brussels, Belgium

4 Laboratory for Neuromechanics and Biorobotics, Department of Automation, Biocybernetics and Robotics, Jožef Stefan Institute, Ljubljana, Slovenia

Abbreviations

BNDR	Bending non-demand return
CoM	Centre of mass
EMG	Electromyography
GRF	Ground reaction force
LBP	Low-back pain
MVC	Maximal voluntary contraction
PLAD	Personal lifting assistive device

Introduction

Low-back pain (LBP) has become a major health problem worldwide (Hoy et al. 2012). With a lifetime prevalence reported to be as high as 84% (Thiese et al. 2014), LBP causes a considerable burden on industry. Negative consequences for companies are, amongst others, decreased productivity (Bergstrom et al. 2014) and increased sickness absence from work (Roelen et al. 2010). The wide range of

Published online: 11 December 2019

complex exposure patterns at workplaces complicates the introduction of interventions, aiming to prevent LBP and support return-to-work management.

An important risk factor for the onset of LBP is mechanical loading of the low back, which can occur during manual materials handling and lifting (Coenen et al. 2014a, b). Despite technical innovations and the knowledge that exposure to occupational lifting should be reduced, physically demanding jobs remain. For example, one-third of employees in the Netherlands indicate that they regularly perform repetitive, physical tasks at work (Statistics Netherlands 2016), indicating that the skills and versatility of employees often cannot be replaced by automated processes.

Within the challenge of reducing mechanical loading of the low back in the work environment, body-worn assistive devices (exoskeletons) have been designed and introduced in industry (de Looze et al. 2016). Exoskeletons, as a mechanical intervention, are intended to reduce the mechanical load on the back by decreasing muscular activity in the back muscles, needed to counteract external moments caused by inertial and external forces. Previous research has shown that this concept of providing an assisting external extension moment can be effective in reducing lumbar L5S1 moments, back muscle activity, and compression forces in the low back (Abdoli et al. 2006; Abdoli-Eramaki and Stevenson 2008; Alemi et al. 2019; Bosch et al. 2016; Frost et al. 2009; Graham et al. 2009; Koopman et al. 2019; de Looze et al. 2016; Ulrey and Fathallah 2013b; Wehner et al. 2009).

Due to the reduction in low back moment and back muscle activity when wearing an exoskeleton, metabolic cost is also expected to decrease. Besides mechanical loading, metabolic cost should also be considered when it comes to preventing LBP. McGill et al. (2007) have shown that the metabolic cost of lifting objects may cause a ventilatory challenge that interferes with control of trunk movement. Additionally, this may cause fatigue, while fatigue-related changes in spinal stability may increase the risk of injury in the low back (Granata and Gottipati 2008). Keeping metabolic cost low, and thereby reducing fatigue, might, therefore, contribute to the prevention of LBP.

Two studies have previously assessed the effect of trunk exoskeletons on metabolic cost during lifting, with different outcomes. Whitfield et al. (2014) did not find differences in oxygen consumption when wearing the PLAD exoskeleton. In contrast, a more recent investigation by Baltrusch et al. (2019b) found decreased metabolic cost when wearing the Laevo exoskeleton. Both studies attributed their findings on metabolic cost to a change of lifting kinematics when wearing the exoskeleton. Specifically, the PLAD induced users to adopt a squat-like technique when lifting (Sadler et al. 2011), while the Laevo induced a stoop-like technique (Baltrusch et al. 2019b). It has been shown previously that the squat technique imposes higher metabolic cost than the stoop technique (Garg and Herrin 2007). Neither study investigated the reduction in joint work at the low back and hip that likely occurs due to the external support of the exoskeleton. Previous research has shown that a passive exoskeleton can take over up to 24% of lumbar moments (Koopman et al. 2019), indicating the potential of an exoskeleton to reduce joint work and by that reducing metabolic cost. Still, such a direct reduction in joint and muscle work has not been investigated yet.

Recently, a novel passive exoskeleton (SPEXOR) was designed to reduce peak and cumulative load on the low back (Näf et al. 2018). Initial measurements showed that support torques of 20–60 Nm can be provided by the exoskeleton, which could substantially reduce muscular effort (Näf et al. 2018). Further, pilot testing has revealed little hindrance by the exoskeleton when performing a set of working tasks, which might limit changes in lifting strategy as previously observed with PLAD and Laevo (Baltrusch et al. 2019b). The aim of this paper is to assess the effect of wearing the SPEXOR exoskeleton on metabolic cost. In addition, to explain potential differences in metabolic cost, we assessed the effect of the exoskeleton on kinematics, and muscle activity.

Methods

Passive exoskeleton

In this study, we tested the passive spinal exoskeleton SPEXOR that was designed and built to reduce peak and cumulative load on the low back (Näf et al. 2018). It consists of two passive elements, connected in series, with misalignment compensation mechanisms, that generate an extension torque around the hip and the L5S1 joint: (1) a pair of passive spring-based, hip joint actuators (MACEPPA 2.0) at the outside of the upper leg (Fig. 1a) and (2) an elastic spinal module at the level of L5S1 and up (Fig. 1b). The spring-based hip actuators were designed to be continuously adjustable. Each actuator can generate a support torque between 10 and 30 Nm, hence total peak torque can reach 60 Nm (both sides). The pre-tension of the springs in the hip actuators can be adjusted to change the provided support torque. The spinal structure comprises a ball joint, a linear slider and elastic carbon fibre beams. By adapting the thickness and number of beams, the stiffness of the spinal module can be adjusted. With three beams with a diameter of 4.7 mm, the current spinal model can generate peak torques of up to 50 Nm. Figure 2 presents the angle–torque relationship of both actuators, measured in a previous study (Näf et al. 2018). An additional feature of the exoskeleton is the implementation of a clutch (Fig. 1c), which can be switched on and off. If switched on, the clutch engages the springs in both

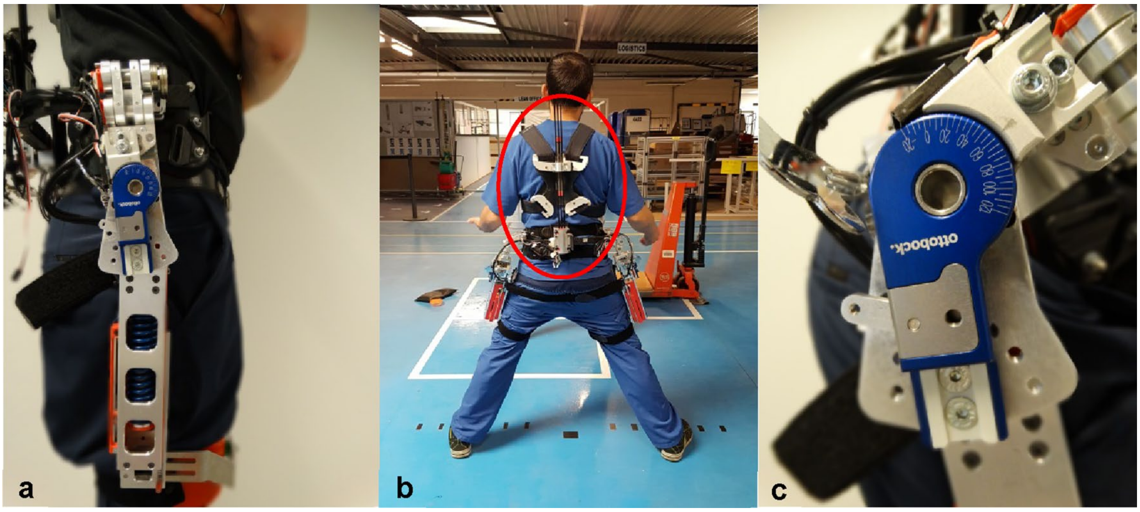


Fig. 1 The SPEXOR exoskeleton was designed to reduce peak and cumulative load on the low back by providing an extension moment around the low back with the help of two passive, in series-connected elements: two spring-based hip actuators (**a**) and an elastic spinal module (**b**, circled red). The implemented clutch allows disengagement of the passive hip actuators (**c**)

hip actuators. If switched off, the actuators stay unengaged and do not provide a support torque to the user, allowing for unobstructed hip flexion.

Participants

We recruited 11 male employees, working in the luggage handling department of the Dutch airline company KLM, who had ample experience with lifting tasks at work. The age, height and body mass of these participants were mean (SD) 47.4 years (7.1 years), 175 cm (7 cm), and 84 kg (15 kg). Data from one participant had to be discarded from the analysis, since this participant failed to complete the 5 min repetitive lifting task without the exoskeleton, due to pain in the back. Since our group of participants included people with and without a history of low-back pain, they were asked to indicate their current pain level before the start of the measurement on a scale from 0 (no pain) to 10 (maximum pain). The average pain level before the start of the measurement was (median [IQR]): 2.5 [0.75–5].

The participants received an information letter prior to the experiment and signed an informed consent form on the measurement day. All procedures performed in studies involving human participants were in accordance with the ethical standards of the institutional and/or national research committee (VUmc, Amsterdam, The Netherlands, NL57404.029.16) and with the 1964 Helsinki Declaration and its later amendments or comparable ethical standards.

Instrumentation

Metabolic cost

For assessing metabolic cost, a breathing gas analysis system was used (Cosmed srl, Quark CPET, Rome, Italy), measuring breathing volume, the rate of oxygen consumption and carbon dioxide production.

Kinematics and mechanical joint work

Ground reaction forces were measured using a single custom-made force plate (1.0 × 1.0 m) at a sample rate of 200 Hz. The force plate is regularly checked for linearity and has a centre of pressure (COP) error of < 4 mm. 3D Kinematics were recorded with an optoelectronic motion capture system (Optotrak, Northern Digital Inc., Waterloo ON, Canada) at a sample rate of 50 Hz. The measurement error of the Optotrak device is < 0.05 mm. Segment kinematics were quantified using a dynamic 3D-linked segment model (Kingma et al. 1996). Cluster markers were attached to lower and upper leg, pelvis, trunk (T10), upper and lower arm, head and box. Given the fact that participants performed symmetric lifting, we only recorded kinematics from the right side of the body. In addition, four cluster markers were attached to the exoskeleton (pelvis frame, hip actuator, beam base, beam top) to measure the 'hip' and 'lumbar' flexion of the exoskeleton. Prior to the measurements, cluster markers were related to anatomical landmarks using pointer measurements (Cappozzo et al. 1995).

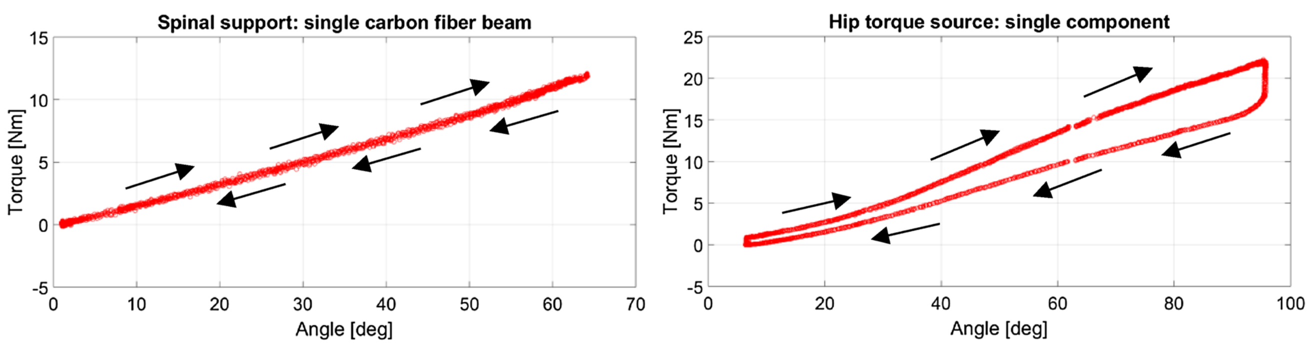


Fig. 2 Angle–torque relationship of the spinal module (**a**) and a single hip actuator (**b**) (adopted from Näf et al., 2018)

Muscle activity

Muscle activity was recorded at a sample rate of 2000 Hz, using surface Electromyography (Porti, TMSi, Oldenzaal, The Netherlands). Bipolar surface electrodes were placed at five bilateral sites on the skin after abrasion and cleaning with alcohol (Ag–AgCl electrodes; interelectrode distance, 20 mm). The recording sites were m. longissimus thoracis (LT) at the T9 level (4 cm lateral), m. iliocostalis lumborum (IL) at the L2 level (6 cm lateral), m. longissimus lumborum (LL) at the L3 level (3 cm lateral), m. external oblique muscles (EO) about 15 cm above the SIAS and m. rectus abdominis (RA), 3 cm lateral from the umbilicus.

Experimental procedure

At the start of the measurement, the oxygen mask was fitted to the participants and resting metabolic rate was measured, while participants were sitting in a chair for 5 min. Subsequently, the exoskeleton was fitted and adjusted to the participant. Anthropometric data were obtained and participants got familiarized with the exoskeleton. EMG electrodes were then placed on the participant, and maximal voluntary isometric contractions (MVCs) were performed. During MVC contractions, participants had to maximally activate the recorded muscles against resistance of upper body weight and added manual resistance for 5 s. The maximum values across three repetitions were later used to normalize EMG data of the subsequent trials. In preparation of the following protocol, cluster markers were attached to the participant's body, calibration measurements were performed.

During the actual experimental conditions, participants were instructed to lift and lower a box of 10 kg (0.39 × 0.37 × 0.11 m, with 2.5 cm diameter handles) from ankle height to hip height, at a rate of eight lifts per minute. The weight of 10 kg and the lifting rate were chosen to ensure that the task can be performed at a constant speed for at least 5 min, without having a break to ensure steady-state oxygen uptake, which is important for data interpretation (Jones and Poole 2005). Ankle height was chosen, since we expected the exoskeleton to generate the biggest support at that height, based on its angle–torque relationship (shown in Fig. 2). Also, at the worksite of our participants, luggage is often lifted from ankle height. Each lifting cycle started in an upright position and consisted of picking up the box, assuming an upright posture with the box, putting down the box, and assuming an upright posture without the box again. A metronome was used to impose lifting rate and participants were instructed to choose their own lifting technique. The instruction was: "Please pick up the box with every beat and put it down again at the next beat. Please adapt your speed to the beat of the metronome to have a smooth movement. You can choose the lifting technique you feel the most comfortable with." This protocol was performed twice, one trial with the exoskeleton (exoskeleton condition) and one trial without the exoskeleton (control condition). In the exoskeleton condition, the clutch was switched on to provide support. Also, great care was taken to avoid contact between the exoskeleton and the EMG electrodes to prevent from the potential confounding factor. The trial order was randomized and counterbalanced between participants. Breaks of at least 5 min were given between the trials. To reach steady state in the oxygen uptake, all trials lasted at least 5 min. Data on kinematics and EMG data were recorded over 5 min. The complete experimental set-up is presented in Fig. 3.

Data analysis

All data collected in the study were processed using MATLAB (R2015b, The MathWorks, Inc., Natick, Massachusetts, USA).

Metabolic cost

Metabolic cost (J/kg/s) was calculated from oxygen uptake and respiratory quotient (Garby and Astrup 1987). Flow

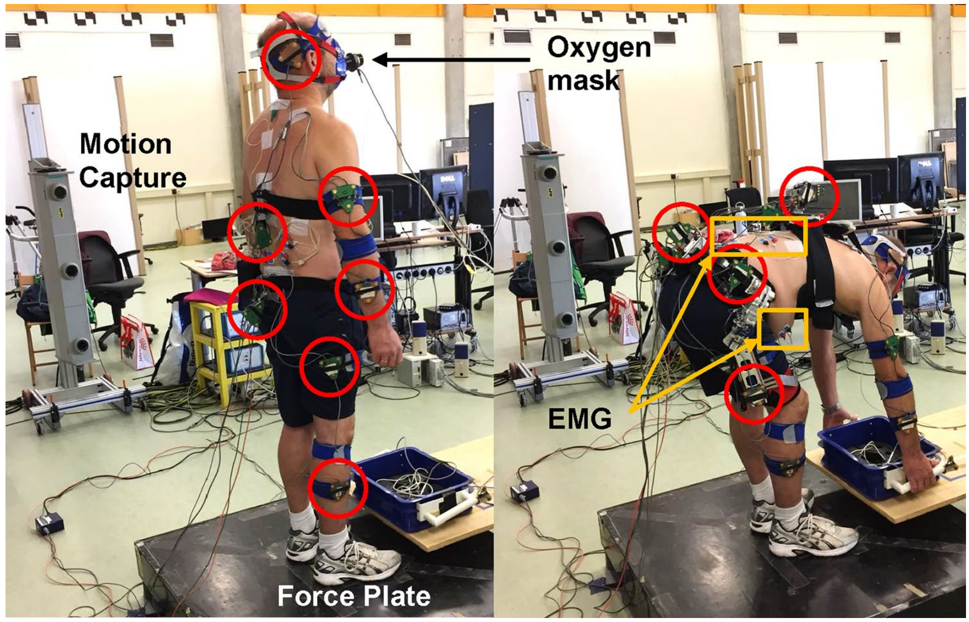


Fig. 3 Experimental setup without (left) and with exoskeleton (right). Red circles show the cluster markers on the body (left) and on the exoskeleton (right). The yellow boxes show the EMG location

rates were averaged over the final two minutes and normalized with respect to body mass. Net metabolic cost was found by subtracting resting metabolism from the metabolic cost during the lifting task.

Kinematics and mechanical joint work

Forces and kinematics were bi-directionally filtered with a second-order Butterworth filter and cut-off frequencies of 10 Hz and 5 Hz, respectively. To assess a potential change in movement behaviour between the control and exoskeleton conditions, we calculated joint angles and the range of motion of the centre of mass (CoM). Joint flexion–extension angles in the knee, hip, trunk and L5S1 were calculated by Euler decomposition of the orientation matrices of the segment local reference frames (Kingma et al. 2010). Since all participants adopted a squat or semi-squat lifting style, peak knee angles were used to define the start and end of a single lifting cycle. The body's centre of mass (CoM) was calculated based on the mass and CoM position of all segments, estimated according to Zatiorsky (2002). Range of motion of the CoM was obtained from the vertical distance travelled by the body centre of mass over a full lifting cycle. Flexion–extension net joint moments around knee, hip and L5S1 were calculated from ground reaction forces (GRF) and body kinematics with a bottom-up inverse dynamics model (Kingma et al. 1996). Angular velocities of the knee, hip and L5S1 joint were defined as the derivative of the joint angles. By multiplying the calculated net joint moments with the calculated angular velocities, we arrived at the joint power for each of the joints. To estimate positive and negative mechanical work generated around the joints, phases of positive and negative joint power were integrated separately over time.

To assess the mechanical work generated by the exoskeleton, we calculated the moment generated by the exoskeleton during lifting, using the measured exoskeleton angles and the known angle–torque relation of the exoskeleton (Näf et al. 2018). By multiplying the estimated support moments of the exoskeleton around the hip and the L5S1 with the joint angular velocities, mechanical power generated by the exoskeleton was calculated. After integrating negative and positive joint power episodes over time, we arrived at the negative and positive work generated by the exoskeleton (exowork) at the hip and the L5S1 joint. Finally, the mechanical work generated by the muscles (muswork) was found by subtracting the work generated by the exoskeleton (exowork) from the joint work (generated by muscles and exoskeleton).

Electromyography

EMG data were filtered using a fourth-order Butterworth band stop filter between 49 and 51 Hz to remove power line hum. Subsequently, the data were high-pass filtered (2nd order, 20 Hz), rectified and low-pass filtered (4th order, 2.5 Hz). Next, we normalized the EMG data to the maximal amplitude of the signal obtained in the MVC trials and to cycle time. The normalized data were averaged over both body sides and over cycles.

Statistics

To test for statistically significant differences between control condition and exoskeleton condition, we conducted paired t tests for all the outcome variables. Critical level of significance was set to $\alpha = 0.05$. After a first inspection, two participants showed errors caused by movement of the markers or loosening of the EMG electrodes. We, therefore, did not include these trials in the analysis of the kinetic, kinematic and EMG data. The number of participants included in the statistical analyses is, therefore, different for different dependent variables and are reported for each outcome.

Results

Metabolic cost

Wearing the exoskeleton decreased net metabolic cost of lifting by 18% [means (SD): 5.63 W/kg (1.26) vs. 4.64 W/kg (1.38); $p = 0.000$] (Fig. 4). One participant showed an increase in metabolic cost when wearing the exoskeleton. This could be explained by the weight of this participant (120 kg), and the resulting lower relative effect of the exoskeleton, compared to the remaining participants.

Kinematics and mechanical joint work

We did not find a significant effect of wearing the exoskeleton on peak angles in knee flexion, hip flexion, lumbar flexion and trunk inclination (Fig. 5). All participants adopted a squat or semi-squat lifting style in the control and exoskeleton conditions.

The average range of motion of the centre of mass (CoM) did not show a significant difference between control condition and exoskeleton condition, either (Fig. 6). Figure 6a, b shows the lowest and highest position of the CoM, averaged over participants. There was no significant difference in these positions, i.e. depth of squatting and full extension, between control and exoskeleton condition.

Positive and negative joint work at the knee during picking up and putting down the box did not change when wearing the exoskeleton (Fig. 7a). Total positive and negative joint work at the hip and L5S1 (generated by muscles and exoskeleton) also did not show significant differences between control and exoskeleton conditions (Fig. 7b, c).

Average positive and negative work generated by the exoskeleton (exowork) at the hip joint amounted to 13 joules (3.4) and − 15 joules (− 3.3), respectively. Despite the significant amount of work done by the exoskeleton, we did not find a significant difference in positive and negative muscle work (muswork) when wearing the exoskeleton, compared to the control condition for unloaded and loaded phases (Fig. 7b).

At the L5S1 joint, average positive and negative work generated by the exoskeleton (exowork) was 24 joules (4.5) and − 24 joules, (− 4.5), respectively. This resulted in a significant decrease in muscle-generated negative and positive work (muswork) in the exoskeleton condition in the unloaded phase (− 51.47 joules (31.84) vs. − 99.45 joules (55.81); $p = 0.02$ and 48.51 joules (29.29) vs. 88.42 joules (29.29); $p = 0.02$; Fig. 7c). Muscle work in the loaded phases did not show significant differences.

Muscle activity

The mean muscle activity in the back muscles, averaged over one lifting cycle, significantly decreased when wearing the exoskeleton, compared to the control condition, for m. longissimus thoracis [19.37% (7.07) vs. 17.39% (5.48); $p = 0.03$), m. iliocostalis lumborum (22.69% (11.28) vs. 18.96% (8.62); $p = 0.01$] and m. longissimus lumborum (26.22% (9.00) vs. 22.55% (5.91); $p = 0.04$) (Fig. 8). The muscle activity in the abdominal muscles did not change when wearing the exoskeleton (Fig. 9).

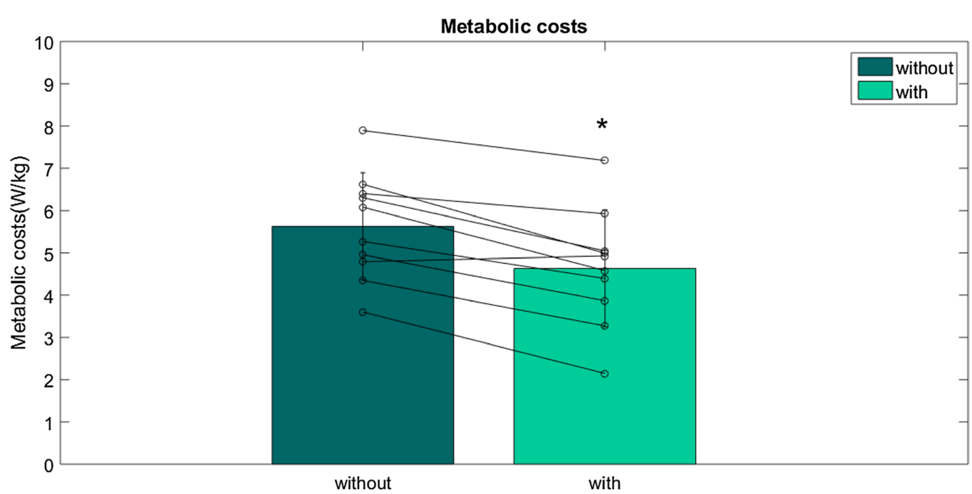


Fig. 4 Metabolic cost of lifting with and without exoskeleton. Values are normalized for bodyweight. $N = 10$. Error bars indicate standard deviations. Black lines indicate individual responses. *Significant change in metabolic cost between control condition (without) and exoskeleton condition (with)

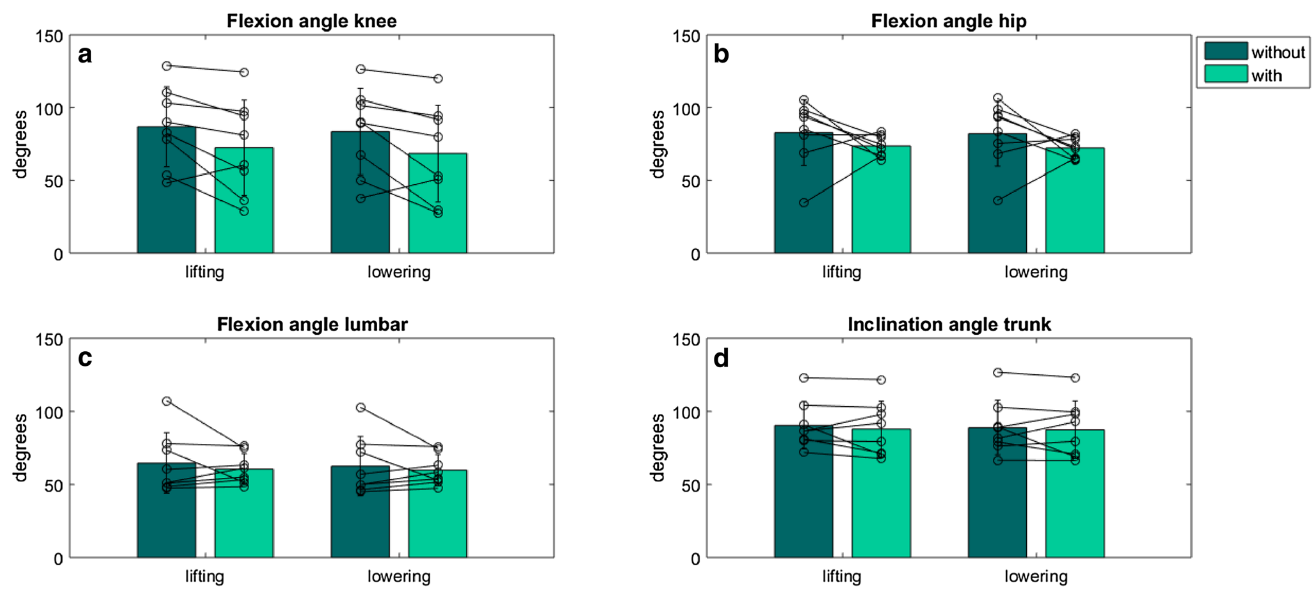
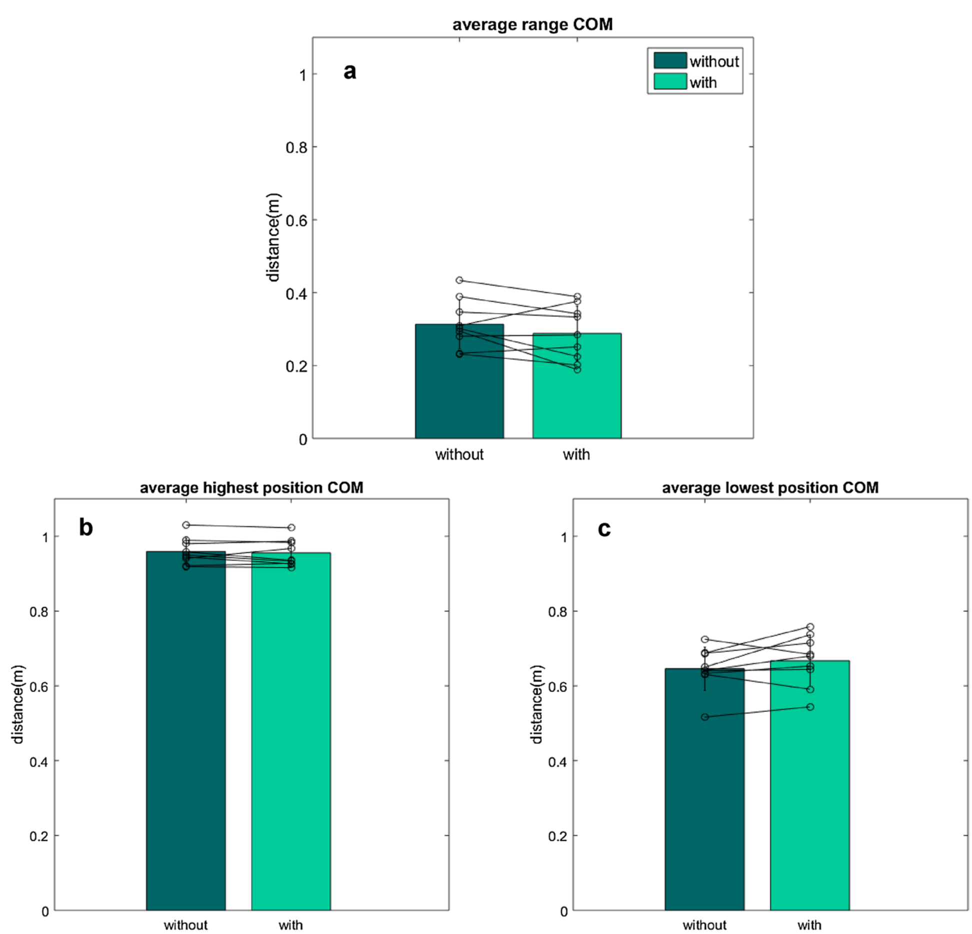


Fig. 5 Peak flexion angles in the knee (**a**), hip (**b**), L5S1 joint (**c**) and trunk (**d**), over one lifting cycle, with and without the exoskeleton, averaged over all participants. $N=8$. Error bars indicate standard deviations. Black lines indicate individual responses. "lifting" refers to bending down without the box and picking up the box to a standing position. "lowering" refers to putting down the box and coming up to a standing position without the box

Fig. 6 The range of motion of the center of mass when lifting with and without the exoskeleton (**a**), averaged over all participants. Highest position of the centre of mass, averaged over all participants (**b**). Lowest position of the centre of mass, averaged over all participants (**c**). $N=9$. Error bars indicate standard deviations. Black lines indicate individual responses



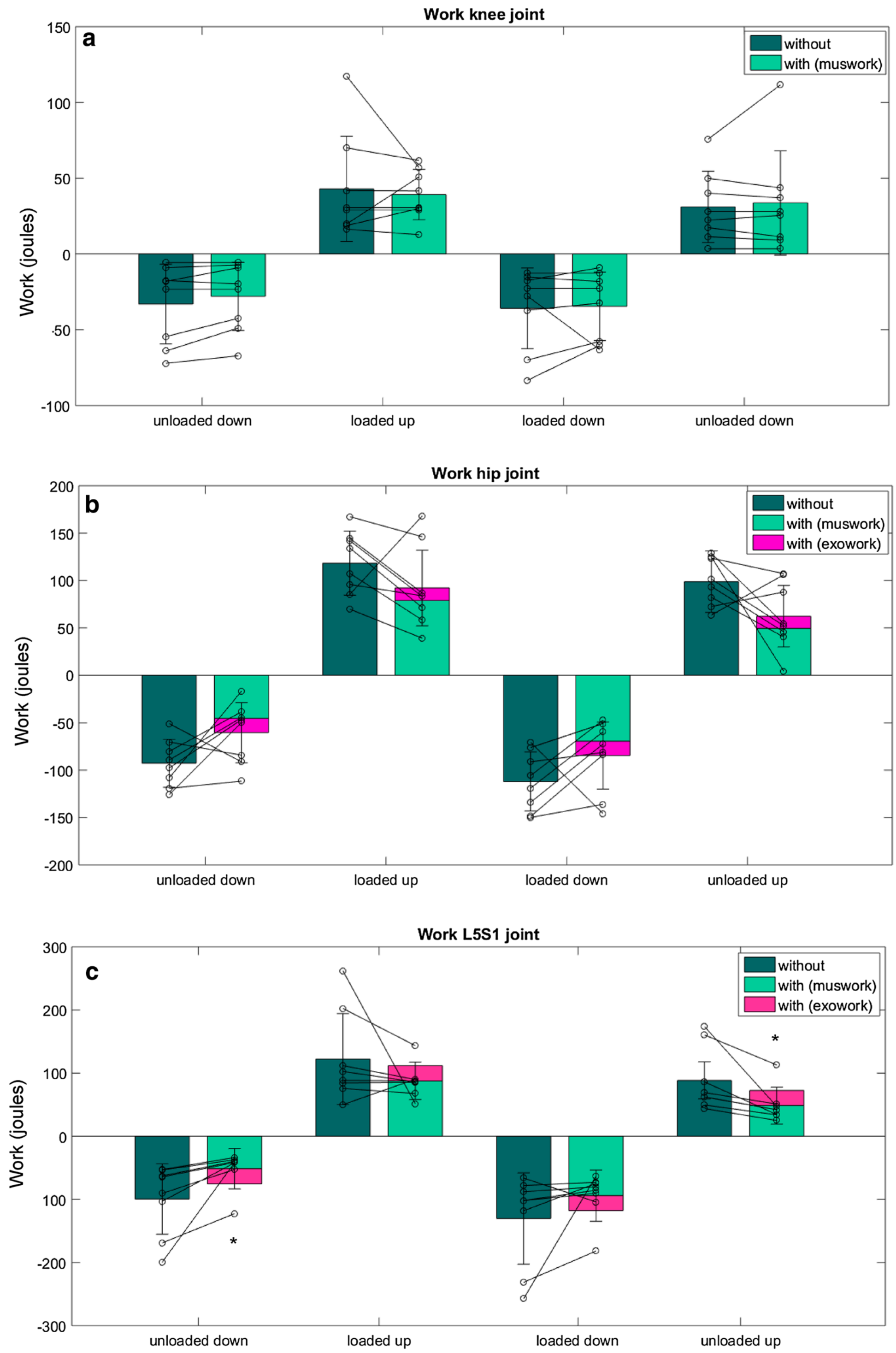


Fig. 7 Negative and positive work at the knee joint (**a**), hip joint (**b**) and L5S1 joint (**c**) over one lifting cycle, with and without the exoskeleton, averaged over all participants. $N = 8$. Error bars indicate standard deviations. Black lines indicate individual responses. "unloaded" refers to bending down and coming up without the box. "loaded" refers to picking up and putting down the box. The exoskeleton condition in **b** and **c** also shows the work generated by the exoskeleton (exowork) and the work generated by the muscles (muswork). *Significant change in work generated by the muscles (muswork) between control condition (without) and exoskeleton condition (with)

Discussion

The aim of this paper was to assess the effect of wearing a novel spinal exoskeleton (SPEXOR) on metabolic cost. We found that the exoskeleton reduces metabolic cost of repetitive lifting by 18%. Using this exoskeleton during repetitive lifting tasks might, therefore, reduce the risk of fatigue-related injury in the low back.

A reduction in metabolic cost can be caused by the exoskeleton taking over part of the mechanical work that has to be generated by the muscles to perform the lifting task. We, therefore, investigated the contribution of the exoskeleton and the muscles to the mechanical work generated

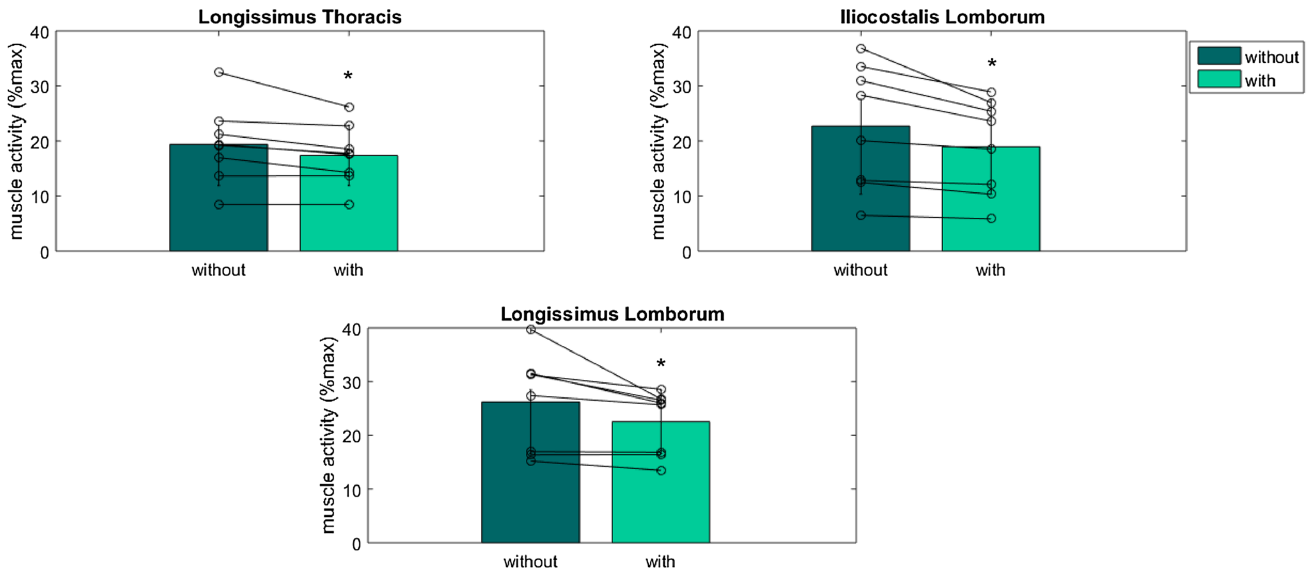


Fig. 8 Muscle activity of the back muscles with and without the exoskeleton, averaged over all participants. $N=8$. Error bars indicate standard deviations. Black lines indicate individual responses. *Significant difference in mean acitivity between control condition (without) and exoskeleton condition (with)

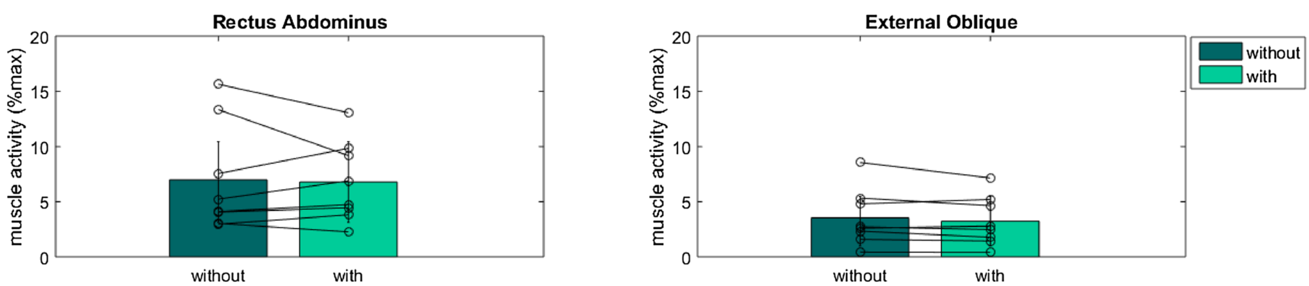


Fig. 9 Muscle activity of the abdominal muscles with and without the exoskeleton, averaged over all participants. $N=8$. Error bars indicate standard deviations. Black lines indicate individual responses

around the hip and L5S1 joint during lifting. The results show that the exoskeleton generated a substantial part of the total mechanical joint work at the hip and L5S1 joint, taking over part of the mechanical work that otherwise had to be generated by the muscles around these joints. At the respective joints, the exoskeleton generated up to 18% of the required work during the unloaded phase and up to 25% during the loaded phase. This is equivalent in magnitude to the reduction of metabolic cost (18%). Although the relation between mechanical work and metabolic work is not necessarily one-to-one, this suggests that metabolic cost is reduced by the exoskeleton taking over muscular work at the hip and the L5S1 joint. Despite the significant contribution to joint work of the exoskeleton, a significant reduction in muscle work was only found for the unloaded phases at the L5S1 joint. The loaded phases at the L5S1 joint and both phases at the hip joints showed the same trend of reduced muscle work, but this did not reach significance. This can be explained by the variation in individual participants' behaviour as can be observed in Fig. 7. Individual participants distributed mechanical work differently over the knee, hip and L5S1 joints, between conditions. This individual variation in work distribution over the joints precludes the statistical significance of the mean reduction in muscle work, appearing around the separate joints. Summing up, muscle work could be an option to avoid this problem. However, potential transport of muscle work between joints through bi-articular muscle, does not allow for such a simple summation (van Ingen Schenau et al. 1990). Still, the trend towards lower muscle work around the joints and the significant work done by the exoskeleton indicates that a transfer from active muscle work to the passive work of the exoskeleton can explain the reduction in metabolic cost.

Reducing muscular effort and unloading the back when wearing an exoskeleton have been assessed by previous studies, reporting reductions in L5S1 peak moments of 15–20%

(Koopman et al. 2019; Frost et al. 2009, Abdoli-E. and Stevenson 2008). This is comparable with our reductions in generated work around the supported joints (18–25%). Besides the effect of taking over muscular effort, a concern when using passive exoskeleton is increased muscular effort in the legs, which has been reported when wearing the HappyBack and the Bendezy (Barret and Fathallah 2001) and when using the BNDR (Ulrey and Fathallah 2013a). However, in the present study, work generated around the knee joint did not show differences between the conditions, indicating that the loading of the legs on average did not increase.

The reduction in muscle work is also supported by the effects on muscle activity. Muscle activity showed clear effects of wearing the exoskeleton. The SPEXOR device reduced back muscle activity by 10–16%, indicating that less muscular effort was needed to perform the lifting task when wearing the exoskeleton. Even though there is not necessarily a linear relation between muscle activity and metabolic cost (Bisi et al. 2011), the reduction in muscle activity is surprisingly similar to the observed reduction of net metabolic cost. Our EMG results are comparable to previous studies that assessed the effect of lifting devices on back muscle activity during repetitive lifting (Abdoli et al. 2006; Whitfield et al. 2014). Some studies reported higher reductions in back muscle activity (Bosch et al. 2016; Koopman et al. 2019; Wehner et al. 2009). However, they assessed the effect of wearing a passive exoskeleton during static bending, in which a continuous support of the exoskeleton is provided. In a dynamic task, such as repetitive lifting, this is not the case and this explains why we did not find as high reductions in back muscle activity. Alemi et al. (2019) reported a reduction of back muscle activity by 29% when using the VT-Lowe's exoskeleton during symmetric lifting. Potential causes for the greater reduction in comparison to the present study are the higher loads lifted by the participants in the study of Alemi et al. (2019) and unknown difference in support provided by the respective exoskeleton. Abdominal muscle activity did not change, indicating that participants did not have to activate their abdominals to overcome resistance of the exoskeleton. Still, the reduced back muscle activity demonstrates the potential of the SPEXOR exoskeleton to unload the low back by reducing back muscle activity and hence reducing metabolic cost.

A reduction in metabolic cost can also be a result of a changed lifting behaviour as shown in previous studies (Baltrusch et al. 2019b; Sadler et al. 2011). Using the SPEXOR exoskeleton in the present study, participants did not systematically change movement behaviour when wearing the exoskeleton. Neither individual joint angles nor the movement of CoM, representing the summation of effects of the different joint movements, was different between the conditions. The COM displacement is an important determinant of work performed in lifting and hence of metabolic cost. Changes in kinematics can, therefore, not explain the change in metabolic cost. Low interference with tasks and minor hindrance by the SPEXOR exoskeleton when performing work-related tasks has been shown in a previous study (Baltrusch et al. (2019a) and might explain why the movement behaviour without and with the exoskeleton remained the same. Despite the relatively high mass of the exoskeleton (6.7 kg), metabolic cost still decreased. This can be explained by the fact that the mass of the exoskeleton is close to the users' COM and, therefore, has less impact on the demand of lifting. The exoskeleton's mass may, however, affect comfort. Baltrusch et al. (2019a) recommended that the mass and the dimension of the same exoskeleton should be reduced to prevent from pressure points on the hip. This, however, was mainly a problem for walking. Squatting and lifting showed lower discomfort values.

The reduction in metabolic cost is in line with a previous study of Baltrusch et al. (2019b), in which it was found that the Laevo exoskeleton reduced metabolic cost during repetitive lifting by 17%. The Laevo exoskeleton generated a torque of up to 23 Nm around the lumbar joint (Koopman et al. 2019), whereas the SPEXOR exoskeleton generates a torque of up to 50 Nm around the L5S1 joint and up to 25 Nm around the hip joint. Given the higher support level, we expected a bigger effect on metabolic cost when wearing the SPEXOR exoskeleton, compared to wearing the Laevo. Still, the reduction of metabolic cost was about the same for both exoskeletons. This might be related to the fact that participants changed to a less demanding lifting strategy when wearing the Laevo and did not change movement behaviour with the SPEXOR. However, more research is needed to assess the difference between these two systems.

The results of this study should be interpreted in the light of some limitations. The lifting task was performed in a laboratory and its duration was chosen to reach steady state. In a real-work environment, lifting tasks can last longer and are often much more variable in terms of technique and frequency. Additional research is needed to assess the effect of the exoskeleton on metabolic cost when performing demanding tasks for a longer period of time in a realistic setting. Also, we did not measure muscle activity around the knee joint, which could have been interesting in relation to knee loading. The results, however, show that the work generated around the knee joint when wearing the exoskeleton did not change. This indicates that the muscles did not generate more work when using the exoskeleton and hence muscle activity probably would not change either. However, this may miss, for example, co-contraction. Another limitation is the fact that the effect of wearing the exoskeleton might be small in people with higher body mass than the participants tested. Besides, artefacts due to skin movement could have influenced kinematic outcomes. However, it is unlikely that these skin movements and the resulting errors are different between conditions. Hence, a measurement error due to skin movement acting as a confounding factor is highly unlikely. Our results cannot be generalized to other exoskeletons, as the effect of wearing an exoskeleton depends on the specific design characteristics of the device. The variation in individual behaviour highlights that different strategies can be used to exploit the effects of the exoskeleton and reduce metabolic cost. More insight into these different strategies and potential relations with participant characteristics might help to understand low back problems and further optimize trunk exoskeletons. However, the current sample is too small to allow further subgroup analyses.

Conclusion

The findings presented in this study demonstrate the potential of the SPEXOR exoskeleton to decrease metabolic costs of lifting and by that reducing the risk of getting fatigued during repetitive lifting. This effect can be explained by the exoskeleton taking over muscular work generated in the hip and the L5S1 joint, reduced back muscle activity, while on average movement strategy remains unchanged. As such, the SPEXOR exoskeleton may contribute to preventing work-related low-back pain for people executing highly demanding tasks, such as repetitive lifting.

Acknowledgements The authors would like to acknowledge the support of Michel Kaldenhoven from KLM, who made this testing possible.

Funding The work presented in this paper was supported by the European Union's Horizon 2020 research and innovation program under Grant Agreement no. 687662-SPEXOR.

Compliance with ethical standards

Conflict of interest The authors declare that they have no conflict of interest.

References

Abdoli-E M, Agnew MJ, Stevenson JM (2006) An on-body personal lift augmentation device (PLAD) reduces EMG amplitude of erector spinae during lifting tasks. Clin Biomech 21:456–465

Abdoli-Eramaki M, Stevenson JM (2008) The effect of on-body lift assistive device on the lumbar 3D dynamic moments and EMG during asymmetric freestyle lifting. Clin Biomech 23:372–380

Alemi MM, Geissinger J, Simon AA, Chang SE, Asbeck AT (2019) A passive exoskeleton reduces peak and mean EMG during symmetric and asymmetric lifting. J Electromyogr Kinesiol 47:25–34

Baltrusch SJ, van Dieën JH, van Bennekom CAM, Houdijk H (2019a) Testing an Exoskeleton that helps workers with low-back pain. Less discomfort with the passive SPEXOR trunk device. IEEE robotics and automation society special issue: Wearable Robotics

Baltrusch SJ, van Dieën JH, Bruijn SM, Koopman AS, van Bennekom CAM, Houdijk H (2019) The effect of a passive trunk exoskeleton on metabolic cost during lifting and walking. Ergonomics 1:1–14

Barret AL, Fathallah FA (2001) Evaluation of Four weight transfer devices for reducing loads on the lower back during agricultural stoop labor. Paper No. 01-8056 of the ASAE Meeting, Sacramento, USA.

Bergstrom G, Hagberg J, Busch H et al (2014) Prediction of sickness absenteeism, disability pension and sickness presenteeism among employees with back pain. J Occup Rehabil 24:278–286

Bisi MC, Stagni R, Houdijk H, Gnudi G (2011) An EMG-driven model applied for predicting metabolic energy consumption during movement. J Electromyogr Kinesiol 21(6):1074–1080

Bosch T, van Eck J, Knitel K, de Looze M (2016) The effects of a passive exoskeleton on muscle activity, discomfort and endurance time in forward bending work. Appl Ergon 54:212–217

Cappozzo A, Catani F, Della Croce U, Leardini A (1995) Position and orientation in space of bones during movement: anatomical frame definition and determination. Clin Bioemech 10(4):171–178

Coenen P, Kingma I, Boot CR, Bongers PM, van Dieen JH (2014a) Cumulative mechanical low-back load at work is a determinant of low-back pain. Occup Environ Med 71:332–337

Coenen P, Gouttebarge V, van der Burght ASAM, van Dieen JH, Frings-Dresen MHW, van der Beek AJ, Burdorf A (2014b) The effect of lifting during work on low-back pain—a health impact assessment based on a meta-analysis. Occup Environ Med. 71:871–877

De Looze MP, Bosch T, Krause F, Stadler KS, O'Sullivan LW (2016) Exoskeletons for industrial application and their potential effects on physical work load. Ergonomics 59(5):671–681

Dupeyron A, Rispens SM, Demattei C, van Dieën JH (2013) Precision of estimates of local stability of repetitive trunk movements. Eur Spine J 22:2678–2685

Frost DM, Abdoli-E M, Stevenson JM (2009) PLAD (personal lift assistive device) stiffness affects the lumbar flexion/extension moment and the posterior chain EMG during symmetrical lifting tasks. J Electromyogr Kinesiol 19:403–412

Garby L, Astrup A (1987) The relationship between the respiratory quotient and the energy equivalent of oxygen during simultaneous glucose and lipid oxidation and lipogenesis. Acta Physiol Scand 129:443–444

Garg A, Herrin GD (2007) Stoop or squat: a biomechanical and metabolic evaluation. IISE Trans 11(4):293–302

Graham RB, Agnew MJ, Stevenson JM (2009) Effectiveness of an on-body lifting aid at reducing low back physical demands during an automotive assembly task: assessment of EMG response and user acceptability. Appl Ergon 40(5):936–942

Graham RB, Sadler EM, Stevenson JM (2012) Local dynamic stability of trunk movements during the repetitive lifting of loads. Hum Mov Sci 31(3):592–603

Granata KP, Gottipati P (2008) Fatigue influences the dynamic stability of the torso. Ergonomics 51(8):1258–1271

Hoy D, Bain C, Williams G et al (2012) A systematic review of the global prevalence of low back pain. Arthr Rheum. 64:2028–2037

Jones AM, Poole DC (2005) Oxygen uptake dynamics: from muscle to mouth–an introduction to the symposium. Med Sci Sports Exerc. 37(9):1542–1550

Kingma I, de Looze MP, Toussaint HM, Klijnsma HG, Bruijnen TBM (1996) Validation of a full body 3-D dynamic linked segment model. Hum Mov Sci 15:833–860

Kingma I, Faber GS, van Dieën JH (2010) How to lift a box that is too large to fit between the knees. Ergonomics 53(10):1228–1238

Koopman AS, Kingma I, Faber GS, de Looze MP, van Dieën JH (2019) Effects of a passive exoskeleton on the mechanical loading of the low back in static holding tasks. J Biomech 83:97–103

McGill SM, Sharratt MT, Seguin JP (2007) Loads on spinal tissues during simultaneous lifting and ventilatory challenge. Ergonomics 38(9):1772–1792

Näf MB, Koopman AS, Baltrusch S, Rodriguez-Guerrero C, Vanderborght B, Lefeber D (2018) Passive back support exoskeleton improves range of motion using flexible beams. Front Robot AI 5(72):1–16

Pataky TC, Robinson MA, Vanrenterghem J (2013) Vector field statistical analysis of kinematic and force trajectories. J Biomech 46(14):2394–2401

Roelen CA, Koopmans PC, Groothoff JW (2010) Subjective health complaints in relation to sickness absence. Work. 37:15–21

Sadler EM, Graham RB, Stevenson JM (2011) The personal lift-assist device and lifting technique: a principal component analysis. Ergonomics 54(4):392–402

Thiese MS, Hegmann KT, Wood EM, Garg A, Moore JS, Kapellusch J, Foster J, Ott U (2014) Prevalence of low back pain by anatomic location and intensity in an occupational population. BMC Musculoskelet Dis 15:238

Ulrey BL, Fathallah FA (2013a) Subject-specific, whole-body models of the stooped posture with a personal weight transfer device. J Electromyogr Kinesiol 23(1):206–215

Ulrey BL, Fathallah FA (2013b) Effect of a personal weight transfer device on muscle activities and joint flexions in the stooped posture. J Electromyogr Kinesiol 23(1):195–205

Van Ingen Schenau GJ, Cavanagh PR (1990) Power equations in endurance sports. J Bioemech 23(9):865–881

Wehner M, Rempel D, Kazerooni H (2009) Lower extremity exoskeleton reduces back forces in lifting. ASME Dyn Syst Control Conf 2:49–56.

Whitfield BH, Costigan PA, Stevenson JM, Smallman CL (2014) Effect of an onbody ergonomic aid on oxygen consumption during a repetitive lifting task. Int J Ind Ergon 44:39–44

Zatiorsky VM (2002) Kinetics of human motion. Human Kinetics, Champaign