Effects of a passive exoskeleton on the mechanical loading of the low back in static holding tasks

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ABSTRACT

With mechanical loading as the main risk factor for LBP in mind, exoskeletons are designed to reduce the load on the back by taking over a part of the required moment. The present study assessed the effect of a passive exoskeleton on back and abdominal muscle activation, hip and lumbar flexion and on the contribution of both the human and the exoskeleton to the L5/S1 net moment, during static bending at five different hand heights. Two configurations of the exoskeleton (LOW & HIGH) differing in angle-torque characteristics were tested. L5/S1 moments generated by the subjects were significantly reduced (15–20% for the most effective type) at all hand heights. LOW generated 4–11 Nm more support than HIGH at 50%, 25% and 0% upright stance hand height and HIGH generated 4–5 Nm more support than LOW at 100% and 75%. Significant reductions (11–57%) in back muscle activity were found compared to WITHOUT for both exoskeletons for some conditions. However, EMG reductions compared to WITHOUT were highly variable across subjects and not always significant. The device allowed for substantial lumbar bending (up to 70°) so that a number of participants showed the flexion-relaxation phenomenon, which prevented further reduction of back EMG by the device and even an increase from 2% to 6% MVC in abdominal activity at 25% hand height. These results indicate that flexion relaxation and its interindividual variation should be considered in future exoskeleton developments.

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1. Introduction

Low-back pain (LBP) is the number one cause of disability in the world (Hartvigsen et al., 2018), with a lifetime prevalence between 75 and 84% (Thiese et al., 2014). Mechanical loading of the low back has been shown to be an important risk factor for the development of LBP (Coenen et al., 2014; Coenen et al., 2013; da Costa and Vieira, 2010; Kuiper et al., 2005; Norman et al., 1998). Still, many workers are exposed to tiring and painful positions and are required to lift heavy loads at least a quarter of the work time (Eurofound, 2012). Although physically demanding jobs have become less prevalent due to mechanization and automation, some remain because they require the mobility and flexibility of the human.

In an attempt to optimize these manual material handling jobs, many studies have addressed the effects of lifting style and task conditions on the physical load that workers are exposed to. Reductions up to 20% were found by for example placing one hand at the thighs during lifting (Kingma et al., 2015) or by optimizing task conditions (e.g. mass, height, horizontal distance etc.) (Faber et al., 2009; Faber et al., 2011; Hoozemans et al., 2008; Kingma et al., 2004; Marras et al., 1999). However, it appeared that workers reduced effects of task improvements by adjusting their lifting behavior, so that effects in real working conditions can be questioned (Faber et al., 2007).

More recently, body worn assistive devices (exoskeletons) have been developed and have been introduced in the work place. The main aim of low-back exoskeletons is to prevent injury, while preserving the versatility of workers during tasks involving forward bending. With mechanical loading as the main risk factor for LBP in mind, exoskeletons are designed to reduce the load on the back by taking over a part of the back muscular activity needed to counteract moments due to gravity on the upper body and loads handled. In so-called passive exoskeletons, passive spring like components are used to generate an extension moment while bending forward, see for an extensive review de Looze et al. (2016).

Several passive exoskeletons (including the EXO tested in this study) have already been evaluated and have shown their effect

in reducing back muscle activity as measured by electromyography (EMG) by 10–40% (Abdoli et al., 2006; Abdoli and Stevenson, 2008; Barrett and Fathallah, 2001; Bosch et al., 2016; de Looze et al., 2016; Graham et al., 2009; Lotz et al., 2009; Ulrey and Fathallah, 2013). However, without knowledge of the lumbar flexion angle, which was in most studies not reported, EMG-based inferences on effects of a device on spine loading, are premature. As has been shown by Ulrey and Fathallah (2013) the effect of the device on EMG activity highly depends on lumbar flexion. This is especially the case when full flexion is approached, due to the occurrence of flexion relaxation, where EMG of back muscles becomes silent as stretched connective tissues generate the extension moment required (Floyd and Silver, 1955). So, in order to interpret the effects of the device, it is important to both report EMG and lumbar flexion. Additionally, it is important to know the actual moment that the device generates. Regrettably however, none of the previous studies reported moments of the device or moment reductions for the subjects, when using the device.

Therefore, the present study assessed the effect of a passive exoskeleton on not only abdominal and back muscle activation, but also on hip and lumbar flexion and on the contribution of both the human and the exoskeleton to the L5/S1 net moment, during static forward bending tasks at a range of five hand heights. Two configurations of the exoskeleton (Laevo BV, The Netherlands) differing in angle-torque characteristics were tested. We hypothesized that both devices reduce the flexion-extension L5/S1 moment generated by the subject and consequently reduce back muscle activity.

2. Methods

2.1. Exoskeleton

The passive exoskeleton (Laevo V2.4 Delft, Netherlands) that was tested in the study is presented in Fig. 1. The device applies forces at three places on the body: thighs, pelvis and on the chest. While bending forward, resistance is applied at the chest and the legs due to a spring-loaded joint in series with an elastic beam, generating a moment in parallel to the back-muscle moment. To be able to determine the contribution of the exoskeletons to the L5/S1 net moment during the static bending tasks, both devices were tested in a calibration trial. Two different types of the device were tested, namely LOW, showing a gradual increase in support between 30 and 140 degrees of Laevo joint flexion, and HIGH, showing a peak support around 50 degrees of Laevo joint flexion (Fig. 2). A substantial difference in support was found between the bending phase (solid) and extending phase (dashed), due to friction within the system. The Laevo joint includes an end stop (i.e. mechanical stop in the joint). Beyond this angle, no further rotation in the joint is possible. Further bending of the human results in deformation of the flexible beams, explaining the sharp increase in force and moment after about 140 degrees in both exoskeletons. Note that, during the static bending tasks the support moment was calculated around the L5S1 joint instead of the Laevo joint.

2.2. Subjects and experimental procedures

Eleven healthy male subjects (age: 24.1 ± 2.7 years, mass: 74.8 ± 7.4 kg, height: 1.84 ± 0.07 m), participated in the study, which was approved by the medical ethical committee of the VU medical center (VUmc, Amsterdam, The Netherlands, NL57404.029.16). After providing written informed consent, subjects were fitted and familiarized with the exoskeleton and anthropometric data were obtained. Thereafter, maximum voluntary contractions were obtained for the back and abdominal muscles in supine posture (McGill, 1991). After all instrumentation was placed on the subjects and calibration measurements were performed, a range of motion (ROM) trial was performed, only in the flexion-extension direction, to determine the maximal lumbar angle. After that, the experiment started in which subjects were asked to bend forward till their hands were at five predetermined heights (100%, 75%, 50%, 25% and floor height 0%) relative to standing upright and to hold this position for around five seconds (Fig. 3). Participants were instructed to hold their knees as straight as possible but without locking the knee joint. This task was repeated three times once without the device (WITHOUT), once with the LOW and once with the HIGH exoskeleton. The order of the device conditions was randomized over subjects to eliminate any order effects. Since changing between device conditions took around 10 min, sufficient rest between tasks was ensured.
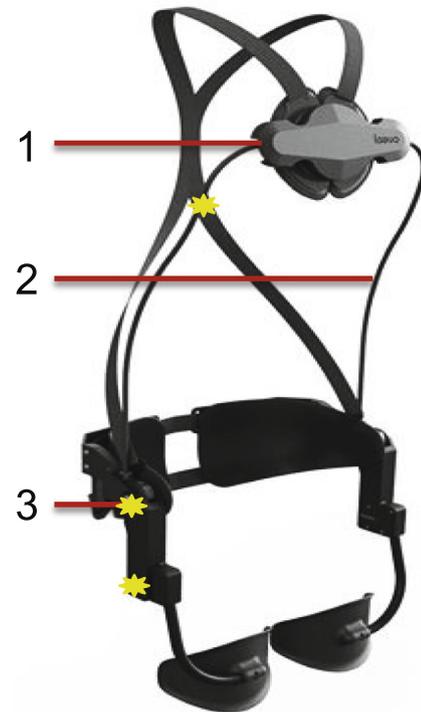

Fig. 1. Laevo (V2.4) (Laevo, Delft, The Netherlands). (1) Rotational chest pad, (2) Flexible beam, (3) Spring loaded joint. During the experiments, three LED's were used to measure the Laevo joint flexion angle. The user's trochanter major is to be aligned with the center of rotation of the device.

2.3. Instrumentation and data pre-processing

A single custom-made 1.0×1.0 m force plate was used to measure ground reaction forces at 200 samples/s. Off-line, forces were bi-directionally filtered with a 2nd order low-pass Butterworth filtered with a cut-off frequency of 10 Hz. Kinematics were collected at a sample rate of 50 samples/s using an opto-electronic 3D movement registration system (Certus, Optotrak, Norton Digital Inc.) and subsequently bi-directionally low-pass filtered with a 2nd order Butterworth filter at 5 Hz. LED cluster markers were attached to body segments (feet with lower legs, upper legs, pelvis, trunk (T10), head, upper arms and forearms with hands). In addition, three single LED markers were attached to the exoskeleton (base, joint, and bar) to measure the 'hip' joint angle of the exoskeleton (LEAVO hip joint). Prior to the measurements, for each participant, cluster markers were related to anatomical landmarks using pointer measurements (Cappozzo et al., 1995). Ten pairs of surface EMG electrodes were attached to the trunk muscles (Rectus Abdominis (RA), External Oblique (EO), anterior part of Internal Oblique (IO), Iliocostalis (IL), and Longissimus pars lumborum (LL); see Kingma et al., 2010) after abrasion and cleaning with alcohol. EMG data were recorded at 2000 samples/s using the Wireless Bipolar Cometa Mini Wave Plus 16-channel EMG system (Cometa, Bareggio, Italy) and online filtered with a band-pass filter (10–1000 Hz). EMG data were synchronized using a pulse generated at the instant the recording of the kinematics and kinetics started.

2.4. Data analysis

L5/S1 flexion-extension moments, generated by subject plus exoskeleton (M_{tot}) were calculated based on the GRF and lower-body kinematics, using a bottom-up inverse dynamics model (Kingma et al., 1996) with improved anthropometric modeling (Faber et al., 2009). A global equation of motion (rather than a segment by segment calculation) was used, as described by (Hof, 1992). Using the measured Laevo angle, the force applied on the chest pad could be calculated based on the angle-force relationship from Fig. 2 (see section about the calibration of Laevo). Subsequently, the Laevo moment around L5/S1 (M_{Laevo_L5S1}) was calculated using a cross product of the 3D moment arm and chest pad force and subtracted from M_{tot} to calculate the L5S1 moment generated by the subject (M_{sub}). Lumbar flexion was calculated based on the orientation of the clusters placed on the pelvis and the trunk. Trunk inclination was calculated relative to the global coordinate system. Off-line, EMG signals were full-wave rectified and low-pass filtered at 2.5 Hz (Potvin et al., 1996). EMG data were normalized to maximum voluntary contractions (McGill, 1991) and averaged over both sides of the body. The mean amplitude over the five seconds in each static posture was taken.
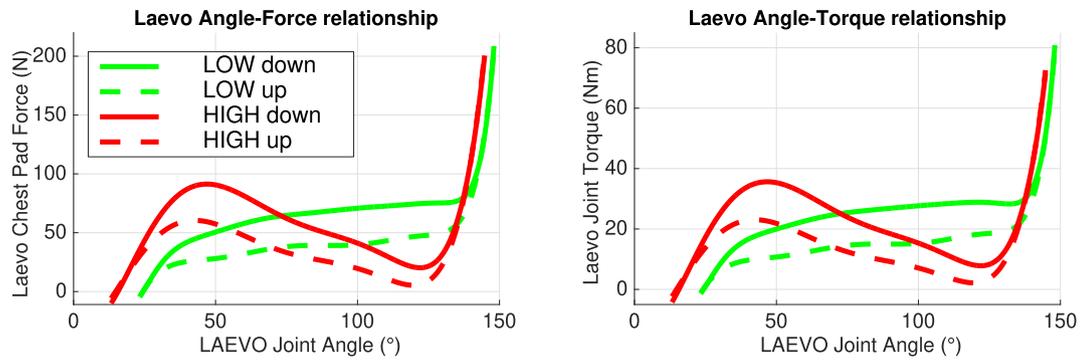
Fig. 2. Angle-force relationships measured with two different subjects for both LOW and HIGH. Force was measured using a force transducer placed on the chest pad. Orientation of the chest pad was tracked using a marker cluster. The Laevo joint flexion angle was captured using three LED's (Fig. 1). Note that an offset was present in this flexion angle during upright standing. The Laevo torque was calculated around the Laevo joint using a cross product of the force and the vector from the point of application of the force to the center of rotation of the joint. A substantial difference in support was found between bending (solid) and extending (dashed), due to friction within the system. Beyond the end stop of the Laevo joint, support rapidly increased due to bending of the flexible beams. Note that, during the static bending tasks the Laevo torque was calculated around L5S1 and not around the Laevo joint itself, like in this figure.
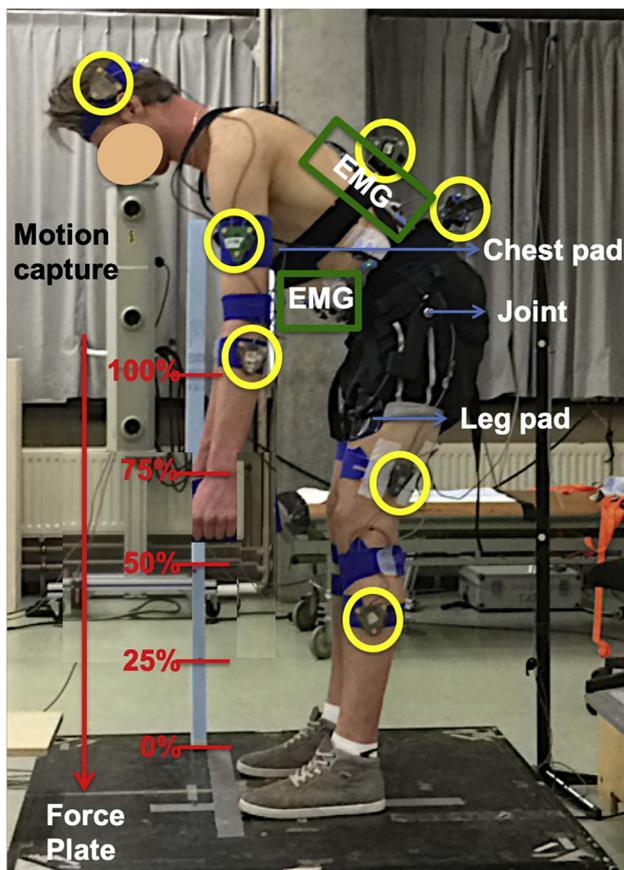

Fig. 3. Overview of the experimental setup showing the force plate, marker clusters (yellow), EMG location (green) and an indication of the five hand heights. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

2.5. Statistics

Outcome variables were moments (M_{tot}, M_{sub} and M_{Laevo_L5S1}), hip flexion, lumbar flexion, trunk inclination and back (IL and LL) and abdominal (RA, EO and IO) muscle activity. For the moment and angle variables, a two-way repeated measures analysis of variance (ANOVA) was conducted with device (WITHOUT, LOW and HIGH) and height (100%, 75%, 50%, 25% and 0%) as within subject factors. Device effects were further explored using Bonferroni post-hoc tests when a significant main effect of device or an interaction between device and height was found. Main effects of height were not further explored because, we were mainly interested in

the effects of the device and how device effects depend on height (the interaction), rather than on effects of height themselves. EMG data were not normally distributed and therefore non-parametric tests (Wilcoxon Signed Rank test) were conducted for each height separately. A significance level of $p < 0.05$ was used.

3. Results

3.1. Kinetic and kinematic variables

A significant main effect of height was found for all variables, indicating that all variables changed depending on hand height (Fig. 4, Table 1). As expected, while the subject-generated moment (M_{sub}) showed a main effect of device condition, but the total L5/S1 moment (including subject and device) and trunk inclination showed no significant main effect of device and no significant interaction with device condition, showing that no major differences in task execution existed between device conditions. M_{Laevo_L5S1}, Hip flexion and M_{sub} did show significant interactions. M_{Laevo_L5S1} significantly differed between LOW and HIGH for all heights except 0%. Support of HIGH was highest at 75% (20 Nm). In contrast, peak support of LOW occurred at 0% (23 Nm). For LOW and HIGH, hip flexion was reduced compared to WITHOUT for hand heights at 50% and lower. At 0%, hip flexion was reduced with 8 and 9 degrees for LOW and HIGH, respectively. M_{sub} differed significantly for all heights between LOW and HIGH compared to WITHOUT. At 50% and 25%, LOW was significantly lower compared to HIGH as a result of the smaller M_{Laevo_L5S1} generated in this angle range by HIGH. At 0% hand height, no significant differences were found between LOW and HIGH. This could be explained by the effect of the end stop, present in both types, leading to a similar increase in support in this range of flexion angles for both LOW and HIGH. No significant differences in lumbar flexion were found between device conditions. Surprisingly, at 75% hand height lumbar flexion was already above 50 degrees which was around 80% of the participants' maximum (Figs. 4F and 5A). As will be further explained below, this had major implications for lumbar back muscle activity.

3.2. Abdominal muscles

Abdominal muscles (Fig. 4) were almost silent (<3% MVC) at 100–50% hand height (except for IO at 100%). At 25% hand height, EO and IO activity with LOW was significantly higher compared to without. At 0%, activity of the abdominals increased to more than 10% MVC, on average. Activity with HIGH at 0% was, for all abdominals, significantly lower compared to LOW. At 0% hand height,
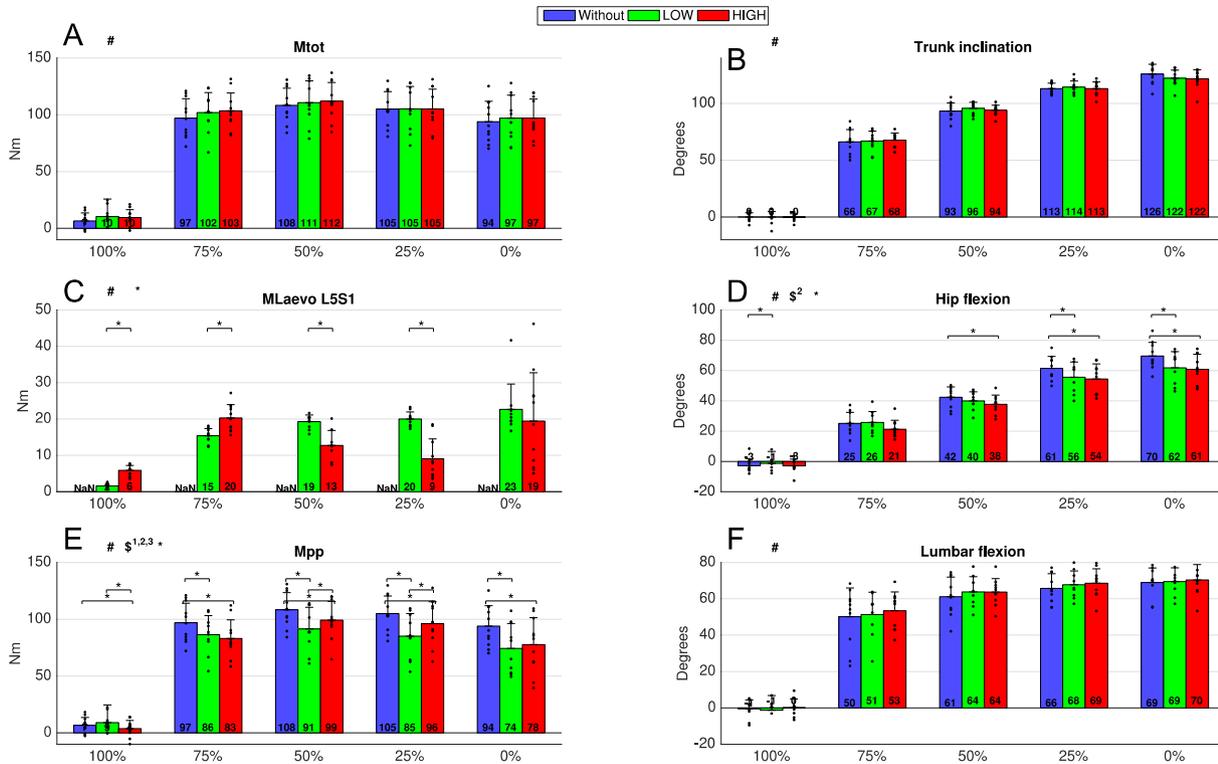
Fig. 4. Mean moments (Mtot, Mlaevo_L5S1 and Mpp in column 1) and angles (trunk inclination, hip flexion and lumbar flexion in column 2). A main effect of height was indicated with #. A main effect of device was indicated with a $ where 1, 2 and 3 stands for significant differences between WITHOUT and LOW, WITHOUT and HIGH and LOW and HIGH, respectively. * Indicated a significant interaction between device and height and subsequently, post hoc Bonferroni corrected *t*-test's were performed. Horizontal bars indicated a significant difference between the two bars.

Table 1
F-values, p-values and effect sizes of repeated measures ANOVA's with Device condition (WITHOUT, LOW and HIGH), Height condition (100%, 75%, 50%, 25% and 0%) and their interaction. Pairwise comparisons were only performed for variables with a significant main effect of device. Significant (p < 0.05) results were indicated in bold.

	Main effect			Main effect			Interaction			Pairwise comparisons device					
	Device			Height			Device * Height			1–2		1–3		2–3	
	F	p	η^2	F	p	η^2	F	p	η^2	Diff	p	Diff	p	Diff	p
M_{tot}	2.62	0.10	0.22	**169.10**	**<0.01**	**0.95**	1.06	0.33	0.11						
Trunk inclination	0.15	0.86	0.01	**1073.27**	**<0.01**	**0.99**	3.02	0.11	0.23						
M_{Laevo_L5S1}	1.63	0.23	0.14	**18.72**	**<0.01**	**0.65**	**6.66**	**0.03**	**0.40**						
Hip flexion	**5.62**	**0.01**	**0.41**	**224.18**	**<0.01**	**0.97**	**5.05**	**0.01**	**0.39**	2.46	0.20	**4.35**	**0.02**	1.89	0.78
M_{sub}	**83.72**	**<0.01**	**0.90**	**113.08**	**<0.01**	**0.93**	**11.57**	**0.01**	**0.56**	**13.53**	**<0.01**	**11.01**	**<0.01**	−2.59	0.19
Lumbar flexion	0.33	0.72	0.04	**225.44**	**<0.01**	**0.97**	0.71	0.42	0.08						

some subjects reached beyond their ROM (>100%), which had been recorded in a separate trial (Fig. 5A). Those subjects showed EO activity up to almost 80% MVC.

3.3. Lumbar back muscles

EMG activity of the lumbar back muscles decreased below 75% hand height and went to almost 0% MVC at 0% hand height. EMG activity of IL was significantly lower compared to without at 50% (LOW) and 25% (HIGH) hand height. Although median IL activity at 75% was substantially reduced in HIGH (6% MCV) compared to WITHOUT (12% MVC), this difference did not reach significance, because some subjects showed an increase due to a small decrease in lumbar flexion. Implications of this will be further addressed in the discussion. For LL, EMG activity was reduced compared to WITHOUT at 50% for LOW only. At 25% hand height some subjects showed a clear reduction in EMG activity when wearing the device. However, a number of subjects showed already low EMG levels at 25% WITHOUT, so that the effect of device, on a group level, was

not statically significant. Moreover, as can be seen from the individual data points (Fig. 5), some subjects had almost no lumbar back muscle activity already at 75% hand height. Indeed, some subjects reached over 100% of their lumbar ROM (Fig. 5A) and showed the flexion-relaxation phenomenon (Fig. 5C & E). When standing up straight (100%), lumbar back muscles activity with LOW was significantly higher compared to WITHOUT and HIGH (related to a higher net moment).

4. Discussion

The present study investigated the effect of two different types (HIGH and LOW) of a passive exoskeleton on muscle activity, hip and lumbar flexion angles and moments. The L5/S1 flexion-extension moment generated by the subjects (M_{sub}) was significantly lower compared to WITHOUT for both types at all flexed positions (75%, 50%, 25% and 0%). Hip flexion was reduced compared to WITHOUT while wearing the exoskeletons at the two most flexed positions (25% and 0%). Inspection of knee angles
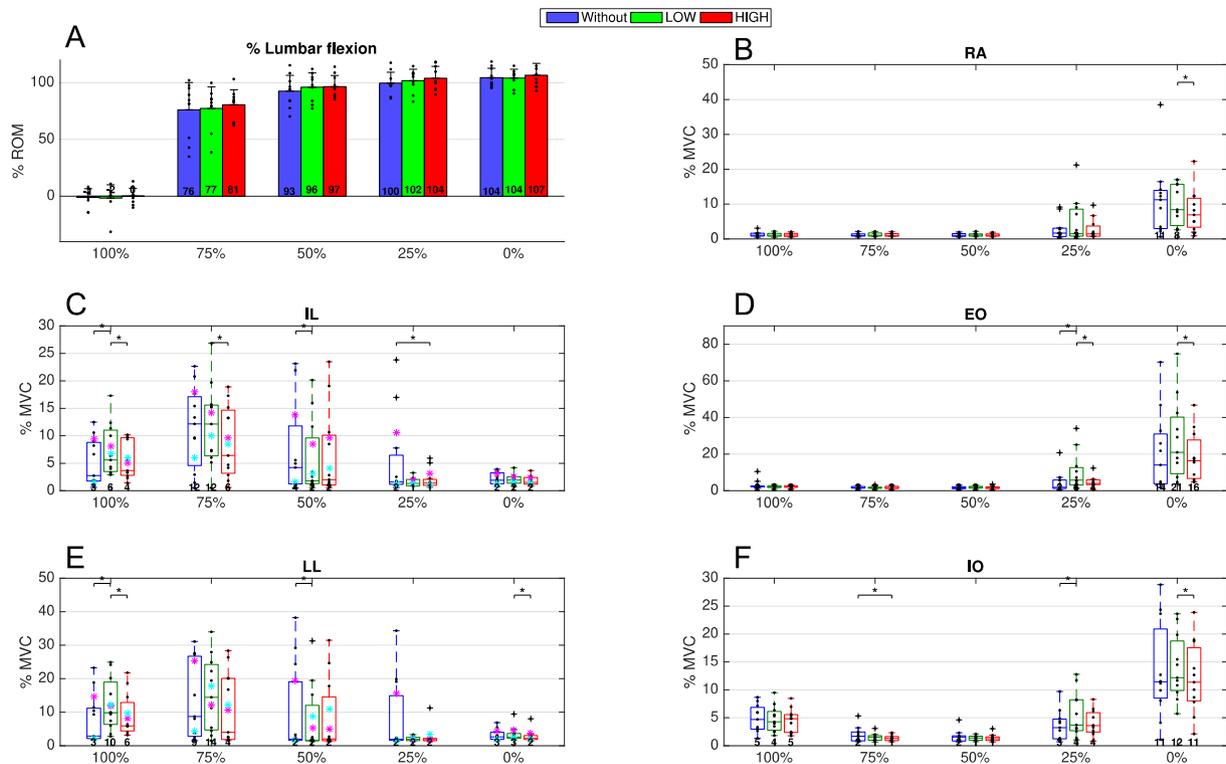
Fig. 5. A, Percentage lumbar flexion with respect to the maximal lumbar flexion angle recorded in the ROM trial. B-F, EMG activity of the Iliocostalis (IL), Longissimus Lumborum (LL), Rectus Abdominus (RA), External Oblique (EO) and Internal Oblique (IO) as percentage of their MVC. In the boxplots, the central mark is the median, the edges of the box are the 25th and 75th percentiles, the whiskers extend to the most extreme datapoints to be not outliers (<2.7 SD), and the outliers (+) are plotted individually. Bonferroni corrected post hoc *t*-test's were performed, horizontal bars indicate a significant (p < 0.05) difference between the two bars.

revealed more extension in the EXO conditions. Lumbar flexion did not significantly differ between device conditions. Some significant reductions in lumbar back muscle activity were found compared to WITHOUT for both LOW and HIGH (Fig. 5C & E), however, not fully consistent with the expectations based on the exoskeleton and subject generated moments (Fig. 4C & E). Below we will first discuss the absolute and relative contribution of the exoskeleton to the required net moment, the effects on EMG activity, and the influence of the flexion-relaxation phenomenon. Finally, we will discuss the limitations of the current study together with possible improvements of future exoskeletons.

Maximum support for LOW and HIGH was around 23 (0% hand height) and 20 Nm (75% hand height), respectively. With respect to the required net moment, which was around 97 Nm and 103 Nm at that instant, the contribution of the exoskeleton was around 24% and 19% for LOW and HIGH. Spinal compression forces are highly correlated with the L5S1 net joint moment (van Dieen and Kingma, 2005). Therefore, in the absence of abdominal co-contraction (see below) the exoskeletons are likely to reduce spine compression by about 20% at the above-mentioned heights. For a given time working in these positions, cumulative loading, which has been shown to be a risk factor for development of LBP (Coenen et al., 2014), will be reduced as well.

Despite the clear reductions observed in the required net moment for the subjects at all heights due to the contribution of the exoskeleton, back muscle activity was only significantly reduced at 50% hand height for LOW and at 25% hand height for HIGH. Interpretation of EMG back muscle activity can be complicated due to the flexion-relaxation phenomenon, as a reduction in EMG doesn't always mean that back loading is reduced. The reduction in EMG could also reflect a shift in loading from active to passive structures. In addition, between individuals flexion-relaxation occurs at different lumbar flexion angles. Therefore,

lumbar EMG can strongly vary between subjects in a given posture, and subtle posture variations can mask exoskeleton effects on EMG. Somewhat surprisingly, at the second highest hand height (75%), lumbar flexion was already around 50 degrees (Fig. 4F), which was around 80% of the average ROM of the participants (Fig. 5A). For some participants in which flexion-relaxation was already present, no further reduction in lumbar EMG activity was possible and this obscured the overall effect, while at the individual level effects were observed in subjects that still had muscle activity in the WITHOUT condition. To illustrate this, a median split was performed on the EMG activity in the WITHOUT condition on each of the hand heights and subsequently, the average of the two groups for the LOW and HIGH conditions was calculated, see blue and purple points in Fig. 5C & E. Consistent and substantial reductions in EMG activity were observed in the group with substantial EMG activity (purple) for both muscle groups. In contrast, unchanged or even a slight increase in EMG was seen in the group with minimum EMG. As a result, overall differences were not significant. Fig. 6, shows the highly non-linear relation between lumbar flexion and back muscle activity for a typical subject in the ROM trial. Although this trial is different from the static bending tasks, it clearly shows that a change in lumbar flexion of less than one degree can lead to a change in muscle activity of more than 20% MVC.

The observed reductions in back muscle activity due to the exoskeleton are comparable with previous literature, especially when our results are compared to Bosch et al. (2016) who evaluated the same device. In other studies, evaluating comparable passive exoskeletons during static holding tasks, reductions of back muscle activity ranged from 10% to around 40% (de Looze et al., 2016; Graham et al., 2009; Ulrey and Fathallah, 2013). However, not all studies reported lumbar flexion angles, so caution should be taken in comparing the results, as lumbar flexion has a major
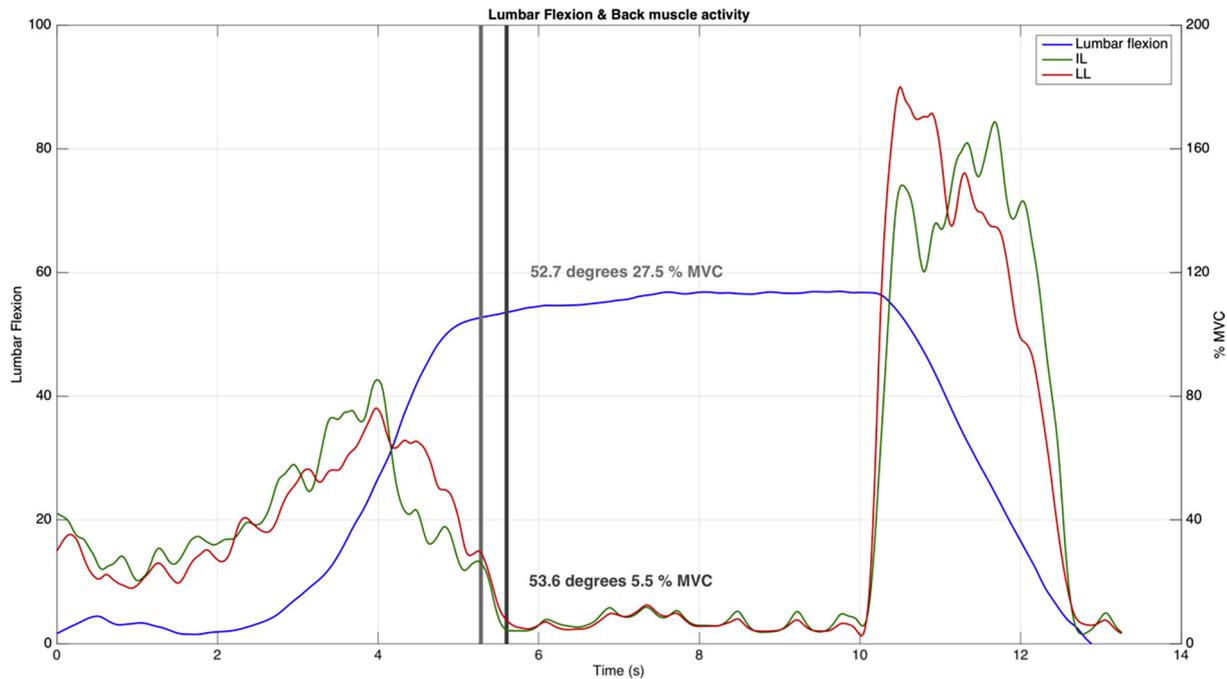
Fig. 6. Typical example of a ROM trial showing the relation between lumbar flexion and muscle activity of the two back muscles. Vertical lines show the EMG values at two different lumbar flexion angles.

effect on back muscle activity. In Ulrey and Fathallah (2013) it was concluded that the device significantly reduced muscle activity of subjects who did not experience flexion-relaxation. This is in line with the effects we observed in subjects who did not experience flexion-relaxation. Our finding of a slightly increased lumbar flexion (2–3 degrees) while wearing the exoskeleton was in contrast with a reduction up to six degrees reported previously (Ulrey and Fathallah, 2013), which may be due to subtle differences in device or task constraints.

With the increase of abdominal muscle activity at hand heights at and below 25% and the presence of the flexion-relaxation phenomenon at the same time in a number of subjects, it is expected that compression forces will be increased when wearing the exoskeleton compared to WITHOUT in these subjects. This is because the sum of the passive moment and the support torque of the exoskeleton exceeds the net required moment. Consequently, subjects have to activate their abdominals (co-contraction) to stay in the required position. However, with uncertainty about the tissues being responsible for the passive moment generating capacity and thus the moment arms that these structures have with respect to L5/S1, it remains to be seen how large the effect on spinal compression would be. The present results clearly showed that differences between subjects and the occurrence of flexion-relaxation had a major impact on the effect of the two devices. Therefore, subject-to-subject variation should be included in the design of the devices, to prevent working against the device for some subjects at deep flexion positions. However, we have to keep in mind that participants were not allowed to bend their knees. In daily life, they probably will do so when approaching their flexion limit, which would avoid the above-mentioned problem.

Although it is expected that, in absence of flexion relaxation, the device has positive effects on back loading, effects cannot be generalized to dynamical lifting. During manual material handling, dynamic lifts with loads around 10–15 kg are no exception. Required moments for these tasks are reported to be as high as 200–300 Nm (Faber et al., 2009; Faber et al., 2011). Because the

support of the current (and other) non-actuated exoskeletons is not dependent on the required moment but solely a function of angle, the relative contribution of the exoskeleton to the required moment may drop to less than 10%. With the large hysteresis present in the device during dynamical movement, support will be even further reduced in the upward phase of a lift, so that the practical value for dynamical lifting tasks of devices with the present support magnitude can be questioned. Another limitation is that results were solely based on male participants as the current version of the device cannot be used by women. As the focus of the study was on the low back, effects around the knee joint were not considered. It might be that loading around the knee was increased as an effect of the EXO. However, these effects are expected to be limited (de Looze et al., 2016). Contact points of the exoskeleton with the participant's' bodies may have differed slightly between participants, which might be a random error. While different sizes of the exoskeleton were available, these were not used as this didn't improve the fit.

Some problems related to functional tasks like walking and tasks involving hip flexion are still present in the exoskeleton tested (Baltrusch et al., 2018). For instance, it was found that metabolic energy expenditure in walking increased by 14% with an exoskeleton (Baltrusch et al., submitted for publication). Some simple adjustments (i.e. switching on/off the support) might already improve the performance of these important tasks and this will be needed to improve user acceptance (Baltrusch et al., 2018).

In conclusion, L5/S1 moments generated by the subjects were significantly reduced at all hand heights and differed between the two exoskeleton types, with LOW being most effective at 50%, 25% and 0% and HIGH at 100% and 75% hand height. This reduction was on average 20%. The device allowed for substantial lumbar bending so that a number of participants did show the flexion-relaxation phenomenon, in which passive tissues are being stretched and generate a major part of the required extension moment. In addition, in absence of back-muscle activity without the device and with increased abdominal activity at 25% and 0% hand height with the device observed in some subjects, compres-

sion forces at the spine are expected to be increased as a result of the user resisting the torque generated by the exoskeleton. Future exoskeletons may need to take these inter-individual variations into account in their design, to prevent generating a supporting torque that together with the passive moment exceeds the net required torque. This study showed the benefit of exoskeletons with different angle-torque relationships, so that workers can make a choice based on the characteristics of task they are exposed to.

Conflict of interest statement

The authors state that there is no conflict of interest to report.

Acknowledgments

References

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

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

Baltrusch, S.J., van Dieën, J.H., Bennekom, C.A.M., Houdijk, H., 2018. The effect of a passive trunk exoskeleton on functional performance in healthy individuals. Appl. Ergon. 72, 94–106.

Baltrusch, S.J., van Dieën, J.H., Bruijn, S.M., Koopman, A.S., Bennekom, C.A.M., Houdijk, H., submitted for publication. The effect of a passive trunk exoskeleton on metabolic costs during lifting and walking.

Barrett, A., Fathallah, F., Year evaluation of four weight transfer devices for reducing loads on lower back during agricultural stoop labor. In: ASAE Annual International Meeting. Sacramento, CA.

Bosch, T., Eck, J., Knitel, K., 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., Croce, U.D., Leardini, A., 1995. Position and orientation in space of bones during movement: anatomical frame definition and determination. Clin Biomech. (Bristol, Avon) 10, 171–178.

Coenen, P., Gouttebarge, V., van der Burght, A.S., van Dieen, J.H., Frings-Dresen, M.H., van der Beek, A.J., Burdorf, A., 2014. 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.

Coenen, P., Kingma, I., Boot, C.R., Twisk, J.W., Bongers, P.M., van Dieen, J.H., 2013. Cumulative low back load at work as a risk factor of low back pain: a prospective cohort study. J. Occup. Rehabil. 23, 11–18.

da Costa, B.R., Vieira, E.R., 2010. Risk factors for work-related musculoskeletal disorders: A systematic review of recent longitudinal studies. Am. J. Ind. Med. 53, 285–323.

de Looze, M.P., Bosch, T., Krause, F., Stadler, K.S., O'Sullivan, L.W., 2016. Exoskeletons for industrial application and their potential effects on physical work load. Ergonomics 59, 671–681.

Eurofound, 2012. Trends in job quality in Europe. Publications Office of the European Union, Luxembourg.

Faber, G.S., Kingma, I., Kuijer, P.P., van der Molen, H.F., Hoozemans, M.J., Frings-Dresen, M.H., van Dieen, J.H., 2009. Working height, block mass and one- vs. two-handed block handling: the contribution to low back and shoulder loading during masonry work. Ergonomics 52, 1104–1118.

Faber, G.S., Kingma, I., van Dieen, J.H., 2007. The effects of ergonomic interventions on low back moments are attenuated by changes in lifting behaviour. Ergonomics 50, 1377–1391.

Faber, G.S., Kingma, I., van Dieen, J.H., 2011. Effect of initial horizontal object position on peak L5/S1 moments in manual lifting is dependent on task type and familiarity with alternative lifting strategies. Ergonomics 54, 72–81.

Floyd, W.F., Silver, P.H., 1955. The function of the erectores spinae muscles in certain movements and postures in man. J. Physiol. 129, 184–203.

Graham, R.B., Agnew, M.J., Stevenson, J.M., 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, 936–942.

Hartvigsen, J., Hancock, M.J., Kongsted, A., Louw, Q., Ferreira, M.L., Genevay, S., Hoy, D., Karppinen, J., Pransky, G., Sieper, J., Smeets, R.J., Underwood, M., Workin, L.L. B.P.S., 2018. What low back pain is and why we need to pay attention. Lancet 391, 2356–2367.

Hof, A.L., 1992. An explicit expression for the moment in multibody systems. J. Biomech. 25, 1209–1211.

Hoozemans, M.J., Kingma, I., de Vries, W.H., van Dieen, J.H., 2008. Effect of lifting height and load mass on low back loading. Ergonomics 51, 1053–1063.

Kingma, I., Bosch, T., Bruins, L., van Dieen, J.H., 2004. Foot positioning instruction, initial vertical load position and lifting technique: effects on low back loading. Ergonomics 47, 1365–1385.

Kingma, I., deLooze, M.P., Toussaint, H.M., Klijnsma, H.G., Bruijnen, T.B.M., 1996. Validation of a full body 3-D dynamic linked segment model. Hum. Mov. Sci. 15, 833–860.

Kingma, I., Faber, G.S., van Dieën, J.H., 2010. How to lift a box that is too large to fit between the knees. Ergonomics 53, 1228–1238.

Kingma, I., Faber, G.S., van Dieën, J.H., 2015. Supporting the upper body with the hand on the thigh reduces back loading during lifting. J. Biomech. 49, 881–889.

Kuiper, J.I., Burdorf, A., Frings-Dresen, M.H., Kuijer, P.P., Spreeuwers, D., Lotters, F.J., Miedema, H.S., 2005. Assessing the work-relatedness of nonspecific low-back pain. Scand. J. Work Environ. Health 31, 237–243.

Lotz, C.A., Agnew, M.J., Godwin, A.A., Stevenson, J.M., 2009. The effect of an on-body personal lift assist device (PLAD) on fatigue during a repetitive lifting task. J. Electromyogr. Kinesiol. 19, 331–340.

Marras, W.S., Granata, K.P., Davis, K.G., Allread, W.G., Jorgensen, M.J., 1999. Effects of box features on spine loading during warehouse order selecting. Ergonomics 42, 980–996.

McGill, S.M., 1991. Electromyographic activity of the abdominal and low back musculature during the generation of isometric and dynamic axial trunk torque: implications for lumbar mechanics. J. Orthop. Res. 9, 91–103.

Norman, R., Wells, R., Neumann, P., Frank, J., Shannon, H., Kerr, M., 1998. A comparison of peak vs cumulative physical work exposure risk factors for the reporting of low back pain in the automotive industry. Clin. Biomech. (Bristol, Avon) 13, 561–573.

Potvin, J.R., Norman, R.W., McGill, S.M., 1996. Mechanically corrected EMG for the continuous estimation of erector spinae muscle loading during repetitive lifting. Eur. J. Appl. Physiol. 74, 119–132.

Thiese, M.S., Hegmann, K.T., Wood, E.M., Garg, A., Moore, J.S., Kapellusch, J., Foster, J., Ott, U., 2014. Prevalence of low back pain by anatomic location and intensity in an occupational population. BMC Musculoskeletal Disorders 15, 283.

Ulrey, B.L., Fathallah, F.A., 2013. Effect of a personal weight transfer device on muscle activities and joint flexions in the stooped posture. J. Electromyogr. Kinesiol. 23, 195–205.

van Dieen, J.H., Kingma, I., 2005. Effects of antagonistic co-contraction on differences between electromyography based and optimization based estimates of spinal forces. Ergonomics 48, 411–426.