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Biomechanical evaluation of a new passive back support exoskeleton

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ABSTRACT

The number one cause of disability in the world is low-back pain, with mechanical loading as one of the major risk factors. To reduce mechanical loading, exoskeletons have been introduced in the workplace. Substantial reductions in back muscle activity were found when using the exoskeleton during static bending and manual materials handling. However, most exoskeletons only have one joint at hip level, resulting in loss of range of motion and shifting of the exoskeleton relative to the body. To address these issues, a new exoskeleton design has been developed and tested.

The present study investigated the effect of the SPEXOR passive exoskeleton on compression forces, moments, muscle activity and kinematics during static bending at six hand heights and during lifting of a box of 10 kg from around ankle height using three techniques: Free, Squat and Stoop.

For static bending, the exoskeleton reduced the compression force by 13–21% depending on bending angle. Another effect of the exoskeleton was that participants substantially reduced lumbar flexion. While lifting, the exoskeleton reduced the peak compression force, on average, by 14%. Lifting technique did not modify the effect of the exoskeleton such that the reduction in compression force was similar.

In conclusion, substantial reductions in compression forces were found as a result of the support generated by the exoskeleton and changes in behavior when wearing the exoskeleton. For static bending, lumbar flexion was reduced with the exoskeleton, indicating reduced passive tissue strain. In addition, the reduced peak compression force could reduce the risk of compression induced tissue failure during lifting.

1. Introduction

Low-back pain (LBP) is the number one cause of disability in the world (Hoy et al., 2014). Literature supports that (work-related) mechanical spine loading is an important risk factor with moderate to strong evidence (Bakker et al., 2009; Coenen et al., 2014; Coenen et al., 2013; da Costa and Vieira, 2010; Griffith et al., 2012; Hoogendoorn et al., 1999; Kuiper et al., 2005; Norman et al., 1998). Despite ongoing mechanization and automation, still 30% of workers are required to lift heavy loads at least a quarter of the work time (Eurofound, 2012), probably because many tasks require the mobility and flexibility of the human (e.g. order pickers, luggage handlers, etc.). As lifting aids are often not used due to their constraints (Baltrusch et al., 2020), focus has shifted towards body worn devices that directly support the user's back. These so-called back support exoskeletons (for reviews see Toxiri et al. (2019) and de Looze et al. (2016)) have been developed to reduce mechanical spine loading. Spine loading is mainly due to muscle forces, needed to counteract the moment at the lower back, induced by gravitational forces on the upper body and handled load. In passive exoskeletons, spring-like components are used to generate an extension moment while bending forward. By applying this supporting moment to the user, the moment, and thereby the force to be generated by the back muscles, is reduced.

Although several passive exoskeletons have shown benefits during static bending in which back muscle activity was reduced by 10–40% (Bosch et al., 2016; Kobayashi and Nozaki, 2008; Koopman et al., 2019b; Ulrey and Fathallah, 2013a, b), effects during lifting are smaller (Koopman et al., 2019a) or similar (Alemi

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et al., 2019). A major drawback of most current exoskeletons (except soft exoskeletons such as (Abdoli-e et al., 2006; Imamura et al., 2014; Inose et al., 2017)) is that they have a single joint at the hip, thereby neglecting partially independent hip and lumbar flexion in humans. As a result, users lose range of motion (Näf et al., 2018), experience unwanted forces in tasks involving hip flexion, e.g., walking and squatting, and shifting of the exoskeleton relative to the body (Baltrusch et al., 2018).

To address these issues, a new exoskeleton design has been developed within the SPEXOR consortium (Babič et al., 2017; Näf et al., 2018). This new design includes separation of hip and lumbar flexion, improved fitting through the use of misalignment compensation mechanisms (Näf et al., 2019), and moment support up to 50 Nm (Näf et al., 2018). The aim of the current study was to biomechanically evaluate this exoskeleton during both static bending and lifting. To do so, compression forces, back and abdominal muscle activity, L5-S1 moments and kinematics were determined in these tasks with and without the device. We hypothesized that, during both static bending and lifting, L5-S1 compression forces would be reduced.

2. Methods

2.1. Exoskeleton

Details about the rationale and design of the tested SPEXOR passive back exoskeleton (Fig. 1) have been reported by Näf et al. (2018). In short, the weight of the exoskeleton is 6.7 kg. It consists of a spine, a pelvis and two hip/thigh modules. The spine module runs along the back and consists of 3 round carbon fiber beams each with a 4.7 mm diameter. At the top, these beams are connected to a back plate, which has a rotational and translational degree of freedom relative to the beams, to account for axial rotation and elongation of the back while bending. During bending, energy is stored in the beams and released during extension, with moments theoretically ranging up to 50 Nm in full flexion. At the pelvis, the beams are connected to a base, which is connected to two overlapping stiff carbon frames, allowing to fit a wide range of pelvic widths to the hip/thigh module, by manually changing the length of the overlap. This hip/thigh modules includes a spring-loaded joint (MACCEPA 2.0) (Vanderborght et al., 2011), augmented with misalignment compensation mechanisms, to account for motion between user and exoskeleton (Näf et al., 2018). Three additional misalignment compensation joints are included to improve the fitting and to reduce shifting of the device. Torque output is tunable by changing the springs' pretension, and ranges from 10 to 30 Nm for each side. To account for the length changes during hip abduction, the module also includes a linear slider along the leg.

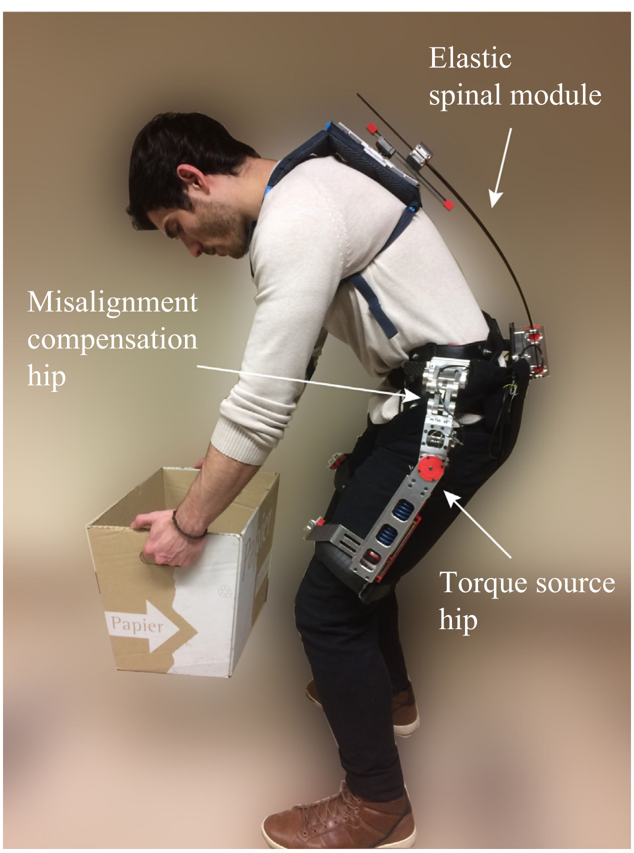

Fig. 1. The SPEXOR passive back support exoskeleton uses a combination of elastic beams and torque generators (MACCEPA) to unload the lower back. Misalignment compensation mechanisms at the hip and the back work towards minimizing discomfort and relative movement. Adapted from "Passive Back Support Exoskeleton Improves Range of Motion Using Flexible Beams," by M. Näf et al., 2018, *Frontiers in Robotics and AI, 5.*

2.2. Subjects and experimental procedures

Following ethical approval of the study (VUmc, Amsterdam, The Netherlands, NL57404.029.16), ten healthy male luggage handlers (mean ± std, age: 46.4 ± 8.7 years, mass: 83.6 ± 16.2 kg, height: 1. 75 ± 0.07 m) from the Dutch airline company KLM, participated in the study after providing written informed consent. First, the exoskeleton was fitted to the participant and ten minutes were given to get familiarized with the device. Subsequently, maximum voluntary contractions were obtained for the back and abdominal muscles in supine posture (McGill, 1991). After the marker clusters were placed on the participants 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.

2.3. Static bending

In the experiment, participants were first asked to bend forward, with the arms vertically downwards. This was done with their hands at six predetermined heights: 100% (standing upright), 95%, 80%, 60%, 20% and 0% (touching the floor) and to hold this position for around five seconds. Participants were instructed to keep their knees straight without locking the knee joint in extension, which was checked visually. Slight knee flexion was allowed if participants were unable to reach the floor with extended knees.

2.4. Lifting

Subsequently, participants were asked to lift a box of 10 kg with the handles 10 cm above ankle height using three different techniques: a Squat lift (flexing the knee joints with the trunk as upright as possible), a Stoop lift (bending the trunk with extended knees) and a Free lift (the participant could freely chose his preferred lifting technique (Fig. 2)). For each technique, three repetitions were performed, with the order of the techniques being randomized over participants. A lifting cycle consisted of picking up the box, returning to an upright posture while holding the box, replacing the box and returning to upright stance once more.

All tasks were performed WITH and WITHOUT exoskeleton, with half of the subjects starting WITH and the other half starting WITHOUT. Because of all the measurement equipment, putting on or taking off the exoskeleton took around 10 min, ensuring suffi-

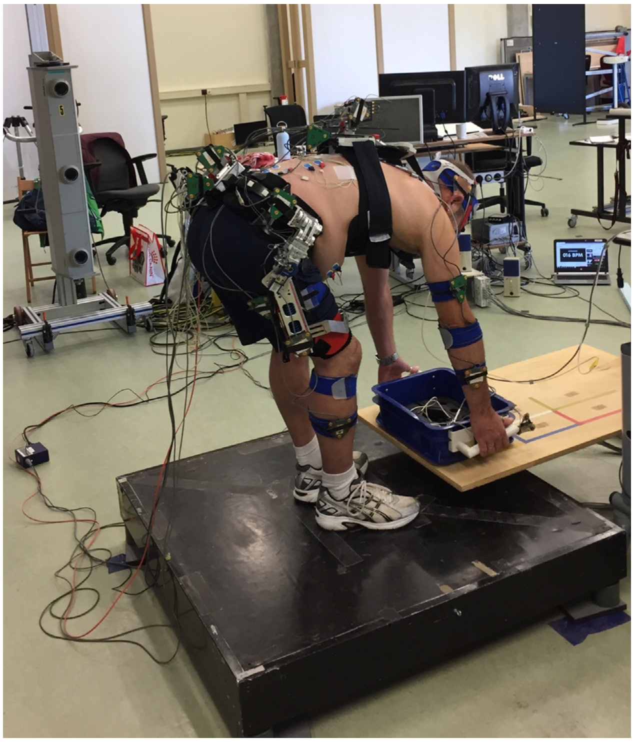

Fig. 2. Picture showing the experimental setup during lifting with the exoskeleton.

cient rest between tasks. In a workplace environment this can be done much faster.

2.5. Instrumentation and data pre-processing

A single custom-made 1.0 $\times$ 1.0 m force plate was used to measure ground reaction forces at 200 Hz. Using an opto-electronic 3D movement registration system (Certus, Optotrak, Northern Digital Inc.), kinematics of the right side of the body were collected at a sample rate of 50 Hz. LED cluster markers were attached to body segments (right foot with lower leg (modeled as one segment), right upper leg, pelvis, trunk (T10), head, right upper arm and right forearm with hand). In addition, marker clusters were attached to relevant parts of the exoskeleton. 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 skin to capture trunk muscles (Rectus Abdominis (RA), External Oblique (EO), Iliocostalis (IL), Longissimus thoracis (TL) and Longissimus pars lumborum (LL); see Kingma et al. (2010)) after abrasion and cleaning with alcohol. EMG data were amplified (Porti-17TM, TMS, Enschede, The Netherlands) and A–D converted (22 bits at 1000 Hz). EMG data were stored synchronized to Optotrak and force plate data using a pulse generated at the instant the recording of the kinematics and kinetics started.

2.6. Data analysis

Data were low-pass filtered using a bi-directional 2nd order Butterworth filter at a cut-off frequency of 5 Hz and 10 Hz, respectively for marker and force plate data. Total L5-S1 flexio-extension moments (M_{L5S1_total}), generated by subject plus exoskeleton were calculated based on the GRF and kinematics, using a bottom-up inverse dynamics model (Kingma et al., 1996). The moment generated by the subject ($M_{L5S1\_subject}$) was calculated by subtracting the moment generated by the device ($M_{L5S1\_exoskeleton}$) from M_{L5S1_total}. $M_{L5S1\_exoskeleton}$ is only dependent on the bending of the beam, and was therefore based on the relation between the bending angle and torque for these beams, which had been characterized during a benchmark bending test previously (Näf et al., 2018). During our measurements, the bending of the beams was measured using the orientation difference of the structures at the trunk and at the pelvis to which the beams were attached. Off-line, EMG signals were band-pass filtered between 20 and 600 Hz, 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 used as input to an EMG driven muscle model. The model has been described in more detail previously (van Dieën, 1997; van Dieën and Kingma, 2005), and consists of 90 muscle slips crossing the L5-S1 joint (Bogduk et al., 1992; McGill, 1996). A best fit between net moments and muscle moments was obtained by optimizing three values for each participant: the gain, i.e. maximum muscle stress, the position of the passive length-tension curve relative to the muscle optimum length, and a scaling factor for the passive length-tension curve. This optimization was performed separately for static and dynamic conditions using data from conditions without exoskeleton only. The optimized values were used in the with exoskeleton condition, without optimizing them again. Finally, to obtain compression forces at the L5-S1 intervertebral joint, muscle forces and net reaction forces were summed after projecting them on the L5-S1 axes system.

2.7. Statistics

Outcome variables were average values (for static tasks) and peak values (for lifting) for L5-S1 compression forces, moments (M_{L5S1_total} and $M_{L5S1\_subject}$), lumbar flexion, trunk angular velocity and lumbar back (averaged over IL and LL) and abdominal (averaged over RA and EO) muscle activity. For lifting, a two-way repeated measures ANOVA was conducted with device and lifting technique as within subject factors. For static bending, a two-way repeated measures ANOVA was conducted with device and height as within subject factors. Device effects were further explored using Bonferroni post-hoc tests, to test if these effects were significant for each technique (in lifting) and each height (in static postures). A significance level of $p < 0.05$ was used. For not normally distributed data, a Wilcoxon signed rank test was used to detect significant differences between WITH and WITHOUT.

3. Results

The fit between ML5S1_subject and the EMG driven model moment, over all conditions and subjects, during lifting was acceptable with correlations (R2) ranging from 0.79 to 0.91, and root mean squared differences ranging from 19.7 to 36.6 Nm (7–13% of highest peak subject moment, averaged over participants). For static bending results were similar, with correlations (R2) ranging from 0.76 to 0.95, and root mean squared differences ranging from 10.0 to 20.4 Nm (8–16% of the highest subject moment, averaged over participants).

3.1. Static bending

Substantial reductions in L5-S1 compression forces were found when wearing the exoskeleton ($p = 0.001$; Table 1, Fig. 3). The effect of the exoskeleton increased with lower hand heights, indicated by a significant interaction effect ($p < 0.001$). As expected, the exoskeleton did not affect compression in upright standing (100% hand height), while compression reduced by 360 ± 104.7 N (13 ± 4%) at 95% hand height up to 798 ± 167 N (21 ± 4%) at 0% hand

Table 1
P-values and effect sizes of the repeated measures ANOVA's for static bending and lifting, with Exoskeleton condition (WITHOUT and WITH) and Height condition (six levels) or Technique condition (Free, Squat and Stoop) and their interactions. Pairwise comparisons were performed for variables with a significant interaction effect with the factor device. Significant ($p < 0.05$) results were indicated in bold. Note that muscle activities in the static tasks were not normally distributed, so no ANOVA was performed for these two variables. Instead, a Wilcoxon sign rank test was performed, results are shown in the figures and text.

	Static			Dynamic		
	Main effect Exoskeleton *p (η2)*	Main effect Height *p (η2)*	Interaction Exoskeleton * Height *p (η2)*	Main effect Exoskeleton *p (η2)*	Main effect Technique *p (η2)*	Interaction Exoskeleton * Technique *p (η2)*
L5S1 Compression	**.001 (.783)**	**<.001(.893)**	**<.001 (.523)**	**.003 (.774)**	**.015 (.450)**	.216 (.205)
M_{L5S1_total}	**.004 (.662)**	**<.001 (.908)**	.212 (.173)	**.029 (.515)**	**<.001(.807**	.799 (.018)
Lumbar back muscle activity	–	–	–	**.003 (.736)**	.135 (.261)	.378 (.123)
$M_{L5S1\_subject}$	**<.001 (.930)**	**<.001 (.872)**	**<.001 (.757)**	**<.001 (.866)**	**<.001 (.855)**	.573 (.075)
Lumbar flexion	**.003 (.701)**	**<.001 (.944)**	**.007 (.422)**	.778 (.012)	**.001 (.637)**	.086 (.345)
Trunk angle velocity	–	–	–	**.009 (.651)**	**.028 (.446)**	.434 (.106)
Abdominal muscle activity	–	–	–	.534 (.058)	.241 (.189)	.467 (.098)

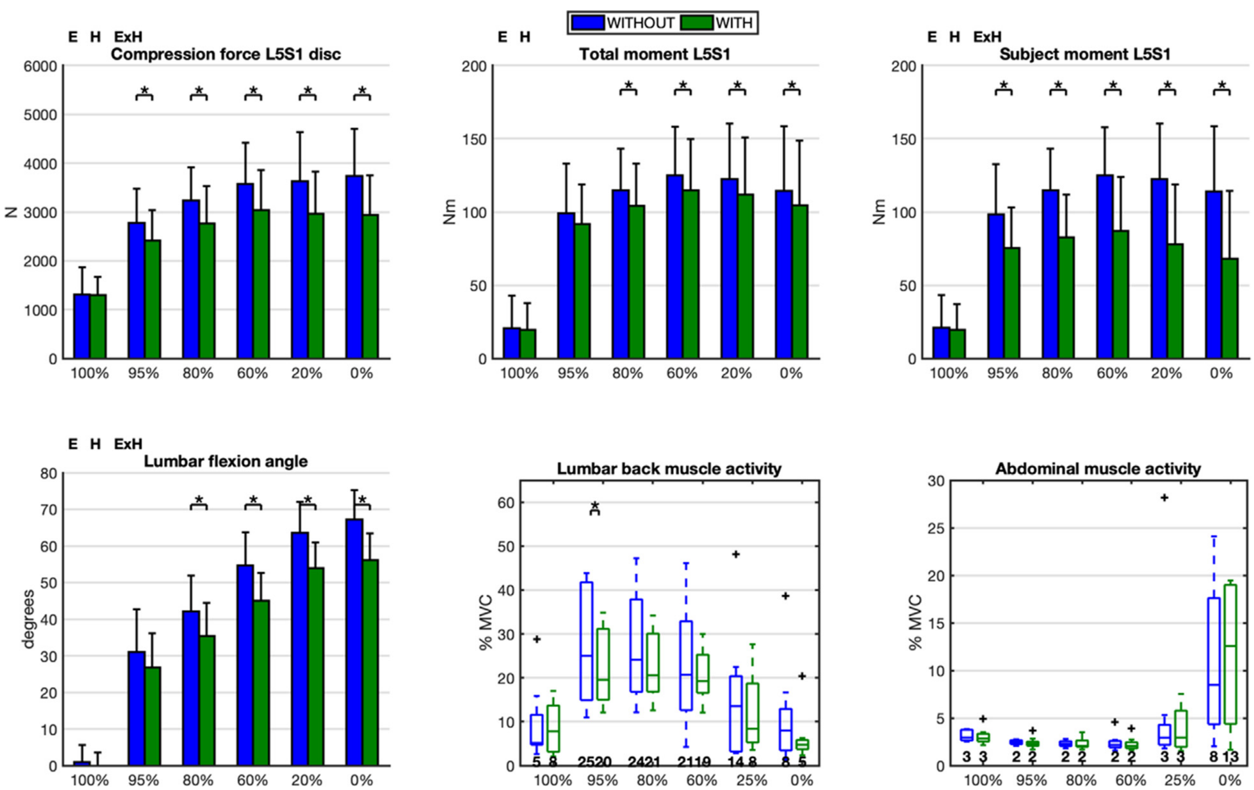

Fig. 3. L5S1 compression, moments (ML5S1_total and ML5S1_subject), lumbar flexion angle and back and abdominal muscle activity. Main effects of Exoskeleton and Height are indicated with E and H, respectively. An interaction effect of Exoskeleton*Height is indicated with ExH. Bonferroni corrected post hoc *t*-test's were performed, horizontal bars indicate a significant ($p < 0.05$) difference between the two bars. No ANOVA was performed for the back and abdominal muscle activity as the data was not normally distributed. Therefore, boxplots are shown. In the boxplots, the central mark is the median (of which the value is shown at the bottom of the graph), the edges of the box are the 25th and 75th percentiles, the whiskers extend to the most extreme data points to be not outliers (<2.7 SD), and the outliers (+) are plotted individually. Wilcoxon sign rank tests were performed, horizontal bars indicate a significant ($p < 0.05$) difference between the two bars.

height (Fig. 3). The supporting moment of the exoskeleton at 95% was 16 ± 1.7 Nm, and increased up to 37 ± 1.1 Nm at 0% hand height. A small but significant average reduction of 8.1 ± 2.0 Nm (8.1 ± 2%) was found in the $L5S1_{\_total}$ moment, which was due to a small decrease of the inclination angle in the WITH condition. The $L5S1_{\_subject}$ moment was reduced more than the compression force, ranging from 23 ± 5.6 Nm (23 ± 6%) at 95% hand height to 46 ± 3.5 Nm (40 ± 3%) at 0% hand height. Despite these substantial reductions in L5-S1 moments and compression forces, back muscle activity was not always significantly reduced. Only for the 95% hand height, a significant reduction of the median of 5.5% MVC (22%) was found ($p = 0.028$). For both 80% and 0% hand height, reductions of 3.6% MVC (15%) and 3.3% MVC (42%) did not reach significance ($p = 0.086$ & $p = 0.066$). Averaged over all hand heights, a significant reduction of lumbar flexion 7.3 ± 1.7 degrees (17 ± 4%) was found when wearing the exoskeleton. For the 0% hand height (hands at floor level) substantial abdominal muscle activity was found, but no significant difference between WITHOUT and WITH was present.

3.2. Lifting

On average, the exoskeleton significantly reduced peak L5-S1 compression forces by 972 ± 216 N (14 ± 3%) with no interaction effect of device condition with lifting technique ($p = 0.216$, Table 1, Fig. 4). As in the static tasks, the moment reduction by the

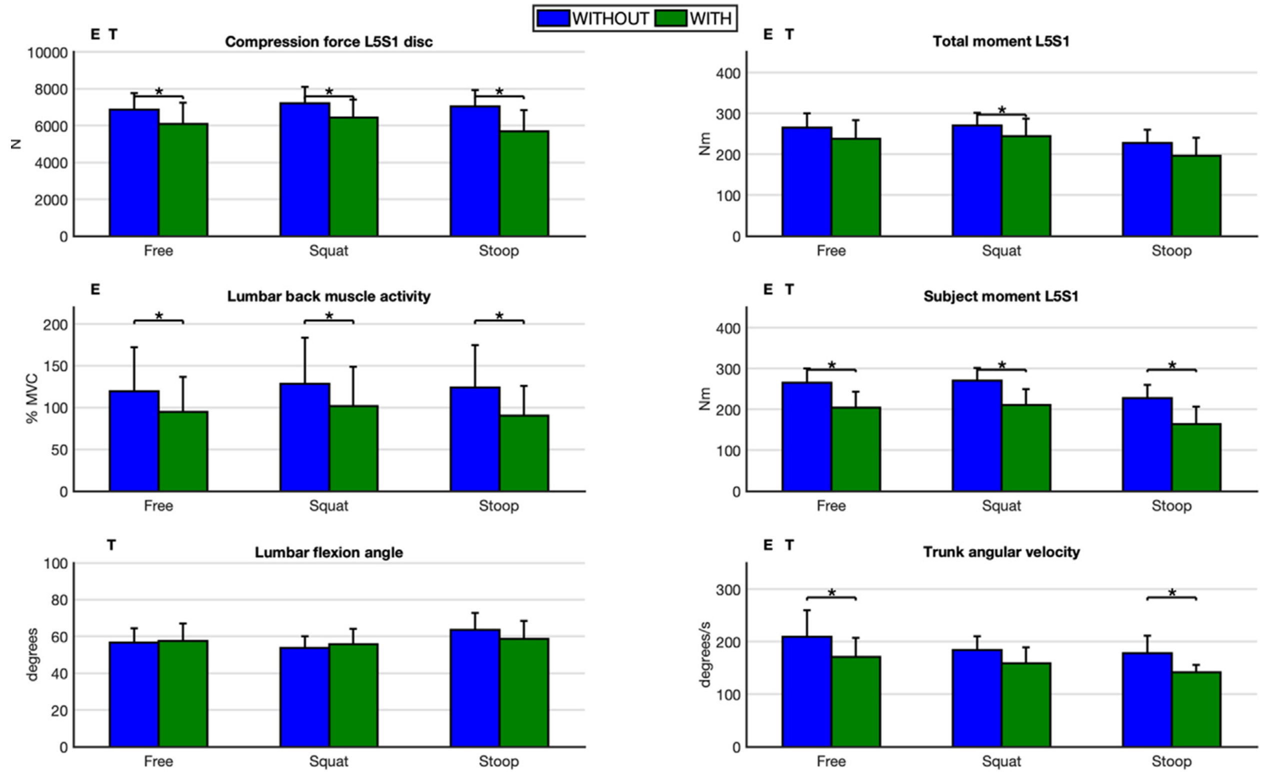

Fig. 4. Peak L5-S1 compression, peak moments (ML5S1_total and ML5S1_subject), peak lumbar flexion angle, peak back muscle activity and peak trunk angular velocity. Main effects of Exoskeleton and Technique are indicated with E and T, respectively. An Interaction effect of Exoskeleton*Technique is indicated with ExT. Bonferroni corrected post hoc *t*-test's were performed, horizontal bars indicate a significant (p < 0.05) difference between the two bars.

exoskeleton was more substantial, by 61.3 ± 9.1 Nm (23 ± 3%) at the instant of peak compression, than the reduction in peak compression force. The moment support of the exoskeleton at this instant was 33.4 ± 1.1 Nm compared to 40.8 ± 1.1 Nm maximally, which shows that the instant of peak loading and peak support do not coincide. The fact that the $L5S1_{\_subject}$ reduction was larger than the support of the EXO, was largely due to a significantly lower peak trunk angular velocity with 33 ± 9 degrees/s (17 ± 5%) while wearing the exoskeleton. Peak activity of the lumbar back muscles was 28 ± 6.3% MVC (22 ± 5%) lower when using the exoskeleton while average peak lumbar flexion was unaffected by the exoskeleton (p = 0.778). While most outcome variables showed a main effect of lifting technique, with peak compression forces showing only significantly larger values for Squat than in Stoop. No significant interaction between lifting technique and exoskeleton condition was found.

4. Discussion

The present study demonstrated reduced back loading when using the new passive spinal exoskeleton due to moment support provided and changes in lifting behavior. In line with our hypothesis, the exoskeleton reduced L5-S1 spine compression forces by 13–21%, depending on hand height, for static bending and by 14% while lifting a box of 10 kg from ankle height using various lifting techniques. The largest reduction, of 21%, was found during static bending at 0% hand height. During lifting, peak back muscle activity was substantially reduced by 22% while no change in peak lumbar flexion angle was present. In contrast, during static bending peak lumbar flexion was reduced by around 7 degrees, while wearing the exoskeleton, leading to insignificant reductions in back muscle activity due to a shift from passive to active force generation, which will be further discussed below. Despite changes in most outcomes with changes in lifting technique, the effect of the exoskeleton was similar between lifting techniques, showing the versatility of the device in the sagittal plane.

4.1. Static bending

For static bending an unexpected significant reduction in the $L5S1_{\_total}$ moment was found, which was due to slightly smaller trunk inclination angles in the WITH condition. This slightly biased our experiment. However, even taking the decreased $L5S1_{\_total}$ moment into account, the exoskeleton still reduced the $L5S1_{\_subject}$ moment by around 16-37 Nm. The exoskeleton had a significant effect on bending behavior as it systematically reduced lumbar flexion. As a result, muscles operated more leftward on their force-length relation and therefore the portion of active force generation was higher. This might explain why reductions in back muscle activity did not reach significance at 80% and 0% hand height. For the 60% and 20% hand heights between subject differences were even more pronounced because some participants were close to flexion relaxation, so that minor changes in lumbar angles had large effects on lumbar muscle activation. Although at 0% abdominal muscle activity was higher compared to the other hand heights, no significant differences were found between WITH and WITHOUT. Again, larger between subject variability in flexibility, but also in body weight and height, could play a role here.

Due to the support of the exoskeleton, substantial reductions in the L5-S1 compression force were found. The highest compression value during static bending was still under 4000 N and this probably is not high enough to cause direct damage in most people (Brinckmann et al., 1988, 1989; Jäger, 2018). However, a substantial reduction in cumulative loading has also shown to be important to reduce the risk of injury (Coenen et al., 2013). The exoskeleton also caused a reduction in lumbar flexion so that it is also likely that the risk of LBP (Griffith et al., 2012) possibly due to passive tissue strain injury (Solomonow et al., 2003) would be reduced.

During static bending, the support of the exoskeleton ranged from 17 to 37 Nm, depending on bending angle, which is somewhat higher than other passive devices that maximally provided around 25 Nm (Abdoli-Eramaki et al., 2007; Koopman et al., 2019b). This contrasts with more consistent reductions in EMG in previous studies (Bosch et al., 2016; de Looze et al., 2016; Graham et al., 2009; Koopman et al., 2019b; Ulrey and Fathallah, 2013a) compared to the current study. Most likely, this is related to the above-mentioned reductions in lumbar flexion when using the current device. Most previous studies did not report lumbar flexion. Differences could also, in part, be due to slight differences in the experimental setup (task constraints, mass, etc.). The maximum of 21% reduction in L5-S1 compression force was substantially higher compared to the only study that also quantified reductions in terms of compression forces with 13% (Ulrey and Fathallah, 2013b).

4.2. Lifting

Lifting technique did not modify the effect of the exoskeleton such that the reduction in compression force was similar between the three different lifting techniques. Moreover, lumbar flexion was unaffected by the exoskeleton in each of these techniques, showing its versatility. However, the exoskeleton did reduce the lifting speed. As a result, $L5S1_{\_total}$ moment was reduced in the WITH condition so that reductions in $L5S1_{\_subject}$ moment were more pronounced than what would be expected based on the actual support of the device. It can be questioned whether this possibly forced reduction in lifting speed is beneficial. Users might feel forced to lift slower and might not be willing to use the device because of this. In addition, it will reduce the productivity of the workers, or, if the same number of lifts is performed, it might increase cumulative loading. In contrast to the static tasks, back muscle activity was significantly reduced for all lifting techniques (20–27%) when using the exoskeleton. This was likely because during lifting no change in peak lumbar flexion was present between WITHOUT and WITH.

Although it is still unclear how much reduction is needed to have a relevant effect on occupational back loading, the average compression reduction of 14% found with the SPEXOR passive spinal exoskeleton is promising, especially because the reduction was independent of lifting style. To translate this to a more practical interpretation, this reduction of 1000 N is approximately equal to the difference between an ankle versus knee height lift with a box of 10 kg (Koopman et al., 2019a). Based on compressive strength data of cadaveric specimens, a 14% lower compression force will substantially reduce the population at risk (Brinckmann et al., 1989; Jäger, 2018). The reductions for back muscle EMG and compression were larger or similar compared to other passive exoskeletons used in similar task conditions (Alemi et al., 2019; Koopman et al., 2019a). Despite the added mass, the exoskeleton reduced back muscle activity, which could potentially prevent or slow down local and global fatigue development. The latter was underscored by a study of Baltrusch et al. (2019), in which oxygen consumption was found to be reduced by approximately 18%, while wearing the same passive spinal exoskeleton. However during walking, it is likely that the weight of the exoskeleton could lead to a slight increase in oxygen consumption.

Peak L5-S1 compression force during lifting a 10 kg box from ankle height were somewhat higher than expected based on other studies (Bazrgari et al., 2008; Kingma et al., 2016; Koopman et al., 2019c; Marras and Davis, 1998), probably mainly due to the anthropometric characteristics of the professionals included in this study.

4.3. Limiations

Potential sources of bias and limitations of this study should be considered. Errors in spinal forces estimated by our EMG-driven model may arise due to factors such as cross-talk, bad representation of deep and wide muscles, EMG normalization, ignoring spine translations and considerations of L5-S1 moments only (Arjmand et al., 2009; DeLuca and Merletti, 1988; Gagnon et al., 2011; Staudenmann et al., 2005; Stokes et al., 2003). However, these sources of error are not likely to affect our comparison between conditions, as these sources of error are not likely to vary strongly between the conditions. In addition, as it is unclear how the mass of the exoskeleton (6.7 kg) is distributed over the body (i.e. the portion of the mass carried by the pelvis), we neglected this effect of the exoskeleton in the inverse dynamic analysis. However, because we applied bottom-up inverse dynamics the weight and center of mass of the exoskeleton is taken into account through the ground reaction forces. Only the added mass in the pelvis is ignored, but we are confident that this effect is small due to the small moment arm relative the L5-S1. 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 exoskeleton. However, these effects are expected to be limited (de Looze et al., 2016), as the device does not generate a moment across the knee joint. Another limitation is that in lifting, lumbar flexion varied less between lifting techniques, compared to some previous studies (Kingma et al., 2004; Kingma et al., 2010). This is likely related to the fact that we recruited participants with ample lifting experience, who may have tended to vary less in lifting style according to instruction, compared to participants not involved in professional lifting. Interestingly, in contrast to static bending, a reduction in lumbar flexion was not seen during lifting. Probably the reduced lumbar flexion during static bending is because participants are already close to experiencing flexion-relaxation without the exoskeleton in the lowest hand positions. If subjects would bend the trunk to the same extent in the WITH condition, they might need to increase abdominal muscle activity, to counteract the support of the exoskeleton, which is counterproductive. While in lifting, although similar lumbar flexion can be reached, flexion relaxation is not present because additional moments are needed due to the dynamics and the load lifted. Therefore, subjects would not need to reduce flexion to avoid abdominal activation. No direct comparison between static and dynamic conditions for the same posture could be performed. However, averaged over the three lifting techniques, peak lumbar flexion was around 60 degrees, resulting in 39 Nm support and 950 N compression reduction. This was similar to the static bending at 0% hand height (60 degrees, 37 Nm, and 800 N, respectively). These numbers show that the absolute effect of the exoskeleton was similar independent whether it was during static or dynamic conditions.

Because the support moment of the exoskeleton is translated to the participant via a force of which the point of application is unknown, the effective moment around the L5-S1 joint might not be the same as the reported exoskeleton moment. This might in part explain the larger percentage reduction in the L5-S1 subject moment compared to the L5-S1 compression force.

Only limited time was available to get used to the device, obviously this was not enough to get fully familiarized with the device. Possibly, longer familiarization period might affect the results, for instance with regard to the lifting speed. However, it is unknown how time of familiarization affects lifting speed. Tasks were only performed in the sagittal plane while it is known that workers often perform asymmetrical tasks in the work place. Although the exoskeleton includes features that enables asymmetric movement and the authors believe the exoskeleton would still be beneficial in asymmetric tasks, future studies should determine its effect in such tasks. In addition, we did not directly compare the new features of the current exoskeleton with other one-joint back exoskeletons. Future work should investigate this further.

4.4. Conclusion

In conclusion, the tested passive spinal exoskeleton reduced the L5-S1 compression force by 13–21% and the peak L5-S1 compression force by 14%, for sagittal plane static bending and lifting, respectively. Although, a part of the reduction was due to resulting changes in behavior due to wearing the exoskeleton. The effect of the exoskeleton was independent of lifting technique and shows the versatility needed in various lifting conditions. This suggest wearing the exoskeleton could reduce the risk of low back pain both during sagittal plane lifting and during static forward bending.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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