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Evaluation of a passive back-support exoskeleton in bed-to-chair patient handling tasks

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ABSTRACT

This study assessed the impact of a passive back-support exoskeleton (Laevo V2.5; Laevo, Netherlands) on trunk and hip postures, lower back muscle activity and heart rate during four common patient handling tasks: task 1, lying to sitting; task 2, sitting to standing; task 3, standing to sitting; task 4, bed-to-wheelchair transfer. Eight participants performed these tasks with and without the exoskeleton. Significant reductions (19%; $p < 0.05$) in erector spinae muscle activity were observed during tasks 2 and 4. Moreover, peak bilateral hip flexion angles decreased by up to 29° across tasks, with a notable decrease in median hip flexion angles in three tasks, except for task 3. These findings suggest that the exoskeleton may offer benefits in reducing lower back muscular strain during certain patient transfer tasks, indicating its potential utility in healthcare settings. Further research is needed to fully assess its effectiveness and practicality in improving patient-handling techniques.

KEYWORDS
back-support exoskeleton; patient transfer; hip posture; muscle activity; heart rate

1. Introduction

Musculoskeletal disorders (MSDs) have been a primary concern for workers in the healthcare industry [1]. In 2017, there were about 3 million registered nurses, 0.8 million home health aides and 1.5 million nursing assistants in the USA [2–4]. The healthcare and social assistance sector alone reported 623,000 injury and illness cases in 2021 [5]. Approximately 35% of nurses revealed current pain in the lower back, and 14% of nurses have experienced a MSD injury [6]. Based on data from the Bureau of Labor Statistics (BLS) [7], hospital workers have nearly twice the rate of overexertion injuries when compared to the average rate of overexertion injuries for all industries. Patient handling activities, including lifting, transferring and repositioning patients, are known to be highly associated with the physical risks of MSDs [8–10]. Patient handling often requires high force (overexertion) and awkward body postures (e.g., bending, stooping, twisting and reaching), which could increase the risk of MSDs [11].

Safe patient handling and mobility (SPHM) programs involving mechanical equipment and safety procedures have demonstrated significant reductions in the injury rate of caregivers [12]. Despite successes with these programs, laws requiring SPHM programs in healthcare exist in only 11 of 50 states and no federal legislation has been implemented to mandate the programs [13]. Additionally, certain specialized healthcare settings, such as home healthcare, long-term care facilities, operating rooms and imaging/radiology facilities, present unique challenges to adopting SPHM programs [14]. The current high incidence rates of MSDs among patient care workers show that severe challenges remain [15]; thus, there is a need to improve the existing intervention methods and explore non-traditional assistive technologies, such as exoskeletons.

Industrial exoskeletons have recently gained great attention as an innovative ergonomic intervention for mitigating the risk of MSDs among workers; passive exoskeletons specifically are designed to provide postural support and redistribute the external load on the human body segments [16]. Commercially available exoskeletons have been implemented in several industries, including shipbuilding, automotive, agriculture, construction and aerospace manufacturing sectors [16–19]. Reported findings regarding the effectiveness of back-support exoskeletons in previous studies are inconsistent. Some studies found that back-support exoskeletons reduced the physical demand on the back during repetitive lifting and manual assembly tasks [16,20–22]. However, other studies have shown that passive exoskeletons may adversely affect body postures or increase discomfort, or did not alter the spinal load [16,20,21,23].

Several studies have investigated or discussed the effectiveness of exoskeletons in the context of patient handling [24–29]. One previous study specifically focused on assessing the impact of three passive back-support exoskeletons on the physical demands of caregivers using three different patient transfer methods during simulated bed-to-wheelchair transfers [24]. However, due to the complexity of the patient transferring methods, it was challenging to determine how the exoskeletons affect the different tasks or sub-tasks involved in the bed-to-chair transfer procedure. Importantly, previous studies have revealed that the effectiveness of exoskeletons varies depending on the specific task being performed [16,24]. They found that the effects of the back-support exoskeleton varied with the symmetry or asymmetry conditions of the tasks, and the designs of the exoskeletons. For the heart rate, there have been inconsistent results regarding the impact of the exoskeletons [30,31]. One study showed no significant

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differences in the heart rate with and without the exoskeleton, whereas another study showed significant reduction of the heart rate with the exoskeleton. This emphasizes the need for thorough testing encompassing a broad spectrum of patient handling activities prior to the implementation of exoskeletons as an aid for healthcare workers in performing patient handling tasks.

Given the prevalence of bed-to-chair patient transfer as a common task in patient handling, the primary objective of this study was to investigate the effects of a back-support exoskeleton (Laevo V2.5, small and large sizes, mass 2.90 kg; Laevo, Netherlands) on trunk and hip postures, lower back muscle activity and heart rate during four selected patient handling tasks associated with this procedure. We hypothesized that utilization of the back-support exoskeleton would result in a reduction in lower back muscle activity and heart rate among caregivers [30], potentially influencing their patient-handling techniques when compared to performing the tasks without wearing the exoskeleton.

2. Methods

2.1. Participants

Eight participants (five males and three females) from the university population were recruited for the study. Participant eligibility criteria included the following: absence of musculoskeletal pain in the past 7 days; no current use of medication associated with MSDs; and no physical activity restrictions. The mean (standard deviation) of height and weight of the participants were 172.75 (8.94) cm and 72.13 (12.82) kg, respectively. Along with the eight participants, an additional participant was included in the study as a consistent simulated patient for all data collection. The simulated patient, with a height of 177 cm and a weight of 87 kg, was specifically chosen to provide standardized conditions throughout the experiment. In order to minimize the inconsistent assistance level from the patient, we only recruited a single simulated patient for the entire study. The simulated patient was trained to offer partial assistance to the caregivers through controlled voluntary actions. The simulated patient underwent extensive practice to achieve a steady level of assistance across all caregiver interactions. Prior to their involvement in the study, all participants provided informed consent in accordance with the protocols approved by the Institutional Review Board (HS22-0131).

2.2. Passive back-support exoskeleton

The Laevo V2.5 was used to test the effects of a back-support exoskeleton. The Laevo exoskeleton was selected because it is one of the passive back-support exoskeletons that are most commonly evaluated in various industrial settings, including the healthcare setting [24,27,28]; specifically, the Laevo exoskeleton was favorable during the prone-positioning patient maneuver [27] and was the only exoskeleton investigated in an intention-of-use study for geriatric care work [28]. The Laevo V2.5 includes gas springs to support the back and body anchor points, which are located at the chest, waist and thighs for load distributions. The engagement angle of the exoskeleton was adjusted for each individual participant to ensure their comfort while wearing and performing movements. The engagement angle was specifically designed for Laevo and adjusting this angle alters the magnitude of the torque generated. The maximum support torque provided by this exoskeleton was up to 40 Nm [32].

2.3. Patient handling tasks

A total of four bed-to-chair patient handling tasks were performed with participants wearing and not wearing the back-support exoskeleton (Table 1) [33]. Prior to the primary data collection, participants received detailed instructions on how to perform each patient handling task and had the opportunity to practice until they felt confident in performing the tasks. Standardized material including photographs of the step-by-step method for each patient handling task was consistently provided to each participant. While two tasks directly involved bed-to-chair interactions, the overall procedure was divided into four subtasks to gain deeper insights into the biomechanical characteristics with and without exoskeleton use. Each task was repeated twice and the order of tasks was randomized. The order of with and without exoskeleton use was counterbalanced. During the study, the simulated patient consistently utilized a transfer belt, an assistive device designed to support individuals facing difficulties in independent standing or walking. The belt was made of durable materials like nylon and cotton, and was designed to securely wrap around the individual's waist. The height of the wheelchair was adjusted to 47 cm while the bed height was set at the lowest height (67.3 cm) for tasks 1, 2 and 4. This adjustment ensured that the patient's feet could safely touch the ground during these tasks, enhancing stability and safety throughout the experimental procedures.

2.4. Apparatus and measures

2.4.1. Kinematic data

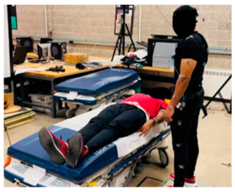
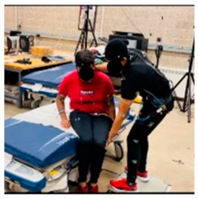
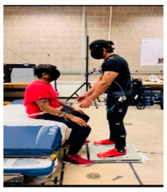
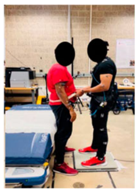
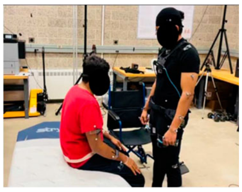
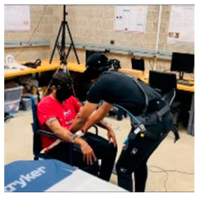
An optical motion capture system with eight cameras (Flex 13; OptiTrack, USA) sampling at a rate of 50 Hz collected kinematic data of the participants. A full body Plug-in Gait marker set, consisting of 39 reflective markers, was attached to anatomical landmarks on the head, torso, arms, hands, pelvis, legs and feet of the participants. The motion capture software Motive version 2.0 was used to capture the three-dimensional (3D) positional data of the markers.

The biomechanical analysis software Visual3D v6 was utilized to compute the joint angles. The 3D joint angles of the trunk and left/right hips were computed between the local coordinate system of the torso and thighs, relative to the pelvis, respectively. The peak (90th percentiles) and the median (50th percentiles) angles of the trunk flexion/extension and left/right hip flexion/extension were extracted from statistical analysis.

2.4.2. Electromyography

The muscle activities of the bilateral erector spinae were collected using a wireless electromyography (EMG) system (Delsys Trigno; Delsys Inc., USA) at 2000 Hz. The Trigno sensors feature a 10-mm electrode distance, coupled with a common mode rejection ratio of 80 dB [34]. These electrodes were attached to the bilateral erector spinae, positioned 3 cm laterally from the L3 spinous process [35]. This particular muscle was chosen as previous studies have shown that the muscle activities of the bilateral erector spinae are significantly affected by various patient handling methods [24,36,37]. Previous studies have also shown that these specific muscles were highly active in patient handling tasks [24,36,37]. The skin

Table 1. Description of patient handling tasks.

Task	Description	Start	End
1	Elevating the patient from a supine position in the bed to an upright sitting position. The participant grasps the patient's back and leg to reposition the patient		
2	Lifting the patient from sitting on the bed to standing. The participant grasps the transfer belt to lift the patient		
3	Repositioning the patient from standing to sitting on the wheelchair. The participant grasps the transfer belt to reposition the patient		
4	Transferring the patient from the bed to a wheelchair. The participant grasps the transfer belt to transfer the patient		

Note: For tasks 1, 2 and 4, the bed height was set at the lowest height (67.3 cm). The simulated patient was trained to partially assist the caregivers by exerting some level of voluntary activities.

preparation and electrode placements were conducted based on the European recommendation for surface EMG [38]. Participants performed a maximum voluntary contraction (MVC), in which participants were asked to maintain an unsupported trunk without resistance in a prone position for 5 s while on a parallel Roman chair [39]. The MVC was repeated three times with a 2-min break between tasks.

The frequency bandwidth for noise rejection of EMG was set at 20–450 Hz. The EMG signal was smoothed using a window size of 0.025 s root mean square (rms) using EMGworks version 3.0. The peak (95th percentile) values for both left erector spinae (LES) and right erector spinae (RES)were extracted through three repetitive tasks of MVCs for each participant. The RMS time series values of each task were then normalized by dividing them by the respective peak MVC value for each muscle (%MVC) [37]. The peak (90th percentile) and median (50th percentile) normalized muscle activity values of each task were obtained [40].

2.4.3. Heart rate

The electrocardiogram (ECG) signals were continuously recorded using a Polar H10 monitor with a Pro Strap (Polar Electro Oy, Finland) during the patient handling task. The signal quality of this wearable heart rate sensor was verified in a previous study [41]. The sensor, equipped with a moistened electrode strap, was securely fastened to the participant's chest area. Following the data collection, the mean heart rate (bpm) was calculated based on the raw ECG signals.

2.5. Statistical analysis

The independent variable was the exoskeleton condition (with and without the back-support exoskeleton). The dependent variables consisted of the peak and median trunk flexion and extension angles, left and right hip flexion angles, peak and median normalized muscle activities of the LES and RES, and mean heart rate. The mixed model approach (generalized linear model [GLM]) was conducted to evaluate the effect of the back-support exoskeleton on the dependent measures. The exoskeleton condition was treated as a fixed effect and the participant was set as a random effect. This analysis was performed for each patient handling task, respectively. The statistical significance (p value) was set as 0.05. The effect sizes based on Cohen's d were computed to assess the practical significance of the measures.

3. Results

3.1. Kinematic data

3.1.1. Trunk postures

As presented in Table 2 and Figure 1, there were no significant differences of trunk flexion and extension angles with

Table 2. Statistical analysis results for peak (90th percentile) and median (50th percentile) trunk flexion and extension angles: *p* values of repeated-measures ANOVA and effect sizes.

Statistic	Variable	Task 1	Task 2	Task 3	Task 4
p	Peak trunk flexion	0.443	0.562	0.919	0.821
	Peak trunk extension	0.109	0.146	0.146	0.951
	Median trunk flexion/extension	0.628	0.855	0.177	0.737
Effect size	Peak trunk flexion	0.174	−0.165	0.028	0.064
	Peak trunk extension	0.616	0.470	0.467	0.021*
	Median trunk flexion/extension	0.153	0.058	0.456	0.107

*$p < 0.05$.
Note: Cohen's *d* was used to assess the effect size. ANOVA = analysis of variance; task 1 = lying to sitting; task 2 = sitting to standing; task 3 = standing to sitting; task 4 = bed-to-wheelchair transfer.

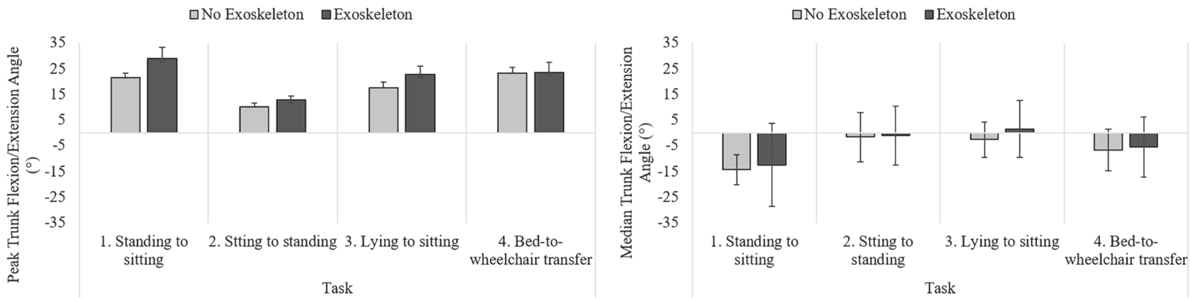


Figure 1. Mean and standard error (whisker) of the peak (90th percentile) and median (50th percentile) trunk flexion(+)/extension(–) angles with and without the back-support exoskeleton for the four different patient handling tasks.

and without the back-support exoskeleton for all patient handling tasks (effect size = −0.165 to 0.616). The trunk extension angle difference with and without the exoskeleton during task 1 showed a medium effect size ($|d| > 0.5$).

3.1.2. Hip postures

As presented in Table 3 and Figure 2, left and right hip flexion angles were significantly affected by the exoskeleton for all patient handling tasks (effect size = −2.087 to −0.072). The exoskeleton significantly reduced the peak bilateral hip flexion angles (up to 29°). The left and right peak hip flexion angle difference with and without the exoskeleton of tasks 1, 2 and 3 showed a large effect size ($|d| > 0.8$).

3.2. Electromyography

The peak (90th percentile) and median (50th percentile) values of the left erector spinae (LES) muscle activity were significantly affected by the exoskeleton during task 2, as presented in Table 4 and Figure 3 (effect size = −0.743 to −0.395). The LES peak muscle activity difference with and without the exoskeleton of task 2 showed a medium effect size ($|d| > 0.5$). The exoskeleton significantly affected the peak and median values of the RES muscle activity during tasks 2 and 4 (effect size = −0.630 to −0.507). The RES peak and median muscle activity difference with and without the exoskeleton during tasks 2 and 4 showed a medium effect size ($|d| > 0.5$). The muscle activities were significantly lower with the exoskeleton compared to no exoskeleton condition (up to 12%MVC).

3.3. Heart rate

As presented in Table 5, there were no significant differences of heart rate with and without the exoskeleton for all patient handling tasks (effect size = −0.295 to 0.297). The heart rate difference with and without the exoskeleton of tasks 1, 2 and 4 showed a small effect size ($|d| > 0.2$). The heart rate varied from 88 to 104 bpm without the exoskeleton and from 92 to 123 bpm with the exoskeleton.

Table 3. Statistical analysis results for left and right peak (90th percentile) and median (50th percentile) hip flexion angles: *p* values of repeated-measures ANOVA and effect sizes.

Statistic	Variable	Task 1	Task 2	Task 3	Task 4
p	Left peak hip flexion	<0.001*	<0.001*	<0.001*	0.0029*
	Left median hip flexion	<0.001*	0.003*	0.818	0.009*
	Right peak hip flexion	<0.001*	<0.001*	<0.001*	0.0014*
	Right median hip flexion	<0.001*	0.007*	0.518	0.002*
Effect size	Left peak hip flexion	−1.926	−1.297	−1.054	−0.579
	Left median hip flexion	−1.576	−0.786	−0.072	−0.622
	Right peak hip flexion	−2.087	−1.287	−1.361	−0.657
	Right median hip flexion	−1.502	−0.817	−0.208	−0.759

*$p < 0.05$.
Note: Cohen's *d* was used to assess the effect size. ANOVA = analysis of variance; task 1 = lying to sitting; task 2 = sitting to standing; task 3 = standing to sitting; task 4 = bed-to-wheelchair transfer.

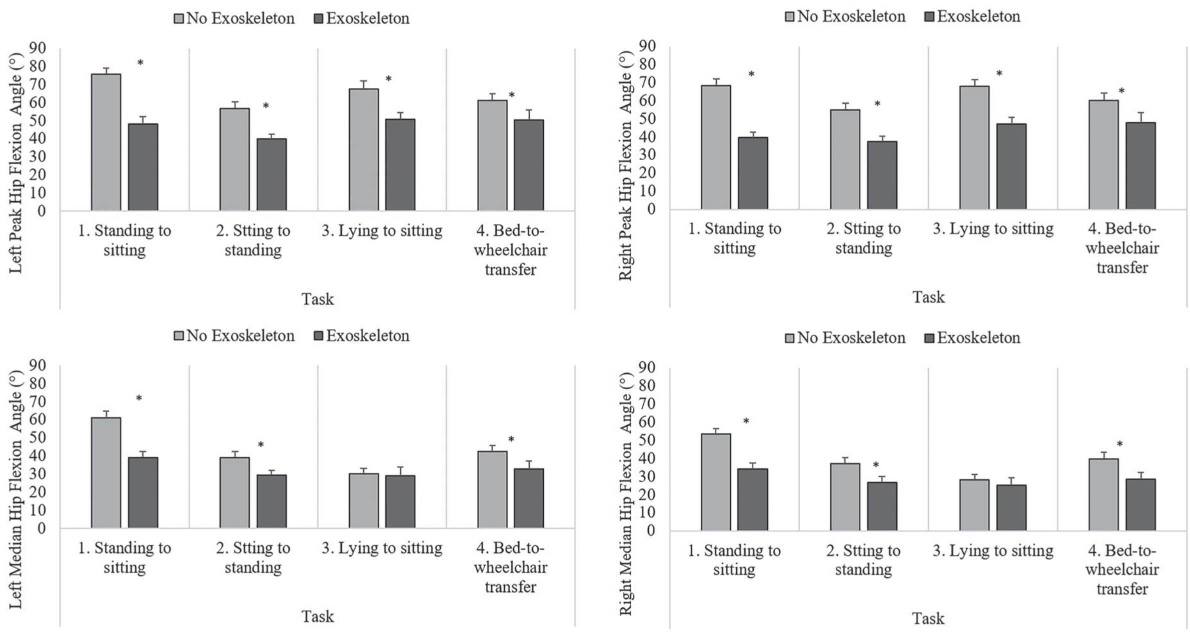


Figure 2. Mean and standard error (whisker) of the left and right peak (90th percentile) and median (50th percentile) hip flexion angles with and without the back-support exoskeleton for the four different patient handling tasks. *Significant difference with and without the exoskeleton ($p < 0.05$).

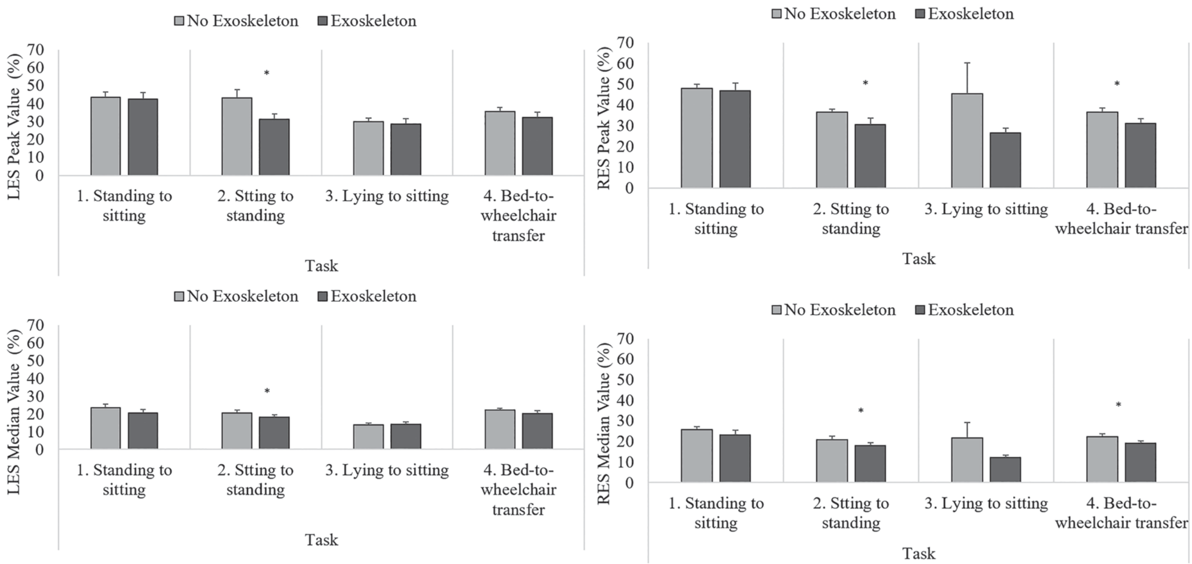


Figure 3. Mean and standard error (whisker) of the peak (90th percentile) and median (50th percentile) LES and RES muscle activities (%MVC) with and without the back-support exoskeleton for the four different patient handling tasks. *Significant difference with and without the exoskeleton ($p < 0.05$).
Note: LES = left erector spinae; MVC = maximum voluntary contraction; %MVC = peak normalized muscle activity; RES = right erector spinae.

Table 4. Statistical analysis results for left and right peak (90th percentile) and median (50th percentile) erector spinae muscle activities: *p* values of repeated-measures ANOVA and effect sizes.

Statistic	Variable	Task 1	Task 2	Task 3	Task 4
p	Left peak erector spinae	0.628	0.004*	0.458	0.12
	Left median erector spinae	0.06	0.049*	0.983	0.083
	Right peak erector spinae	0.638	0.02*	0.181	0.003*
	Right median erector spinae	0.075	0.017*	0.206	0.007*
Effect size	Left peak erector spinae	−0.082	−0.743	−0.134	−0.335
	Left median erector spinae	−0.347	−0.395	0.004	−0.321
	Right peak erector spinae	−0.103	−0.586	−0.441	−0.630
	Right median erector spinae	−0.345	−0.507	−0.439	−0.599

*$p < 0.05$.
Note: Cohen's *d* was used to assess the effect size. ANOVA = analysis of variance; task 1 = lying to sitting; task 2 = sitting to standing; task 3 = standing to sitting; task 4 = bed-to-wheelchair transfer.

Table 5. Statistical analysis results for heart rate: *p* values of repeated-measures ANOVA and effect sizes.

Statistic	Variable	Task 1	Task 2	Task 3	Task 4
p	Heart rate (bpm)	0.858	0.286	0.352	0.132
Effect size	Heart rate (bpm)	−0.044	−0.295	0.297	0.212

Note: Cohen's *d* was used to assess the effect size. ANOVA = analysis of variance; task 1 = standing to sitting; task 2 = sitting to standing; task 3 = lying to sitting; task 4 = bed-to-wheelchair transfer.

4. Discussion

This study evaluated the effect of the back-support exoskeleton on trunk and hip postures, lower back muscle activity and heart rate during bed-to-chair patient handling tasks. The exoskeleton significantly affected the hip postures and lower back muscle activities; trunk postures and heart rate were not significantly affected by exoskeleton use. The biomechanical impact of the exoskeleton was task-dependent. The exoskeleton reduced the lower back muscle activities for sitting-to-standing and bed-to-chair transfer tasks.

For the bed-to-chair patient handling tasks, the impact of the exoskeleton use was more substantial in hip flexion than in trunk flexion and extension. A previous study tested the Laevo exoskeleton [42] and reported that the back-support exoskeleton had a larger effect on decreasing hip extensor (up to 28%) demand than the trunk extensor demand (up to 6%) during symmetric and asymmetric lifting [42]. This could be related to the specific design mechanism of the Laevo exoskeleton, as the anchor points are located at the thigh and designed to produce the hip extensor moment with other supporting anchor points. This could alternate the participant's hip postures by reducing hip flexion during patient handling activities. For the lying-to-sitting task (task 3), there were no discernible differences in the median values of hip flexion angles with and without the exoskeleton. This observation may be attributed to the comparatively lower demand for hip postures (28°–30°) during the lying-to-sitting task, in contrast to the range observed in other tasks (37°–53°). The tension induced by the gas springs of the exoskeleton could function more effectively in high demand postures, which is supported by previous findings [24,43]. It was noted that there was a lack of statistical significance in the median angles of hip flexions, particularly for task 3 (standing to sitting). In general, this task required less hip flexion (< 30°) compared to other tasks, as seen in Figure 1. As the hip flexion was already small during the no exoskeleton condition, the device was not needed to further correct the hip flexion, which emphasizes that the impact of the back-support exoskeleton depended on the type of the tasks performed by caregivers.

Although trunk flexion and extension angles were not statistically different, the exoskeleton increased the peak trunk flexion angles compared to no exoskeleton (3°–8°). This was in line with previous findings that found back-support exoskeletons increased the trunk angle compared to no exoskeleton during squat and stand pivots and scoot patient transfer tasks [24]. This suggests that caregivers wearing a back-support exoskeleton could potentially adjust their patient handling strategies by reducing the hip flexion angles, but increasing the trunk extension angles. The findings indicated divergent effects of the back-support exoskeleton on the muscle activities of the erector spinae and the trunk angles. Specifically, the exoskeleton exhibited a reduction in erector spinae muscle activities during specific patient-handling tasks, while the trunk angles did not show statistical variations with and without the exoskeleton. This implies that the support torque assistance provided by the exoskeleton may play a more prominent role in diminishing muscle activities rather than inducing alterations in trunk motion. Although the increased trunk angle was not associated with the muscle activities of the erector spinae in the current study, careful consideration and further investigation should be done to assess unanticipated outcomes while using exoskeletons. The co-activation of the surrounding trunk muscles including the erector spine, abdominal muscles and oblique muscles was known to increase trunk stiffness and spinal loadings [44]. For example, the antagonistic abdominal muscle activity could be recorded and analyzed to determine whether increased trunk stiffness and stabilization is required with exoskeleton use [44].

Reduction of erector spinae muscle activity was observed for the sitting-to-standing and bed-to-wheelchair transfer tasks (5–12%MVC) with back-support exoskeletons. Furthermore, it appeared that exoskeleton use only reduced the muscle activity of the right side of the erector spinae muscle during bed-to-wheelchair transfer tasks, which could be due to the asymmetric exertions required for this task. The participant was required to transfer a patient from the center to a left side. This left twisting movement could exert more muscle activities on the right side of erector spinae, which could be assisted more by the exoskeleton support.

Previous studies have shown that back-support exoskeletons reduce muscle activities of the erector spinae [43,45], which may be a result of the assistive back torque produced by the exoskeleton. A prior study demonstrated an erector spinae muscle activity reduction of up to 11.2% using the Laevo exoskeleton during patient transfer tasks between a wheelchair and a bed [24]. In contrast, the current study revealed a higher reduction of muscle activity, reaching up to 19%. This variance might be attributed to difference characteristics inherent in the patient handling tasks conducted in each study. The reduction of the muscle activity of the erector spinae was greatest during the sitting-to-standing task, which indicates that the effect of the exoskeleton was more apparent in symmetric exertions compared to asymmetric exertions. These results are similar to previous literature where symmetric lifting exertion showed a greater reduction of back muscle activities compared to asymmetric lifting exertion [42,46]. This suggests that there is a need to improve the design mechanisms of back-support exoskeletons to alleviate the physical demand of asymmetric exertion. Moreover, the generation of support torque from the exoskeleton may exhibit inconsistencies based on the specific patient handling tasks undertaken, warranting a thorough investigation into this aspect. Asymmetric trunk postures commonly occur during patient handling tasks [47,48] and patient handling-specific exoskeletons should consider these aspects.

There were no significant changes in heart rate when using the exoskeleton during patient handling tasks. This was similar to a previous study showing that a back-support exoskeleton did not change the heart rate during manual waste collection [31]. Conversely, another study reported that a back-support exoskeleton significantly reduced the heart rate during a simulated sorting task [30]. The heart rate serves as a reliable indicator of physical activity intensity, and there is strong correlation between Borg's [49] perceived exertion rating and the heart rate. In the present study, use of the exoskeleton did not appear to have a significant impact on task intensity as measured by heart rate. Moreover, similar to the findings in the sorting task study conducted by Bar et al. [50], the tasks employed in our study may have been of insufficient duration to detect significant changes in cardiovascular strain over the course of a complete working shift. Additionally, advanced heart variability analysis could be conducted on the raw ECG data to examine the complex interactions between the autonomic nervous system and the cardiovascular system. Further investigation is warranted to explore these effects in future studies.

Although this study was carefully designed, several limitations were noted. First, only one model of back-support exoskeleton (Laevo V2.5) was tested in this study. Various back-support exoskeleton models produced by different manufacturers likely involve different assistive torque mechanisms, which may lead to different results. Second, a low sample size ($N = 8$) was evaluated in this study. The data were collected during the pandemic period, when it was challenging to recruit many participants to conduct a study. In order to mitigate the impact of the low sample size, we carefully designed the study to improve reliability. Although the results revealed a practical difference in outcome measures supported by the effect sizes, the statistical power could be reduced due to the low sample size. In addition, the study may not have enough statistical power to investigate the sex differences in the exoskeletons' effects. Additionally, future research could delve more thoroughly into the task characteristics. Examining tasks that distinguish between symmetric and asymmetric aspects could aid in discerning the differential impact of exoskeletons. Exploring both directions, not only from chair to bed but also vice versa, would provide a comprehensive understanding of the effectiveness of exoskeletons. Lastly, we evaluated the effectiveness of the back-support exoskeleton in a controlled laboratory setting. The feasibility and adaptability of exoskeletons in healthcare workplaces should be explored in the future.

5. Conclusion

The impact of the exoskeleton varied depending on the task at hand. Notably, a significant reduction in erector spinae muscle activities ($p < 0.05$) was observed during task 2 (sitting to standing) and task 4 (bed-to-wheelchair transfer) when utilizing the exoskeleton. Across all four tasks, the exoskeleton demonstrated a marked decrease in peak bilateral hip flexion angles (up to 29°) and a reduction in median hip flexion angles in three tasks, excluding task 3 (standing to sitting). While no statistically significant differences were found in trunk flexion and extension angles, the exoskeleton resulted in an increase in trunk extension angles ranging from 3° to 8°. The heart rate showed no significant difference between wearing and not wearing the exoskeleton. Furthermore, the exoskeleton exhibited a more pronounced effect on reducing lower back muscle activities in symmetric exertions compared to asymmetric exertions. This indicates a potential need for design improvements in the exoskeleton to address the specific physical demands associated with asymmetric exertions. Participants, when equipped with exoskeletons, adjusted their patient handling strategies by decreasing hip flexion and increasing trunk extension. To comprehensively understand whether the exoskeleton alters spine stiffness and stability during patient handling, further investigation is warranted. The insights gained from this study could contribute to refining the design of back-support exoskeletons tailored to the demands of patient handling activities.

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