Effects of using a whole-body powered exoskeleton during simulated occupational load-handling tasks: A pilot study

Hanjun Park, Sunwook Kim, Maury A. Nussbaum and Divya Srinivasan*
Department of Industrial and Systems Engineering, Virginia Tech, Blacksburg, VA, USA

CORRESPONDING ADDRESS: Divya Srinivasan
Department of Industrial and Systems Engineering,
Virginia Tech, 250 Durham Hall (0118), Blacksburg, VA 24061, USA.
Phone: (540) 231-0083. E-mail: sdivya1@vt.edu.

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ABSTRACT

Whole-body powered exoskeletons (WB-PEXOs) can be effective in reducing the physical demands of heavy occupational work, yet almost no empirical evidence exists on the effects of WB-PEXO use. This study assessed the effects of WB-PEXO use on back and leg muscle activities during lab-based simulations of load handling tasks. Six participants (4M, 2F) completed two such tasks (load carriage and stationary load transfer), both with and without a WB-PEXO, and with a range of load masses in each task. WB-PEXO use reduced median levels of muscle activity in the back (~42–53% in thoracic and ~24–43% in lumbar regions) and legs (~41-63% in knee flexors and extensors), and mainly when handling loads beyond low-moderate levels (10–15 kg). Overall, using the WB-PEXO also reduced inter-individual variance (smaller SD) in muscle activities. Future work should examine diverse users, focus on finding effective matches between WB-PEXO use and specific tasks, and identify applications in varied work environments.

1. INTRODUCTION

A growing interest has emerged in the use of exoskeletons (EXOs) as a new ergonomic intervention to reduce work-related musculoskeletal disorder (WMSD) risks (e.g., de Looze et al., 2016). EXOs are wearable devices designed to assist and/or augment the user with supportive forces or moments during diverse physical activities and in different work environments, with potential to reduce physical demands and enhance task performance without limiting human flexibility. EXO technologies can be categorized (Lee et al., 2012) broadly as either *passive* (using restorative energy from mechanical springs, dampers, etc.) or *active/powered* (using powered actuators and/or motors to generate supportive forces and moments). Passive EXOs have been studied extensively in terms of their impacts on a user's physical demands during various work tasks, including manual lifting (Abdoli-E et al., 2006; Alemi et al., 2019; Bosch et al., 2016; Koopman et al., 2020; Wei et al., 2020; Whitfield et al., 2014), overhead work (Alabdulkarim & Nussbaum, 2019; Kim et al., 2018a; 2018b) and assembly-related tasks involving trunk bending (Kim et al., 2020; Luger et al., 2019). Limited research has been presented on powered EXOs for occupational use, however, perhaps because passive technology is currently simpler, more mature, and affordable; and the majority of commercially-available EXOs for occupational applications are passive (e.g., exoskeletonreport.com).

Current evidence on passive EXOs supports their efficacy as an ergonomic intervention to reduce physical demands, although the benefits and limitations of EXO use can be substantially influenced by the specific EXO design and task conditions (Alemi et al., 2020; Amandels et al., 2019; Hensel & Keil, 2019; Madinei et al., 2020). For example, Alemi et al. (2020) compared two back support EXOs (BSEs; Laevo^{TM} and SuitX^{TM}) during symmetric and asymmetric repetitive lifting tasks and found both BSEs to be beneficial in terms of reducing back muscle activities. However, larger reductions in trunk extensor muscle activity were evident in symmetric vs. asymmetric lifting, and mixed results were observed in terms of perceived discomfort. Similarly, Madinei et al. (2020) compared the BackX^{TM} and Laevo^{TM} during several different conditions of precision manual assembly tasks, and found that reductions in trunk muscle activity were substantially posture-dependent (larger trunk extensor muscle activity reductions in the task conditions closer to the mid-sagittal plane; ≤ 47% and ≤ 24% reductions in trunk extensor muscle activity were found for BackX^{TM} and Laevo^{TM} respectively).

This task dependency or specificity is primarily rooted in the passive EXO design approach. Specifically, a supportive force/moment generation mechanism responds to body motions or postures (e.g., bending the trunk for a back-support exoskeleton, elevating the arm for an arm-support exoskeleton), and thus the level of assistance/support is typically a function of the angle between the two body segments involved. The support level is often adjustable, yet it is not possible to adjust in real-time during a task. Levels of support are also limited, since a passive EXO provides support regardless of body movement directions. The user thus may work against the support (e.g., lowering an arm), and may experience high contact pressure and discomfort where the body segments interface with EXO components. In contrast, powered EXOs can control supportive force/moment levels in response to the user's intention and can enable dramatic strength augmentation, rendering powered EXO technologies more versatile and flexible. Powered EXOs are thus considered an important aspect of the future workforce with the advent of the Industry 4.0 era (Romero et al., 2016).

The design of powered EXOs typically comprises three major components: actuator(s), transmission, and wearable structures. Based on the transmission and structural designs, powered EXOs can be categorized as either rigid or soft (Sanchez-Villamañan et al., 2019; Toxiri et al., 2019). Rigid EXOs are built with rigid linkages aligned parallel to human segments and deliver assistive torque to one or more target joints. Rigid body EXOs reduce physical demands with assistive torques, and thereby may reduce the risks of WMSDs (Huysamen et al., 2018; Toxiri et al., 2018; von Glinski et al., 2019). Soft EXOs rely on cable transmission and/or garment-like functional textile-based wearable structures to transfer power from the actuator(s) to the user through linear forces along with the musculoskeletal system. Compared to rigid EXOs, soft EXOs are more effective in minimizing problems of joint misalignment, and their lighter weight provides more versatility (Ding et al., 2018). However, soft EXOs lack a weight-supporting framework, and thus could less effect for implementations in heavy-duty tasks (Lee et al., 2017).

The WB-PEXO assessed in the current study is a rigid system capable of dramatically augmenting human strength to perform heavy-duty tasks. Existing reports on powered EXOs have focused largely on enhancing basic design elements, such as assistive strategies (Hamaya et al., 2017; Krausz et al., 2020), structural designs (e.g., degrees-of-freedom and joint actuators; Jafari et al., 2010; Zoss et al., 2006), and force sensor integration (Grosu et al., 2015) to follow user intention. As recently discussed by Toxiri et al. (2019), an important challenge in powered EXOs is to generate appropriate supportive forces/moments to match the user's intention during physical activities. Human-subjects testing of powered EXOs designed for a specific body region (e.g., low back or shoulder) has shown that these devices can effectively assist the user during physical activities. For example, use of a powered back-support EXO prototype (Robomate; www.robo-mate.eu) reduced trunk extensor muscle activity by up to 15% (Huysamen et al., 2018) and lumbosacral compression forces by ~18% during various lifting tasks (Koopman et al., 2019). The Hybrid Assistive Limb (HAL®), a powered back-support EXO, also reduced trunk extensor muscle activity during symmetric lifting, by up to ~20% (von Glinski et al., 2019). Muscle activity of the anterior deltoid was decreased by up to ~58% during three different simulated overhead tasks using a powered arm-support EXO ("Lucy"; Otten et al., 2018).

Although whole-body powered exoskeletons (WB-PEXO) were first conceptualized and developed decades ago (i.e., Hardiman between 1965 and 1971; Makinson, 1971), this technology has only recently become viable for practical use. In contrast to powered EXOs that are designed to support a specific body region, WB-PEXOs can transfer external loads/forces to the ground without the need to re-distribute loads over different, un-augmented body parts. WB-PEXOs thus offer a greater potential to control the physical demands imposed on a user and to permit "super-human" strength in highly-demanding tasks. Yet, available evidence on WB-PEXOs is generally limited to the technical specifications (e.g., maximum payload, motor power), design, and development of WB-PEXO elements. In one example of human-subjects testing, Fontana et al. (2014) discussed their Body Extender system while presenting single user data during several activities (e.g., trunk rotation and squatting, lifting) and walking, but the impacts on the user were not reported. Recently, our research group reported preliminary results using a WB-PEXO research prototype (Model P1, Sarcos Robotics) for one-arm lifting (Kim et al., 2019); we found a substantial reduction in arm muscle activities (trapezius and anterior deltoid) and a low-moderate increase of muscle activity in the lumbar region when operating a load of 11.3 kg. It is unclear, however, whether a WB-PEXO would offer different benefits depending on task types and load levels.

To enable a better understanding of the potential occupational impacts of using WB-PEXOs, and to facilitate their effective future adoption, we completed an exploratory study to assess how using a state-of-the-art WB-PEXO (Alpha prototype of Guardian® XO®, www.sarcos.com) impacts a human operator in terms of the demands on the trunk and leg musculature. Specifically, two different load handling tasks were considered that are common across various industry sectors – load carriage and stationary load lifting/lowering – and a range of load masses were handled. Load handling tasks were of particular interest as they can impose high demands on the low back (e.g., Da Costa & Vieira, (2010), and lifting and carrying heavy loads is considered an important potential use-case of occupational WB-PEXOs (Fontana et al., 2014). We expected that task type (load carriage vs. stationary load handling) and load mass would influence the potential benefits of WB-PEXO use, in terms of muscular demands. Results from the current study are intended to help guide future improvements in WB-PEXO design and identify specific occupational use-cases.

2. METHODS

2.1 Participants

A convenience sample of six healthy volunteers (4 males and 2 females) completed this exploratory study. Mean (SD) stature, body mass, and age were 1.81 (0.04) m, 81.6 (17.2) kg, and 28.3 (6.3) years, respectively. All participants were right-handed, and none had any self-reported musculoskeletal injuries or disorders in the last 12 months. This study protocol was approved by the Virginia Tech Institutional Review Board, and all participants provided written informed consent prior to data collection. To minimize learning effects during the experiment, all participants first received extensive training (>8 hours, over multiple sessions) in using the WB-PEXO, which was continued until they reported that they could operate it competently to perform basic tasks (walking, bending, lifting, etc.).

2.2 WB-PEXO

The device used in the current study is the alpha prototype of the Guardian® XO® developed by Sarcos Robotics. This system has a mass of 160 kg, an anthropomorphic design, and 24 active degrees-of-freedom, including: the shoulders (flexion/extension and abduction/adduction), elbows (flexion/extension), humeral (axial rotation), wrist (axial rotation), trunk (axial rotation and lateral bending), hips (flexion/extension, abduction/adduction, and axial rotation), knees (flexion/extension), shank (axial rotation), and ankles (flexion/extension). Designed for occupational purposes, this WB-PEXO is intended to provide an operator with the ability to safely lift and manipulate loads up to 90 kg, with external joint torques being applied at the major body joints. This ability is achieved through a patented "Get-Out-Of-The-Way" control scheme to mimic human movements and augment joint torques (Jacobsen, S. C., Olivier, M. X., & Maclean, 2010). The WB-PEXO consists of various tunable parameters, including actuation gains, along with payload and gravity compensation, that can be adjusted for a specific operator. Being a prototype version, the WB-PEXO's hardware and control implementations continue to be refined, to achieve further benefits on the musculoskeletal loading experienced by the operator: the current study was conducted at a defining point in development, to benchmark the benefits of the current version through user-evaluations, and to identify specific use-cases to guide further design optimizations.

2.3 Load Handling Tasks

With load handling as the broad task type of interest, we studied two specific load handling scenarios (Figure 1): (1) load carriage, involving lifting and carrying loads from one place to another; and (2) stationary load transfers, involving a large range-of-motion of major body joints. We chose these two tasks to compare the effects of using the WB-PEXO during lifting/lowering different loads vs. carrying different loads (with and without loads), the latter being where the human user is ambulatory and balancing the WB-PEXO.

(Figure 1 about here)

The load carriage task involved using both hands to: lift a loaded carrier bag placed on the ground in front of the participant, carry it along a 7.5 m walkway, turn around, carry it back to the starting point, and place it on the ground. A hook-shaped end effector, attached to each wrist of the WB-PEXO, was used to pick up and carry the loads. Five different levels of load mass included (4.5, 10, 16, 20, and 26 kg). Stationary load transfers involved moving a loaded bag between three levels of a storage rack with one arm. The vertical height of the bottom, middle, and top levels of the rack were set at 11, 103 and 168 cm, and were selected to approximate the foot, elbow, and head heights of an average U.S. adult, respectively (Fryar et al., 2016). Participants stood in front of the rack, at a distance of roughly one arm length, though they could adjust this distance until they felt comfortable reaching the bottom shelf without adjusting the location of their feet. Load transfers began with the weighted bag placed on the middle shelf (Figure 1), and the task was performed with seven different loads (0, 4.5, 5.7, 9.5, 20, 32, and 47 kg).

2.4 Procedures

Participants completed the experiment in a single experimental session (~2 hrs). A repeated-measures design was used with two independent variables for each of the two tasks: *Intervention* (WB-PEXO and control conditions) and *Load Mass* (5 levels for load carriage and 7 levels for stationary load transfer). The fit of the WB-PEXO and its tunable parameters (such as harness adjustments, controller gains, and gravity compensation) were set according to individual initial preferences in the beginning of the initial training session. These parameter values were further optimized (continually adjusted) based on the user's feedback, during the same session. Specifically, parameter values were adjusted with constant intervals (either in increasing or decreasing steps) until the user was comfortable, and felt competent to perform simple lifting or walking tasks with the XO. For the load carriage task, participants completed three trials of each task condition at a self-selected pace and were asked to carry the bag without specific instructions. Sufficient rest was provided between tasks to minimize muscle fatigue. To reduce potential learning effects, the order of *Intervention* conditions was first randomized, and the order of *Load Mass* was then randomized within each *Intervention* condition.

2.5 Instrumentation and data processing

Muscle activity was monitored at 1.5 kHz using a telemetered surface electromyographic (EMG) system (Ultium^{TM}, Noraxon, AZ, USA). After initial skin preparation, pairs of pre-gelled, bipolar, Ag/AgCl electrodes were placed bilaterally over two accessible trunk muscle groups based on procedures described earlier (Cram, 2010; Jia et al., 2011): the lumbar erector spinae (LES) at the L3 level; and the thoracic erector spinae (TES) at the T10 level. Additional electrode pairs were placed unilaterally (right-side) over four accessible muscle groups in the lower extremity: vastus lateralis (RVL), biceps femoris (RBF), anterior tibialis (RTA), and medial gastrocnemius (RMG). At the start of each experimental session, maximum voluntary isometric contractions (MVICs) were completed for each muscle group. All MVIC testing was done using a commercial dynamometer (Biodex 3 Pro, Biodex Medical Systems Inc., NY, USA), with a custom fixture to restrain the pelvis and legs. For the thoracic and lumbar muscles, participants performed maximal trunk extension while standing upright, their feet slightly separated, their pelvis and legs secured, and the trunk flexed to ~20° (Jia et al., 2011). For the leg muscles, participants were secured using straps on the dynamometer chair and performed maximal right knee flexion and extension with the knee flexed at multiple angles between ~50°and ~90° (Babault et al., 2001; Bouillard et al., 2014). For each muscle group, MVIC trials were replicated twice, and with non-threatening verbal encouragement. EMG signals obtained during both MVICs and task trials were band-pass filtered (20–450 Hz, 4th-order Butterworth, bidirectional), and root-mean-squared (RMS) amplitudes were subsequently obtained with a 300 *ms* time constant. Normalized EMG (nEMG) values were obtained using the corresponding maximum values obtained during MVICs for each muscle group. For each trial of a given work task, median (50th percentile) nEMG was the primary outcome measure and was used as an indicator of overall muscular activation. Peak (90th percentile) nEMG was also computed, with results presented in the Appendix as a secondary outcome measure. In the load carriage task, outcome measures were calculated during the times when participants were walking with the load. In the load transfer task, outcome measures were averaged over a full lifting and lowering cycle.

2.6 Statistical analysis

Summary results are presented as means (SDs). Separate two-way, repeated-measures analyses of variance (ANOVAs) were used to assess the effects of *Intervention* (WB-PEXO, control) and *Load Mass* (five levels during load carriage, and seven levels during load transfer) on each outcome measure. Significant interaction effects were followed by Tukey's HSD post hoc pairwise comparisons of *Intervention* effect. Gender was not included in the model due to the small sample size. We observed no substantial departures from parametric model assumptions. ANOVA effect sizes are reported using eta-squared (η^2) values and post hoc effect sizes are reported as *Hedge's g* (Rosenthal et al., 1994). All statistical analyses were performed using JMP Pro (v. 15, SAS, Cary, NC), with the restricted maximum likelihood (REML) method, and statistical significance was determined when $p < 0.05$.

3. RESULTS

3.1 Load carriage

Effects of *Intervention* and *Load Mass* on median nEMG are summarized in Table 1. Across the loads examined, median nEMG values were typically 10-50% in the bilateral TES and LES, and 7-37% in the leg muscles (Figure 2). As can be seen from the trends in Figure 2, muscle activities were higher in the WB-PEXO vs. control condition when carrying lighter loads, and then they "crossed over" each other at loads between 10 and 20 kg: muscle activities were lower when using the WB-PEXO at higher loads. Such a cross-over was not evident for either the RTES or RBF, and both muscles had lower median nEMG values when using the WB-PEXO with all load masses. Higher activity occurred in the RTA when using the WB-PEXO with all load masses (by 60% overall). In terms of statistical results, there were significant main effects of *Intervention* on RTES, RVL, RBF, RTA and RMG; and there was a significant interaction effect of *Intervention* and *Load Mass* on the remaining muscles (LTES, RLES and LLES). Statistical results and the effects of *Intervention* and *Load Mass* on peak muscle activities were largely consistent with those for median activities (see Table A1 and Figure A1 in the Appendix). An exception was the RTES, for which the cross-over point occurred at a higher load (15-20 kg).

(Figure 2 about here)

(Table 1 about here)

3.2 Stationary load transfers

A summary of ANOVA results for median nEMG is presented in Table 2. In general, median nEMG values in the control condition ranged from 7 to 49% in the back muscles and from 3 to 40% in the leg muscles (Figure 3). A similar pattern was observed in the bilateral TES in both WB-PEXO and control conditions, in that increases in load mass led to similar increases in muscle activity up to ~20 kg. With loads above 20 kg, however, muscle activity seemed to increase at a slower rate when using the WB-PEXO (as can be seen in Figure 3). While *Intervention* had a significant main effect on RTES, *Intervention* and *Intervention* × *Load Mass* interaction had significant effects on LTES. Both main and interaction effects of *Intervention* and *Load Mass* were significant on the RLES and LLES muscles. From Figure 3, it seems that all muscle groups showed a cross-over point with a load less than ~20 kg, above which muscle activities were lower when using the WB-PEXO. The two leg muscles (RBF, RMG) had similar increases in median nEMG up to ~30 kg, with activation increasing more rapidly beyond ~30 kg. Activity in most leg muscles was similar between *Intervention* conditions when loads were <20 kg, however a reduction (30% on average) occurred using the WB-PEXO with greater loads. Similar to the load carriage task, results of the statistical analysis and the effects of *Intervention* and *Load Mass* on peak nEMG (Table A2 and Figure A1 in the Appendix) largely mirrored those for median nEMG.

(Figure 3 about here)

(Table 2 about here)

4. DISCUSSION

The main purpose of our exploratory study was to gather initial evidence on whether using a WB-PEXO is viable, and to quantify the effects of using a WB-PEXO on the physical demands of the operator when performing an initial set of occupationally-relevant tasks. In a broader sense, we also hoped to identify relevant task characteristics that can aid in selecting potential applications for using a WB-PEXO, and to guide further design optimizations for this specific prototype that is under development.

4.1 Muscle activity: Influence of different load masses and task types

Effects of using the WB-PEXO varied between task types and load masses, though larger reductions were observed overall with higher load masses (Figures 2 and 3), some of which were statistically significant. In the load carriage task, Figure 2 qualitatively shows that activity in most muscles increased monotonically with load mass, though at a slower rate in the WB-PEXO condition. Exceptions were observed in the right vastus lateralis (RVL), right tibialis anterior (RTA), and right medial gastrocnemius (RMG), for which activity levels were independent of load mass when the WB-PEXO was used. In the stationary load transfer task, activity in some muscles increased with load mass, whereas in other muscles (bilateral TES and right LES) the activity first increased and then plateaued at higher loads. Using the WB-PEXO seemed to result in a greater reduction in muscle activity in the stationary task than the load carriage task, as indicated by the magnitude of changes seen in Figures 2 and 3 and larger post hoc effect sizes observed across most muscle groups. Furthermore, there was a difference between these tasks in terms of the cross-over point – the load at which using the WB-PEXO led to beneficial effects in terms of muscle activity. This cross-over point was lower in the stationary task.

We believe that some of these task-related differences stemmed from users having to compensate for the substantial inertia of the WB-PEXO to maintain the balance of the human+EXO system during the load carriage tasks, compared to being stationary in the other task. Although there is limited evidence in the literature, it has been suggested that maintaining balance while walking with a WB-PEXO can be difficult without the assistance of active balance control (Fontana et al., 2014). Such balance issues may have led users to increase muscle activation, especially via co-contraction, to stiffen their joints and thereby compensate for the inertia of the human+EXO system in a dynamic condition. Developers of the prototype examined here continue to explore ways to better accommodate users via inertial compensation, but the version tested did not include active balance control. Implementing active balance control is not straightforward, though, as such control could interfere with a user's intended movement and lead to undesirable scenarios (e.g., falling). Further work is needed to better understand the effects of implementing active balance control on a user's control strategies and muscle activation.

4.2 Comparisons to other powered exoskeletons

Although not directly comparable, there are a few reports on powered back EXOs for repetitive lifting, which indicated reductions in back extensor activity of 12-30% (Huysamen et al., 2018; Toxiri et al., 2018; von Glinski et al., 2019). Our results were largely similar, with 13-27% reductions in muscle activity found when handling loads of 5.7 and 20 kg as used in these prior studies. That a WB-PEXO provides similar reductions in back muscle activity as a back-support exoskeleton is quite promising in terms of applications and impact, as a WB-PEXO also confers benefits to other major muscle groups in the body, such as the arms and legs. von Glinski et al. (2019) assessed the effects of using the HAL® for Care Support device (powered back exoskeleton) during repetitive lifting, and reported decreases in back muscle activities (11% and 4.5% respectively for thoracic and lumbar) and an increase in quadriceps muscle activities (~18.7%). Our results showed considerably higher reductions in back muscle activity (30% and 23% for thoracic and lumbar, compared to 11% and 4.5% respectively for comparative load levels), and a similar increase in quadriceps (RVL) muscle activity during lifting tasks (11%), and mainly when the load was below 10 kg. An increase in quadriceps activity may have stemmed from the control strategy currently used by the WB-PEXO, which provides limited assistance gain when handling loads at low elevations. Users may have used more hip flexion to compensate for the weight of the WB-PEXO during the bending phases of the lifting task at such elevations. This speculation, though, needs to be confirmed using kinematic measures. Overall, the WB-PEXO examined here seems to be comparable to, and in some cases even outperform, contemporary powered EXOs that have been tested, in terms of reducing trunk and leg muscle activity in controlled lifting and load carriage tasks. It should be noted that while the descriptive values were similar, they were not found to be statistically significant and that if these values are deemed to be operationally relevant, it will be important to design confirmatory studies that can detect these effect sizes.

Regarding the higher level of muscle activation in the lower limbs observed here during gait, Russell & Apatoczky, (2016) reported higher activities in gastrocnemius and tibialis anterior muscles when individuals walked both faster and slower than their preferred/self-selected cadences. Here, all participants walked at a slower pace when using WB-PEXO compared to the control condition. Further research, however, is required to confirm how gait stability and EXO-control strategies affect different muscle groups and coordination while operating a complex, heavy, and powerful WB-PEXO.

4.3 Powered vs. passive exoskeletons

Previous research (Abdoli-E et al., 2006) found ~28% reductions in back muscle activity using the PLAD during symmetrical lifting of three different loads (5, 15, 25 kg). Similarly, several studies (e.g., Alemi et al. (2020)) have reported ~15-25% reductions in thoracic and iliocostalis lumborum activities during symmetrical lifting tasks when using the SuitX™ and Laevo™ EXOs. Reductions in back muscle activity observed here are comparable to these previous reports. It is important to note that while this low-moderate load range (5-20 kg) has been the most commonly studied when assessing passive EXOs (to date), it is when loads exceeded this range that the WB-PEXO likely becomes more beneficial, which was also supported by the significant pairwise differences observed in the higher load levels in our statistical analysis. This beneficial effect of the device, specifically when loads exceed ~20kg, suggests a clear potential for powered EXOs to augment workers to do tasks that were previously considered infeasible for human operators. Therefore, in terms of occupational applications, both passive and powered EXOs will likely have distinct applications depending on use-case requirements. Other practical considerations may also affect the choice of EXO, such as task configuration, power requirement, space availability, and cost.

While most passive EXOs are designed to deliver pre-specified levels of support that are infeasible to adjust in real-time during operation, powered EXOs can have their control parameters tuned to provide appropriate assistance during operation (Toxiri et al., 2018). In the current study, however, the WB-PEXO was operated using constant assistance gain, similar to passive EXOs. Gain selection thus may not have been optimal for the full range of load masses tested. Hence, to see comparable beneficial results regardless of payloads, the WB-PEXO prototype is undergoing a subsequent stage of development, which includes improving control assistance through enhanced task-dependent predictions of user intent (i.e., posture, load level being handled).

4.4 Inter-Individual Variability

Large inter-individual variability in nEMGs was evident in the no-EXO (control) condition during both tasks, which in many cases increased with load mass (Figures 2 and 3). Since all EMGs were normalized to maximal voluntary capacity, this large variability implies a large variance in strength and/or differences in technique among the current participants. Interestingly, using the WB-PEXO notably reduced inter-individual variability across all loads examined in the thoracic muscles, vastus lateralis, and biceps femoris (i.e., observably smaller SDs in Figures 2 and 3). On one hand, this reduction in variability suggests that using a WB-PEXO may serve as an effective intervention, especially for weaker/older or more diverse populations in occupational settings, as WB-PEXO use could produce levels of muscle activity comparable to those among stronger individuals. On the other hand, if reduced inter-individual variance observed when using the WB-PEXO was secondary to restrictions on the range of movement strategies that diverse users could employ, it is a concern that needs further investigation.

Reduced inter-individual variability with WB-PEXO use, however, was not evident in all of the muscle groups monitored; both lumbar (RLES and LLES) and lower leg (RTA and RMG) muscle groups had relatively similar levels of variability in the WB-PEXO and control conditions. Further research is thus needed to ensure that a WB-PEXO effectively accounts for individual differences (i.e., strength, lifting techniques, and gait speed) and to examine if WB-PEXO use similarly reduces inter-individual variability when tested among a more diverse sample, or when user strength is intentionally manipulated as an experimental variable.

Finally, movement speed can also affect muscle activity. Individuals performed tasks here at self-determined paces, in all tasks and experimental conditions. These speeds were not directly measured, and hence their effects on inter-individual differences in muscle activity could not be determined. Future work should consider measuring and reporting the effects of pace, especially in walking tasks.

4.5 Limitations

Some limitations of the current study should be acknowledged. First, a relatively small and homogenous convenience sample was included (young and healthy participants), due to the elaborate training and safety protocols involved with operating the WB-PEXO prototype examined. Recruiting and testing a larger and more diverse sample in the future will help provide a more accurate and generalizable quantification of the effects of WB-PEXO use. Second, although all participants were considered to be experienced using heuristic criteria and subjective opinions of the investigators, it is still an open research question regarding how to precisely and objectively define what constitutes expertise in operating a complex WB-PEXO, or how long it may take one to achieve such expertise. Third, all work tasks performed here were simulations conducted in a controlled laboratory setting, and the generalizability to actual occupational tasks is unknown. Fourth, there was a certain degree of misfit between the EXO interface and the bodies of the participants, and this fit changed dynamically as participants adopted different postures. The effects of such fitting issues, due to individual anthropometric differences, on EXO effectiveness and participant comfort are open research questions (Stirling et al., 2020). Fifth, only short-term effects of WB-PEXO use were investigated, so caution should be taken in generalizing results to more prolonged situations. Finally, we only examined the effects of WB-PEXO in terms of muscle activities. However, several factors can affect muscle activity, such as changes in muscle length and velocity. Hence, results obtained from the current study should be utilized for musculoskeletal modeling analysis to better understand how WB-PEXO use affects internal joint loading. Furthermore, kinematic and metabolic data need to be assessed to more comprehensively understand potential use cases and the benefits of a WB-PEXO.

5. Conclusions

We found that using a prototype WB-PEXO becomes beneficial in terms of trunk and leg muscle activities when load mass increased beyond low-moderate levels (~10-15 kg), both for stationary and load carriage tasks involving level walking. Using the WB-PEXO reduced median activity of back muscles (by a range of 8%-53% for thoracic and 5%-43% for lumbar) and leg muscles (by a range of 3%-63%). From the descriptive results shown in figures 2 and 3, the beneficial effects of using the WB-PEXO also seem to be task-specific, with the WB-PEXO showing potential for greater benefits in a stationary task compared to a load carriage task. Given the exploratory nature of the current study, though, it remains unclear regarding the extent to which our results will generalize to a larger sample of diverse individuals with different WB-PEXO operation skill levels, and in other occupationally-relevant tasks. Future research is needed to provide more insights on the tradeoff between EXO assistance and required control efforts from human operators, user adaptations, and the movement control strategies employed when using a WB-PEXO to accomplish diverse tasks.

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REFERENCE

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. *Clinical Biomechanics*, *21*(5), 456–465. https://doi.org/10.1016/j.clinbiomech.2005.12.021

Alabdulkarim, S., & Nussbaum, M. A. (2019). Influences of different exoskeleton designs and tool mass on physical demands and performance in a simulated overhead drilling task. *Applied Ergonomics*, *74*(February 2018), 55–66. https://doi.org/10.1016/j.apergo.2018.08.004

Alemi, M. M., Geissinger, J., Simon, A. A., Chang, S. E., & Asbeck, A. T. (2019). A passive exoskeleton reduces peak and mean EMG during symmetric and asymmetric lifting. *Journal of Electromyography and Kinesiology*, *47*, 25–34. https://doi.org/10.1016/j.jelekin.2019.05.003

Alemi, M. M., Madinei, S., Kim, S., Srinivasan, D., & Nussbaum, M. A. (2020). Effects of Two Passive Back-Support Exoskeletons on Muscle Activity, Energy Expenditure, and Subjective Assessments During Repetitive Lifting. *Human Factors*, *62*(3), 458–474. https://doi.org/10.1177/0018720819897669

Amandels, S., het Eyndt, H. O., Daenen, L., & Hermans, V. (2019). Introduction and testing of a passive exoskeleton in an industrial working environment. In *Advances in Intelligent Systems and Computing* (Vol. 820, Issue February). Springer International Publishing. https://doi.org/10.1007/978-3-319-96083-8_51

Babault, N., Pousson, M., Ballay, Y., & Van Hoecke, J. (2001). Activation of human quadriceps femoris during isometric, concentric, and eccentric contractions. *Journal of Applied Physiology*, *91*(6), 2628–2634. https://doi.org/10.1152/jappl.2001.91.6.2628

Bosch, T., van Eck, J., Knitel, K., & de Looze, M. (2016). The effects of a passive exoskeleton on muscle activity, discomfort and endurance time in forward bending work. *Applied Ergonomics*, *54*, 212–217. https://doi.org/10.1016/j.apergo.2015.12.003

Bouillard, K., Jubeau, M., Nordez, A., & Hug, F. (2014). Effect of vastus lateralis fatigue on load sharing between quadriceps femoris muscles during isometric knee extensions. *Journal of Neurophysiology*, *111*(4), 768–776. https://doi.org/10.1152/jn.00595.2013

Cram, J. (2010). Eleanor. Jones & Bartlett Publishers, 2010. *Introduction to Surface Electromyography*. https://doi.org/10.1016/S0376-7361(09)70018-4

Da Costa, B. R., & Vieira, E. R. (2010). Risk factors for work-related musculoskeletal disorders: A systematic review of recent longitudinal studies. *American Journal of Industrial Medicine*, *53*(3), 285–323. https://doi.org/10.1002/ajim.20750

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*(5), 671–681. https://doi.org/10.1080/00140139.2015.1081988

Ding, Y., Kim, M., Kuindersma, S., & Walsh, C. J. (2018). Human-in-the-loop optimization of hip assistance with a soft exosuit during walking. *Science Robotics*, *3*(15), 1–9. https://doi.org/10.1126/scirobotics.aar5438

Fontana, M., Vertechy, R., Marcheschi, S., Salsedo, F., & Bergamasco, M. (2014). The body extender: A full-body exoskeleton for the transport and handling of heavy loads. *IEEE Robotics and Automation Magazine*, *21*(4), 34–44. https://doi.org/10.1109/MRA.2014.2360287

Fryar, C. D., Gu, Q., Ogden, C. L., & Flegal, K. M. (2016). *Anthropometric reference data for children and adults; United States, 2011-2014*.

Grosu, V., Guerrero, C. R., Brackx, B., Grosu, S., Vanderborght, B., & Lefeber, D. (2015). Instrumenting complex exoskeletons for improved human-robot interaction. *IEEE Instrumentation and Measurement Magazine*, *18*(5), 5–10. https://doi.org/10.1109/MIM.2015.7271219

Hamaya, M., Matsubara, T., Noda, T., Teramae, T., & Morimoto, J. (2017). Learning assistive strategies for exoskeleton robots from user-robot physical interaction. *Pattern Recognition Letters*, *99*, 67–76. https://doi.org/10.1016/j.patrec.2017.04.007

Hensel, R., & Keil, M. (2019). Subjective Evaluation of a Passive Industrial Exoskeleton for Lower-back Support: A Field Study in the Automotive Sector. *IISE Transactions on Occupational Ergonomics and Human Factors*, *7*(3–4), 213–221. https://doi.org/10.1080/24725838.2019.1573770

Huysamen, K., de Looze, M., Bosch, T., Ortiz, J., Toxiri, S., & O'Sullivan, L. W. (2018). Assessment of an active industrial exoskeleton to aid dynamic lifting and lowering manual handling tasks. *Applied Ergonomics*, *68*, 125–131. https://doi.org/10.1016/j.apergo.2017.11.004

Jacobsen, S. C., Olivier, M. X., & Maclean, B. J. (2010). *Control Logic for Biomimetic Joint Actuators*. https://europepmc.org/article/pat/wo2010025403

Jafari, A., Tsagarakis, N. G., Vanderborght, B., & Caldwell, D. G. (2010). A novel actuator with adjustable stiffness (AwAS). *IEEE/RSJ 2010 International Conference on Intelligent Robots and Systems, IROS 2010 - Conference Proceedings*, 4201–4206. https://doi.org/10.1109/IROS.2010.5648902

Jia, B., Kim, S., & Nussbaum, M. A. (2011). An EMG-based model to estimate lumbar muscle forces and spinal loads during complex, high-effort tasks: Development and application to residential construction using prefabricated walls. *International Journal of Industrial Ergonomics*, *41*(5), 437–446. https://doi.org/10.1016/j.ergon.2011.03.004

Kim, S., Lawton, W., Nussbaum, M. A., & Srinivasan, D. (2019). Effects of Using a Prototype Whole-Body Powered Exoskeleton for Performing Industrial Tasks. *Proceedings of the Human Factors and Ergonomics Society Annual Meeting*, *63*(1), 1086–1087. https://doi.org/10.1177/1071181319631469

Kim, S., Madinei, S., Alemi, M. M., Srinivasan, D., & Nussbaum, M. A. (2020). Assessing the potential for "undesired" effects of passive back-support exoskeleton use during a simulated manual assembly task: Muscle activity, posture, balance, discomfort, and usability. *Applied Ergonomics*, *89*. https://doi.org/10.1016/j.apergo.2020.103194

Kim, S., Nussbaum, M. A., Mokhlespour Esfahani, M. I., Alemi, M. M., Alabdulkarim, S., & Rashedi, E. (2018). Assessing the influence of a passive, upper extremity exoskeletal vest for tasks requiring arm elevation: Part I – "Expected" effects on discomfort, shoulder muscle activity, and work task performance. *Applied Ergonomics*, *70*(December 2017), 315–322. https://doi.org/10.1016/j.apergo.2018.02.025

Kim, S., Nussbaum, M. A., Mokhlespour Esfahani, M. I., Alemi, M. M., Jia, B., & Rashedi, E. (2018). Assessing the influence of a passive, upper extremity exoskeletal vest for tasks requiring arm elevation: Part II – "Unexpected" effects on shoulder motion, balance, and spine loading. *Applied Ergonomics*, *70*(March), 323–330. https://doi.org/10.1016/j.apergo.2018.02.024

Koopman, A. S., Näf, M., Baltrusch, S. J., Kingma, I., Rodriguez-Guerrero, C., Babič, J., de Looze, M. P., & van Dieën, J. H. (2020). Biomechanical evaluation of a new passive back support exoskeleton. *Journal of Biomechanics*, *105*. https://doi.org/10.1016/j.jbiomech.2020.109795

Koopman, A. S., Toxiri, S., Power, V., Kingma, I., Dieën, J. H. Van, Ortiz, J., & Looze, M. P. De. (2019). The effect of control strategies for an active back-support exoskeleton on spine loading and kinematics during lifting. *Journal of Biomechanics*, *91*, 14–22. https://doi.org/10.1016/j.jbiomech.2019.04.044

Krausz, N. E., Lamotte, D., Batzianoulis, I., Hargrove, L. J., Micera, S., & Billard, A. (2020). Intent Prediction Based on Biomechanical Coordination of EMG and Vision-Filtered Gaze for End-Point Control of an Arm Prosthesis. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, *28*(6), 1471–1480. https://doi.org/10.1109/TNSRE.2020.2992885

Lee, H., Kim, W., Han, J., & Han, C. (2012). The technical trend of the exoskeleton robot system for human power assistance. *International Journal of Precision Engineering and Manufacturing*, *13*(8), 1491–1497.

Lee, Y., Kim, Y. J., Lee, J., Lee, M., Choi, B., Kim, J., Park, Y. J., & Choi, J. (2017). Biomechanical Design of a Novel Flexible Exoskeleton for Lower Extremities. *IEEE/ASME Transactions on Mechatronics*, *22*(5), 2058–2069. https://doi.org/10.1109/TMECH.2017.2718999

Luger, T., Cobb, T. J., Seibt, R., Rieger, M. A., & Steinhilber, B. (2019). Subjective evaluation of a passive lower-limb industrial exoskeleton used during simulated assembly. *IISE Transactions on Occupational Ergonomics and Human Factors*, 1–10.

Madinei, S., Tech, V., Alemi, M. M., Kim, S., Srinivasan, D., & Nussbaum, M. A. (2020). Biomechanical Evaluation of Passive Back-Support Exoskeletons in a Precision Manual Assembly Task: "Expected" Effects on Trunk Muscle Activity, Perceived Exertion, and Task Performance. *HUMAN FACTORS*, *62*(3), 441–457. https://doi.org/10.1177/0018720819890966

Makinson, B. J. (1971). Research and Development Prototype for Machine Augmentation of Human Strength and Endurance - Hardiman I Project. In *GENERAL ELECTRIC CO SCHENECTADY NY SPECIALTY MATERIALS HANDLING PRODUCTS OPERATION.*

Otten, B. M., Weidner, R., & Argubi-Wollesen, A. (2018). Evaluation of a Novel Active Exoskeleton for Tasks at or above Head Level. *IEEE Robotics and Automation Letters*, *3*(3), 2408–2415. https://doi.org/10.1109/LRA.2018.2812905

Romero, D., Stahre, J., Wuest, T., Noran, O., Bernus, P., Fast-Berglund, Å., & Gorecky, D. (2016). Towards an Operator 4.0 Typology: A Human-Centric Perspective on the Fourth Industrial Revolution Technologies Digital Lean Manufacturing View project Produktion2030 View project TOWARDS AN OPERATOR 4.0 TYPOLOGY: A HUMAN-CENTRIC PERSPECTIVE ON THE FOURTH . *Researchgate.Net*, 29–31. https://www.researchgate.net/publication/309609488

Rosenthal, R., Cooper, H., Research, L. H.-T. handbook of, & 1994, U. (1994). Parametric measures of effect size. In *The handbook of research synthesis*. https://books.google.com/books?hl=ko&lr=&id=p-aFAwAAQBAJ&oi=fnd&pg=PA231&dq=hedges+d+effect+size&ots=TWxKOihFTD&sig=xscly9reZcvX61KSBQ-7RPU8Xmw

Russell, D. M., & Apatoczky, D. T. (2016). Walking at the preferred stride frequency minimizes muscle activity. *Gait and Posture*, *45*, 181–186. https://doi.org/10.1016/j.gaitpost.2016.01.027

Sanchez-Villamañan, M. D. C., Gonzalez-Vargas, J., Torricelli, D., Moreno, J. C., & Pons, J. L. (2019). Compliant lower limb exoskeletons: A comprehensive review on mechanical design principles. *Journal of NeuroEngineering and Rehabilitation*, *16*(1), 1–16. https://doi.org/10.1186/s12984-019-0517-9

Stirling, L., Kelty-Stephen, D., Fineman, R., Jones, M. L. H., Daniel Park, B. K., Reed, M. P., Parham, J., & Choi, H. J. (2020). Static, Dynamic, and Cognitive Fit of Exosystems for the Human Operator. *Human Factors*, *62*(3), 424–440. https://doi.org/10.1177/0018720819896898

Toxiri, S., Koopman, A. S., Lazzaroni, M., Ortiz, J., Power, V., de Looze, M. P., O'Sullivan, L., & Caldwell, D. G. (2018). Rationale, implementation and evaluation of assistive strategies for an active back-support exoskeleton. *Frontiers Robotics AI*, *5*(MAY), 1–14. https://doi.org/10.3389/frobt.2018.00053

Toxiri, S., Näf, M. B., Lazzaroni, M., Fernández, J., Sposito, M., Poliero, T., Monica, L., Anastasi, S., Caldwell, D. G., & Ortiz, J. (2019). Back-Support Exoskeletons for Occupational Use: An Overview of Technological Advances and Trends. *IISE Transactions on Occupational Ergonomics and Human Factors*, *7*(3–4), 237–249. https://doi.org/10.1080/24725838.2019.1626303

von Glinski, A., Yilmaz, E., Mrotzek, S., Marek, E., Jettkant, B., Brinkemper, A., Fisahn, C., Schildhauer, T. A., & Geßmann, J. (2019). Effectiveness of an on-body lifting aid (HAL® for care support) to reduce lower back muscle activity during repetitive lifting tasks. *Journal of Clinical Neuroscience*, *63*, 249–255. https://doi.org/10.1016/j.jocn.2019.01.038

Wei, W., Wang, W., Qu, Z., Gu, J., Lin, X., & Yue, C. (2020). The effects of a passive exoskeleton on muscle activity and metabolic cost of energy. *Advanced Robotics*, *34*(1), 19–27. https://doi.org/10.1080/01691864.2019.1707708

Whitfield, B. H., Costigan, P. A., Stevenson, J. M., & Smallman, C. L. (2014). Effect of an on-body ergonomic aid on oxygen consumption during a repetitive lifting task. *International Journal of Industrial Ergonomics*, *44*(1), 39–44. https://doi.org/10.1016/j.ergon.2013.10.002

Zoss, A. B., Kazerooni, H., & Chu, A. (2006). Biomechanical design of the Berkeley Lower Extremity Exoskeleton (BLEEX). *IEEE/ASME Transactions on Mechatronics*, *11*(2), 128–138. https://doi.org/10.1109/TMECH.2006.871087

TABLES

Table 1

Summary of ANOVA results [p value; (F statistic, v_1, v_2)] and η^2 for the effects of *Intervention* and *Load Mass* on median levels of normalized EMG (nEMG) of the load carriage task. Statistically significant effects are highlighted in bold. Tukey's HSD *post hoc* differences (pairwise comparisons between XO and no-XO conditions at each load level) were performed, and the resulting effect sizes are reported as *Hedge's g*. Large effect sizes (|g|>0.8) are highlighted in bold.

Muscle Group	Intervention (I)	η_{I}^2	Load Mass (L)	η_{L}^2	I x L	$\eta_{I\times L}^2$	*Hedge's g* Intervention effect at each load level (L1 - L5)
Left Thoracic Erector Spinae (LTES)	0.13 (2.39, 1, 36)	0.01	**<0.0001** **(17.20, 4, 36)**	**0.39**	**0.001** **(6.00, 4, 36)**	**0.14**	**L1: 0.97** L2: 0.22 L3: 0.29 **L4: 0.98** **L5: 1.28**
Right Thoracic Erector Spinae (RTES)	**0.004** **(9.21, 1, 36)**	**0.07**	**0.046** **(2.69, 4, 36)**	**0.09**	0.72 (0.52, 4, 36)	0.02	L1: 0.17 L2: 0.46 L3: 0.44 L4: 0.66 L5: 0.61
Left Lumbar Erector Spinae (LLES)	0.50 (0.46, 1, 45)	0.002	**<0.0001** **(17.17, 4, 45)**	**0.30**	**0.02** **(3.43, 4, 45)**	**0.06**	**L1: 0.98** L2: 0.45 L3: 0.38 L4: 0.18 L5: 0.70
Right Lumbar Erector Spinae (RLES)	0.89 (0.02, 1, 45)	0.0001	**<0.0001** **(15.85, 4, 45)**	**0.30**	**0.03** **(2.86, 4, 45)**	**0.06**	**L1: 1.12** L2: 0.39 L3: 0.04 L4: 0.26 L5: 0.71
Right Vastus Lateralis (RVL)	**0.01** **(6.41, 1, 45)**	**0.04**	**0.008** **(3.96, 4, 45)**	**0.11**	**0.01** **(3.60, 4, 45)**	**0.10**	**L1: 1.74** **L2: 1.53** L3: 0.47 L4: 0.02 L5: 0.37
Right Biceps Femoris (RBF)	**0.001** **(11.96, 1, 45)**	**0.08**	**<0.0001** **(14.37, 4, 45)**	**0.37**	0.28 (1.31, 4, 45)	0.03	L1: 0.14 L2: 0.50 L3: 0.27 **L4: 1.13** **L5: 1.55**
Right Tibialis Anterior (RTA)	**<0.0001** **(51.61, 1, 36)**	**0.04**	**0.04** **(2.88, 4, 36)**	**0.04**	0.16 (1.77, 4, 36)	0.02	**L1: 1.36** **L2: 0.92** **L3: 0.80** L4: 0.49 L5: 0.23
Right Medial Gastrocnemius (RMG)	0.55 (0.37, 1, 27)	0.002	**0.002** **(5.67, 4, 27)**	**0.12**	0.10 (2.14, 4, 27)	0.04	L1: 0.36 L2: 0.13 L3: 0.20 L4: 0.31 L5: 0.51

Table 2

Summary of ANOVA results [p value; (F statistic, ν_1, ν_2)] and η^2 for the effects of *Intervention* and *Load Mass* on median levels of nEMGs in the stationary load transfer task. Statistically significant effects are highlighted in bold. Tukey's HSD *post hoc* differences (pairwise comparisons between XO and no-XO conditions at each load level) were performed, and the resulting effect sizes are reported as *Hedge's g*. Large effect sizes (\|g\|>0.8) are highlighted in bold.

Muscle Group	Intervention (I)	η_{I}^2	Load Mass (L)	η_{L}^2	I x L	$\eta_{\mathrm{I\times L}}^2$	*Hedge's g* Intervention effect at each load level (L1 – L7)
Left Thoracic Erector Spinae (LTES)	**<0.0001 (29.78, 1, 52)**	**0.11**	**<0.0001 (17.27, 6, 52)**	**0.38**	**0.03 (2.62, 6, 52)**	**0.06**	L1: 0.66 L2: 0.56 L3: 0.61 L4: 0.64 **L5: 0.80** **L6: 0.90** **L7: 1.34**
Right Thoracic Erector Spinae (RTES)	**0.006 (8.25, 1, 56)**	**0.03**	**<0.0001 (24.42, 6, 56)**	**0.55**	0.10 (1.85, 6, 56)	0.04	L1: **1.69** L2: 0.50 L3: 0.11 L4: 0.20 L5: 0.57 L6: 0.49 **L7: 0.74**
Left Lumbar Erector Spinae (LLES)	**0.02 (5.68, 1, 62)**	**0.01**	**<0.0001 (42.54, 6, 62)**	**0.43**	**0.0003 (5.10, 6, 62)**	**0.05**	L1: 0.73 L2: 0.33 L3: 0.09 L4: 0.10 L5: 0.38 L6: 0.74 L7: 0.78
Right Lumbar Erector Spinae (RLES)	**0.001 (11.73, 1, 62)**	**0.05**	**<0.0001 (18.55, 6, 62)**	**0.45**	**0.03 (2.59, 6, 62)**	**0.06**	L1: 0.43 L2: 0.09 L3: 0.10 L4: 0.43 L5: 0.78 **L6: 1.12** **L7: 0.88**
Right Vastus Lateralis (RVL)	**0.04 (4.56, 1, 58)**	**0.03**	**<0.0001 (5.90, 6, 58)**	**0.26**	**0.03 (2.58, 6, 58)**	**0.11**	L1: 0.40 L2: 0.05 L3: 0.08 L4: 0.34 L5: 0.06 **L6: 0.86** **L7: 0.93**
Right Biceps Femoris (RBF)	0.09 (3.00, 1, 62)	0.01	**<0.0001 (17.09, 6, 62)**	**0.48**	**0.033 (2.47, 6, 62)**	**0.07**	**L1: 1.02** L2: 0.48 L3: 0.26 L4: 0.31 L5: 0.21 L6: 0.24 **L7: 1.09**
Right Tibialis Anterior (RTA)	0.21 (1.61, 1, 51)	0.01	**<0.0001 (7.55, 6, 51)**	**0.25**	0.21 (1.45, 6, 51)	0.05	**L1: 1.13** L2: 0.22 L3: 0.08 L4: 0.15 L5: 0.08 L6: 0.37 L7: 0.49
Right Medial Gastrocnemius (RMG)	**0.03 (5.14, 1, 39)**	**0.03**	**<0.0001 (14.42, 6, 39)**	**0.46**	**0.01 (3.11, 6, 39)**	**0.10**	L1: 0.29 L2: 0.23 L3: 0.38 L4: 0.08 L5: 0.15 **L6: 0.88** **L7: 1.04**

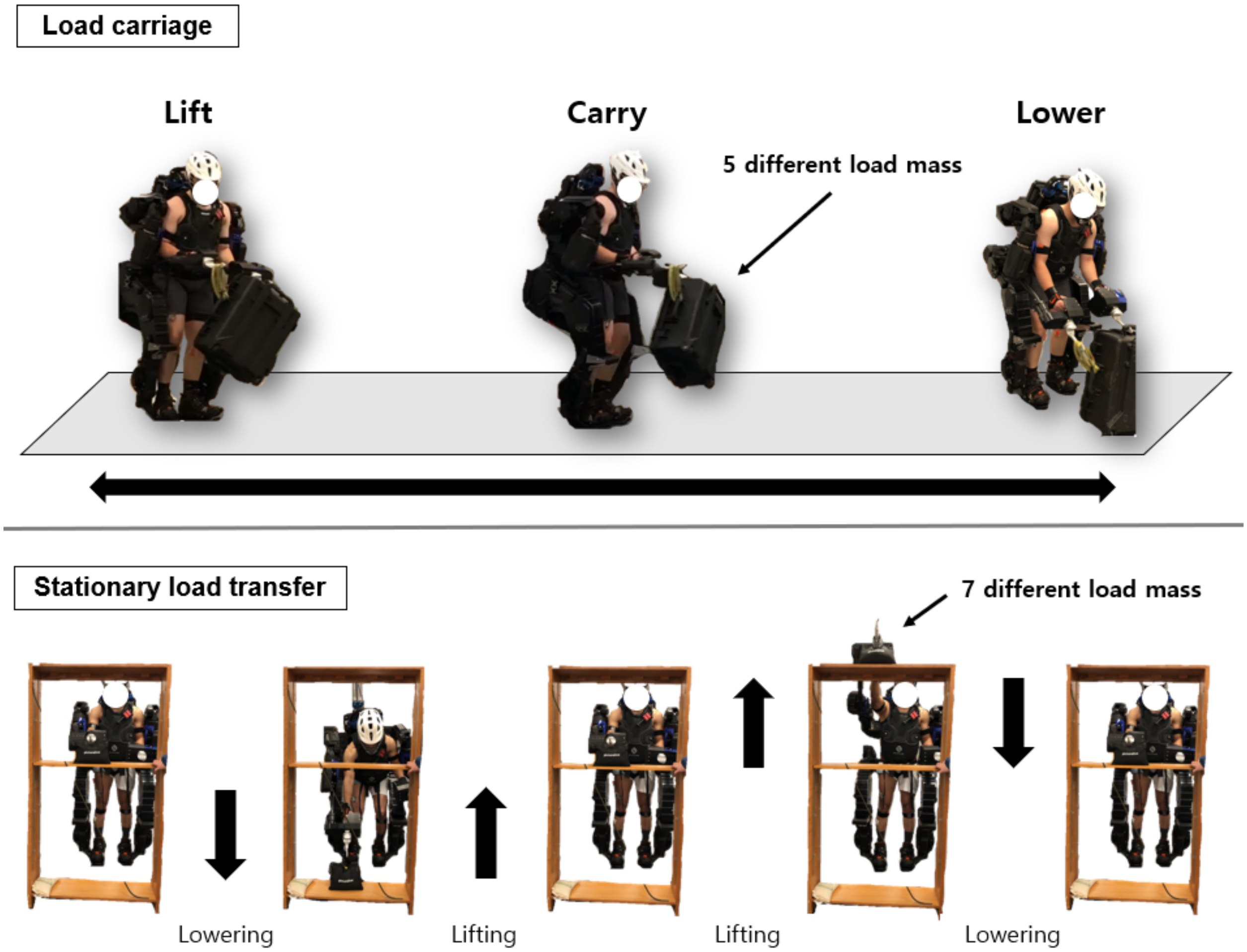
FIGURE CAPTIONS

Figure 1. Illustrations of the load carriage (Top) and stationary load transfer tasks (Bottom). Participants performed three replications each of the load carriage task (involving a 15m round trip) and the stationary load transfer task.

Figure 2. Muscle activity (normalized EMG = nEMG) during the load carriage task in two bilateral muscle groups monitored in the back (Top) and four muscle groups in the leg (Bottom). Each data point represents the mean value of median nEMG across participants at each load mass. Hatched and grey regions indicate ± 1 standard deviation (SD) in the WB-PEXO and control conditions, respectively. Dashed lines denote significant main effects of *Intervention*, and the symbol * indicates a significant paired difference between WB-PEXO and control conditions at a given load mass.

Figure 3. Muscle activity (normalized EMG = nEMG) during the stationary load transfer task in two bilateral muscle groups monitored in the back (Top) and four muscle groups in the leg (Bottom). Each data point represents the mean value of median nEMG across participants at each load mass. Hatched and grey regions indicate ± 1 standard deviation (SD) in the WB-PEXO and control conditions, respectively. Dashed lines denote significant main effects of *Intervention*, and the symbol * indicates a significant paired difference between WB-PEXO and control conditions at a given load mass.

FIGURES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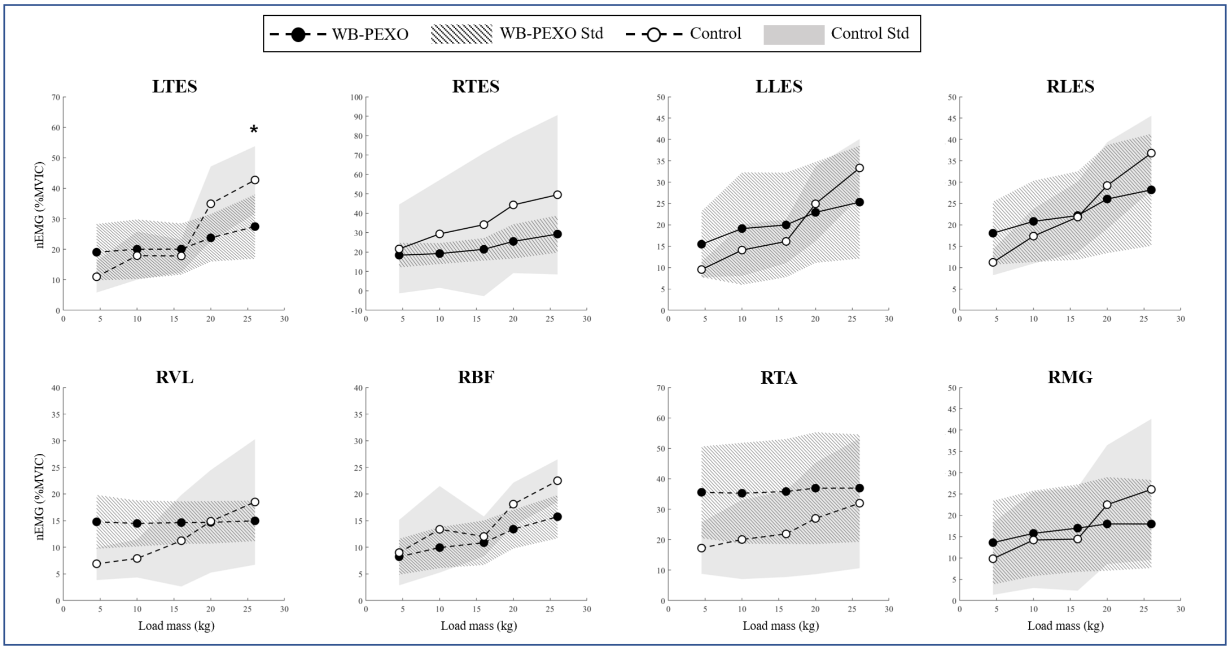
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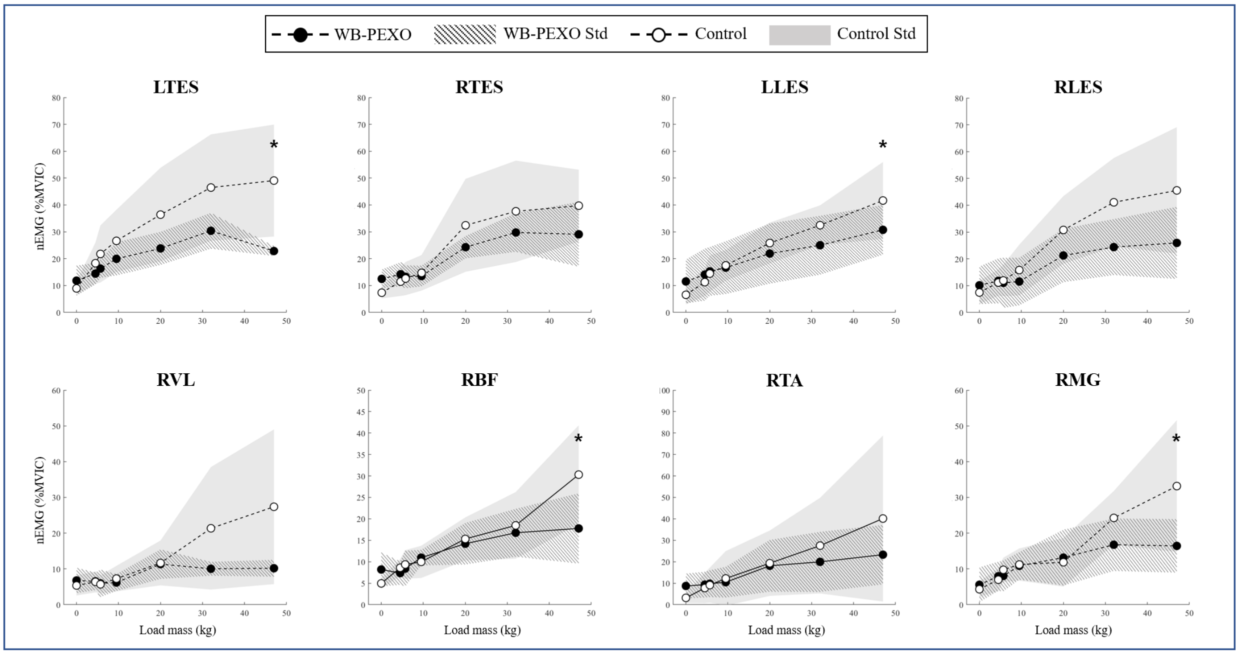
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APPENDICES

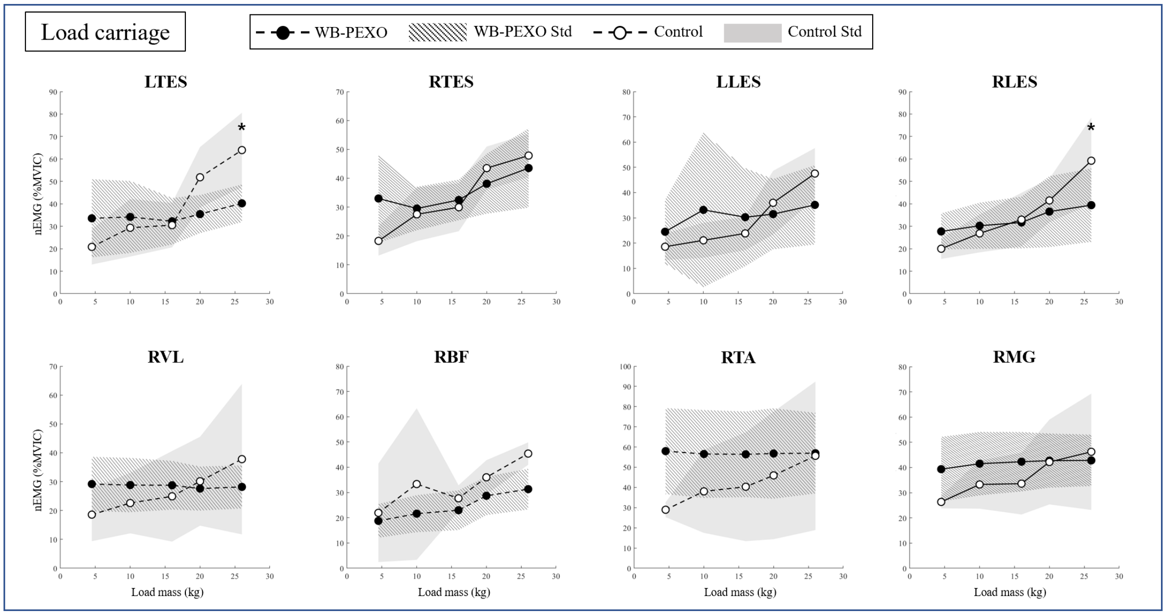

Figure A1. Peak normalized muscle activity (nEMG) during the <u>load carriage task</u> in two bilateral muscle groups monitored in the back (Top) and four muscle groups in the leg (Bottom). Each data point represents the peak value of median nEMG across participants at each load mass. Hatched and grey regions indicate ± 1 standard deviation (SD) in the WB-PEXO and control conditions, respectively. Dashed lines denote significant main effects of *Intervention*, and the symbol * indicates a significant paired difference between WB-PEXO and control conditions at a given load mass.

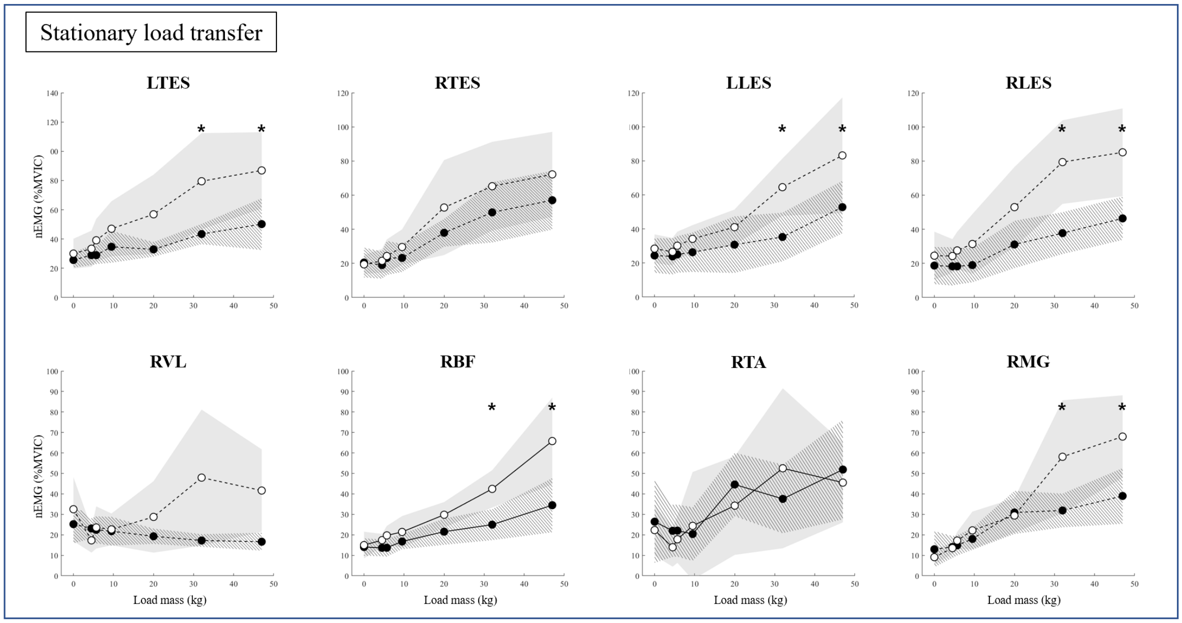

Figure A2. Peak normalized muscle activity (nEMG) during the <u>stationary load transfer task</u> in two bilateral muscle groups monitored in the back (Top) and four muscle groups in the leg (Bottom). Each data point represents the peak value of median nEMG across participants at each load mass. Hatched and grey regions indicate ± 1 standard deviation (SD) in the WB-PEXO and control conditions, respectively. Dashed lines denote significant main effects of *Intervention*, and the symbol * indicates a significant paired difference between WB-PEXO and control conditions at a given load mass.

Table A1

Summary of ANOVA results [*p* value; (*F* statistic, ν_1, ν_2)] and η^2 for the effects of *Intervention* and *Load Mass* on peak levels of nEMGs in the load carriage task. Statistically significant effects are highlighted in bold. Tukey's HSD *post hoc* differences (pairwise comparisons between XO and no-XO conditions at each load level) were performed, and the resulting effect sizes are reported as *Hedge's g*. Large effect sizes (|*g*|>0.8) are highlighted in bold.

Muscle Group	Intervention (I)	η^2_{I}	Load Mass (L)	η^2_{L}	I x L	$\eta^2_{\mathrm{I\times L}}$	*Hedge's g* Intervention effect at each load level (L1 - L5)
Left Thoracic Erector Spinae (LTES)	0.21 (1.62, 1, 35)	0.01	**<0.0001 (10.38, 4, 35)**	**0.33**	**0.002 (5.19, 4, 35)**	**0.17**	**L1: 0.85** L2: 0.30 L3: 0.15 **L4: 1.26** **L5: 1.63**
Right Thoracic Erector Spinae (RTES)	0.16 (2.08, 1, 31)	0.01	**<0.0001 (17.75, 4, 31)**	**0.40**	**0.01 (3.92, 4, 31)**	**0.09**	**L1: 1.08** L2: 0.21 L3: 0.30 L4: 0.52 L5: 0.34
Left Lumbar Erector Spinae (LLES)	0.63 (0.24, 1, 45)	0.002	**0.002 (4.92, 4, 45)**	**0.18**	0.10 (2.05, 4, 45)	0.07	L1: 0.56 L2: 0.50 L3: 0.41 L4: 0.31 **L5: 0.87**
Right Lumbar Erector Spinae (RLES)	0.23 (1.47, 1, 45)	0.01	**<0.0001 (13.00, 4, 45)**	**0.34**	**0.01 (3.67, 4, 45)**	**0.10**	**L1: 1.09** L2: 0.34 L3: 0.10 L4: 0.35 **L5: 1.03**
Right Vastus Lateralis (RVL)	0.52 (0.42, 1, 45)	0.004	0.24 (1.42, 4, 45)	0.06	0.14 (1.83, 4, 45)	0.07	**L1: 1.05** L2: 0.58 L3: 0.28 L4: 0.19 L5: 0.46
Right Biceps Femoris (RBF)	**0.008 (7.57, 1, 45)**	**0.08**	**0.005 (4.25, 4, 45)**	**0.19**	0.75 (0.48, 4, 45)	0.02	L1: 0.20 L2: 0.50 L3: 0.66 **L4: 0.94** **L5: 2.00**
Right Tibialis Anterior (RTA)	**0.0002 (17.45, 1, 36)**	**0.10**	0.25 (1.40, 4, 36)	0.03	0.20 (1.58, 4, 36)	0.04	**L1: 1.72** **L2: 0.79** L3: 0.60 L4: 0.36 L5: 0.04
Right Medial Gastrocnemius (RMG)	**0.04 (4.59, 1, 27)**	**0.04**	0.06 (2.63, 4, 27)	0.10	0.26 (1.39, 4, 27)	0.05	**L1: 1.23** L2: 0.64 L3: 0.63 L4: 0.03 L5: 0.17

Table A2

Summary of ANOVA results [p value; (F statistic, ν_1, ν_2)] and η^2 for the effects of *Intervention* and *Load Mass* on peak levels of nEMGs in the stationary load transfer task. Statistically significant effects are highlighted in bold. Tukey's HSD *post hoc* differences (pairwise comparisons between XO and no-XO conditions at each load level) were performed, and the resulting effect sizes are reported as *Hedge's g*. Large effect sizes (\|g\|>0.8) are highlighted in bold.

Muscle Group	Intervention (I)	η^2_{I}	Load Mass (L)	η^2_{L}	I x L	$\eta^2_{\mathrm{I \times L}}$	*Hedge's g* Intervention effect at each load level (L1 – L7)
Left Thoracic Erector Spinae (LTES)	**<0.0001 (49.39, 1, 49)**	**0.30**	**<0.0001 (15.98, 6, 49)**	**0.15**	**0.04 (2.41, 6, 49)**	**0.04**	L1: 0.46 L2: 0.39 **L3: 0.80** L4: 0.71 **L5: 0.93** **L6: 1.14** **L7: 1.25**
Right Thoracic Erector Spinae (RTES)	**0.0001 (17.39, 1, 58)**	**0.05**	**<0.0001 (31.59, 6, 58)**	**0.56**	0.13 (1.73, 6, 58)	0.03	L1: 0.13 L2: 0.31 L3: 0.11 L4: 0.62 L5: 0.63 L6: 0.62 L7: 0.61
Left Lumbar Erector Spinae (LLES)	**<0.0001 (49.89, 1, 62)**	**0.10**	**<0.0001 (31.15, 6, 62)**	**0.38**	**<0.0001 (6.11, 6, 62)**	**0.07**	L1: 0.40 L2: 0.26 L3: 0.49 L4: 0.73 L5: 0.69 **L6: 1.73** **L7: 0.98**
Right Lumbar Erector Spinae (RLES)	**<0.0001 (43.21, 1, 62)**	**0.14**	**<0.0001 (23.25, 6, 62)**	**0.46**	**0.004 (3.65, 6, 62)**	**0.07**	L1: 0.42 L2: 0.53 **L3: 0.80** **L4: 0.80** **L5: 1.05** **L6: 1.99** **L7: 1.63**
Right Vastus Lateralis (RVL)	0.0005 (13.46, 1, 61)	0.11	0.10 (1.86, 6, 61)	0.09	0.004 (3.51, 6, 61)	0.17	L1: 0.52 **L2: 0.98** L3: 0.12 L4: 0.11 L5: 0.69 **L6: 1.20** **L7: 1.49**
Right Biceps Femoris (RBF)	**<0.0001 (35.93, 1, 62)**	**0.11**	**<0.0001 (27.41, 6, 62)**	**0.51**	**0.0007 (4.58, 6, 62)**	**0.08**	L1: 0.16 **L2: 0.96** **L3: 1.24** L4: 0.71 **L5: 1.21** **L6: 1.93** **L7: 1.53**
Right Tibialis Anterior (RTA)	0.75 (0.099, 1, 50)	0.0004	**<0.0001 (11.54, 6, 50)**	**0.31**	0.28 (1.30, 6, 50)	0.03	L1: 0.22 L2: 0.65 L3: 0.32 L4: 0.17 L5: 0.46 L6: 0.45 L7: 0.25
Right Medial Gastrocnemius (RMG)	**0.002 (11.25, 1, 39)**	**0.04**	**<0.0001 (27.49, 6, 39)**	**0.60**	**0.001 (4.73, 6, 39)**	**0.10**	L1: 0.50 L2: 0.11 L3: 0.44 L4: 0.49 L5: 0.14 **L6: 1.12** **L7: 1.47**