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Musculoskeletal models determine the effect of a soft active exosuit on muscle activations and forces during lifting and lowering tasks

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ABSTRACT

Exosuits have the potential to mitigate musculoskeletal stress and prevent back injuries during industrial tasks. This study aimed to 1) validate the implementation of a soft active exosuit into a musculoskeletal model of the spine by comparing model predicted muscle activations versus corresponding surface EMG measurements, and 2) evaluate the effect of the exosuit on peak back and hip muscle forces. Fourteen healthy participants performed squat and stoop lift and lower tasks with boxes of 6 and 10 kg, with and without wearing a 2.7 kg soft active exosuit. Participant-specific musculoskeletal models, which included the exosuit, were created in OpenSim. Model validation focused on the back and hip extensors, where temporal agreement between EMG and model estimated muscle activity was generally strong to excellent (average cross-correlation coefficients ranging from 0.84 to 0.98). Root mean square errors of muscle activity (0.05–0.10) were similar with and without the exosuit, and compared well to prior model validation studies without the exosuit (average root mean square errors ranging from 0.05 to 0.19). In terms of performance, the exosuit reduced the estimated peak erector spinae forces during lifting and lowering phases across all lifting tasks but reduced peak hip extensor muscles forces only in a squat lift task of 10 kg. These reductions in total peak muscle forces were approximately 1.7–4.2 times greater than the corresponding exosuit assistance force, which were 146 ± 19 N and 102 ± 14 N at the times of peak erector spinae forces in lifting and lowering, respectively. Overall, the results support the hypothesis that exosuits reduce soft tissue loading, and thereby potentially reduce fatigue and injury risk during manual materials handling tasks. Incorporating exosuits into musculoskeletal models is a valid approach to understand the impact of exosuit assistance on muscle activity and forces.

1. Introduction

Workers who perform manual materials handling tasks often experience high internal loads on their backs and are more likely to develop back pain (Ferguson and Marras, 2013; Norman et al., 1998). Exosuits or exoskeletons are a promising approach for reducing musculoskeletal demands during lifting tasks. There are numerous back exosuit designs, with various strengths and weaknesses (Ali et al., 2021; Huysamen et al., 2018; Kermavnar et al., 2021), but a key goal of most exosuits is to reduce muscle loads and the risk of injury or fatigue (Kermavnar et al., 2021; Zelik et al., 2022). Prior work has shown that exosuits can reduce back and hip extensor biological moment demands and muscle activity by 6–48 % (Bär et al., 2021; Lamers et al., 2018; Quirk et al., 2024), suggesting reduced internal demands. Quantification of exosuit effects on tissue loading could provide important insights and support improved design and performance.

Musculoskeletal modeling can quantify tissue loading, and some studies have used them to evaluate the effects of back exosuits on musculoskeletal loads. A simple single-muscle equivalent model supported the feasibility of a parallel force application design, suggesting a disproportionate reduction in back extensor muscle forces leading to a net reduction in spinal compression loads (Abdoli-Eramaki et al., 2007). Subsequent modeling studies have confirmed reductions of *in vivo* spine loading using various exosuits (for reviews see: Ali et al., 2021;

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Kermavnar et al., 2021). However, model validation by comparison of model outcomes such as muscle activity or musculoskeletal loads to corresponding direct or indirect measurements is critical for appropriate interpretation of model estimations (Hicks et al., 2015; Zander et al., 2016). Prior models used to evaluate back exosuits have not had their validity confirmed during the use of an exosuit.. Moreover, the effects on various muscle groups, including the relative efficiency of applied exosuit force in reducing muscle forces, has not been reported. Investigating this with a detailed musculoskeletal model may provide important insights for exosuit design or control features to improve usability and effectiveness.

Therefore, the objective of this study is to 1) confirm the validity of a previously presented musculoskeletal model with an exosuit (Banks et al., 2024) by comparing predicted muscle activations with directly recorded electromyography (EMG) measurements; and 2) evaluate the effect of the exosuit on model-estimated back and hip extensor muscle forces relative to the assistance force. We hypothesize that model-estimated muscle activations will correspond well with recorded EMG, with and without the exosuit, and that the exosuit will reduce back and hip extensor muscle forces.

2. Methods

Participant, experimental setup and design, data processing, and musculoskeletal modeling details of this study have been presented previously (Banks et al., 2024; Chung et al., 2024; Quirk et al., 2024). Therefore, only key information is presented here.

2.1. Participants

A priori sample size calculation determined that 14 participants were required to establish significant differences with an 80 % power and alpha of 0.05 (Quirk et al., 2023). Healthy adults from the local community who were 18–55 years of age, physically active, and reported no history of musculoskeletal or neurological disorders or recent back pain were recruited to participate. All participants provided written consent to a protocol approved by the Institutional Review Board of Harvard Medical School (IRB18-0960).

2.2. Soft active back exosuit

A soft active exosuit was worn by participants during portions of the testing (Chung et al., 2024; Quirk et al., 2023, 2024). The exosuit consisted of a backpack connected via an actuated strap to bilateral cuffs secured to the upper thighs (Fig. 1). The assistive force (maximum 250 N) was controlled based on embedded inertial measurement units (IMUs) and recorded by an inline load cell (LSB200, FUTEK Advanced Sensor Technology Inc., CA, USA).

2.3. Experimental procedure

Participants performed box lifting tasks in eight scenarios comprised of two permutations of box mass (6 or 10 kg), lift type (stoop or squat-style), and exosuit condition (exosuit or no exosuit). Full-body kinematics were recorded (Qualisys AB, Goteborg, Sweden) by 49 motion capture markers (Fig. 1). Each task consisted of a 2.5 s lift phase (bending to pick up the box and returning to standing) and a 2.5 s lowering phase (bending to put down the box and returning to standing), performed for 10 consecutive repetitions, with 7 s of rest between each repetition. Timing of the lift and lowering phases was paced with an audible 50-beats-per-minute metronome.

2.4. Electromyography (EMG) collection and processing

Surface EMG (Delsys Duo; Delsys Inc., USA) recorded (2148 Hz) muscle activity at five sites including 1) the thoracic erector spinae (ES) at 6 cm lateral to the 9th thoracic spinous process (T9); 2) lumbar ES at 6 cm lateral to the 1st lumbar spinous process (L1); 3) lumbar ES at 3 cm lateral to the 3rd lumbar spinous process (L3); 4) the gluteus muscles; and 5) the biceps femoris. The gluteus and biceps femoris were placed according to SENIAM guidelines (Hermens et al., 1999). All tasks were symmetric in the sagittal plane; therefore, activations were only collected unilaterally from the right side of the body. Off-line, muscle activations were bandpass filtered (50–450 Hz), full-wave rectified, 6 Hz low pass filtered, normalized to the peak from two maximal voluntary contraction postures performed against mechanical static restraints, time-normalized over 201 data points (0–100 %; see Data Analysis), and ultimately calculated as an ensemble average of all repetitions of the task for each muscle (Quirk et al., 2024).

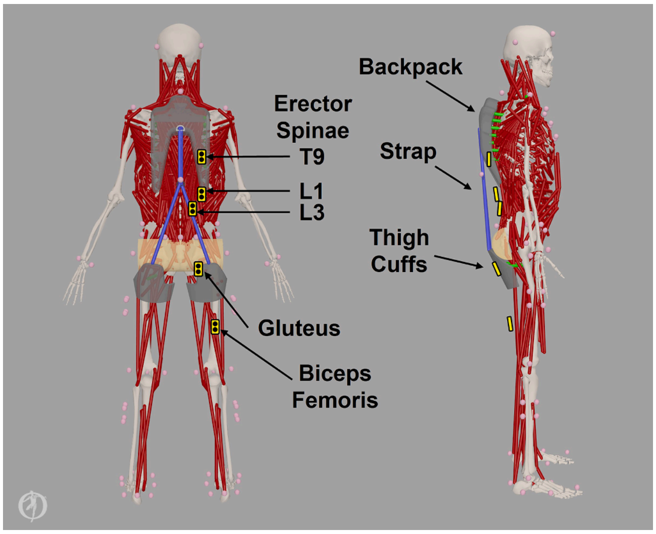

Fig. 1. Rear (left) and sagittal (right) view of the musculoskeletal model with exosuit. Light pink spheres and yellow rectangles represent the motion tracking markers and electromyography sensor placements, respectively. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

2.5. Musculoskeletal models

An OpenSim full-body thoracolumbar model without an exosuit was used to model the no exosuit conditions. This model has been previously validated for estimating spinal loads and muscle exertion during static and dynamic tasks without an exosuit (Akhavanfar et al., 2024; Alemi et al., 2023; Bruno et al., 2015, 2017). To incorporate the exosuit into this model, a rigid body was connected to the T6 vertebrae via a custom joint, representing the exosuit pack of mass 2.4 kg (Fig. 1). The model also included the exosuit's actuated strap spanning from thigh cuffs to the exosuit pack, splitting off at a buckle and wrapping around the gluteus to estimate a realistic path. The exosuit assistance force was assigned as a uniform tension between rigid bodies and promoted trunk and hip extension. Point actuators were also positioned between the exosuit and the ribs, shoulders, and thighs to represent potential contact pressures (for additional model details, see Banks et al., 2024). The handled box was modeled by attaching a rigid body of half the measured mass and estimated inertial moment properties of the actual box to each hand (Akhavanfar et al., 2022).

Participant-specific models were constructed from the appropriate male or female base model scaled according to weight and marker calibration poses (Burkhart et al., 2020), and trunk muscle size and position adjusted based on regressions accounting for age, sex, height, and weight (Allaire et al., 2023). Spine curvatures were further personalized using calibration pose IMU data (Banks et al., 2024). Body and joint kinematics were computed using the OpenSim Inverse

Kinematics Tool to fit the marker positions to each participant-specific model (Delp et al., 2007). Coordinate coupling constraints were applied to the model to reduce the fifty-one degrees of freedom associated with the spine to a determinate three (Alemi et al., 2021). All modeled kinematics were filtered with a zero-lag low-pass (4 Hz) fourth-order Butterworth filter (Winter, 2009). Muscle activations and forces from ES and gluteus muscles were estimated from the tracked kinematics using the OpenSim Static Optimization Tool (Crowninshield and Brand, 1981). During exosuit trials, the recorded exosuit assistive force was input to the exosuit's actuated strap in the model. Custom MATLAB (The MathWorks Inc., USA) API scripts were used to run each modeling step (Lee and Umberger, 2016).

2.6. Data analysis

For each task repetition, the starting time was defined from when the C7 marker velocity began to exceed 40 mm/s prior to picking up the crate, and ended when the marker velocity was less than 40 mm/s upon returning to standing without the crate. The lifting and lowering phases were then defined as the first and last half of each repetition, respectively. Due to the time it took to process trials and the repetitive nature of the lifting tasks, only the fifth through ninth lift (five in total) repetitions were modeled.

Model outcomes of interest were muscle activities and muscle forces during the lifting trials. Modeled muscle activities were calculated by averaging the activities across the appropriate model muscle fascicles located near the nominal location of each EMG sensor (Alemi et al., 2023). Specifically, ES activity of the iliocostalis muscle fascicles at T9, iliocostalis muscle fascicles at L1, and all ES muscle fascicles at L3 were averaged to compare with the recorded T9, L1, L3 activities, respectively. The activity of the three gluteus maximus fascicles plus the most posterior fascicle of the gluteus medius were averaged, and the activity of the biceps femoris long head and short head fascicles were averaged to compare with the recorded gluteus and biceps femoris activities, respectively. Total ES muscle force was calculated by summing all ES fascicles, and total hip extensor muscle force by summing all the gluteus and the biceps femoris long head fascicles. Bilateral model-based activations and muscle forces were averaged over the five modeled repetitions, and time-normalized to match the measured EMG.

The validity of the exosuit model was quantified in two ways (Alemi et al., 2023). First, the maximum absolute normalized cross-correlation (MANCC) coefficient was calculated to assess the temporal similarity of the recorded and modeled muscle activity. MANCC values were interpreted as >0.90, >, 0.70, and >0.40 indicating excellent, strong, and moderate similarity, respectively. The lag between signals for the MANCC was reported as a percentage of the overall task motion. Second, to assess the similarity between activation magnitudes, the root mean square error (RMSE) between the modeled and EMG-measured muscle activity was calculated, while correcting for the lag-time. MANCC and RMSE values for all muscle groups were calculated for each participant and averaged for all lifting and lowering scenarios.

Mixed effects regressions evaluated the effects of exosuit condition (yes / no), lift type (squat / stoop), and box mass (6 kg / 10 kg) on peak total muscle forces of the ES and the hip extensors for both lifting and lowering. Participants were modeled as a random effect. Linear combinations of coefficients were evaluated to estimate the effect of the exosuit on peak muscle forces. Corresponding exosuit forces from the recorded exosuit load cell data at the times of peak muscle forces were also extracted and used to determine timing and response differences with peak exosuit and muscle forces (as a percentage of lift/lower). Statistical analyses were performed in Stata/IC 13.1 (StataCorp LP, College Station, TX), with significance set at $\alpha = 0.05/4 = 0.0125$ to account for four mixed effects regressions performed (for ES and hip extensor muscle forces, in lifting and lowering separately).

3. Results

3.1. Participant demographics

Fourteen participants (10 men, 4 women) consented and successfully completed the experimental protocol. The average ± standard deviation participant was 1.75 ± 0.09 m in height, 75.7 ± 13.4 kg in mass, and 31 ± 4 years of age.

3.2. Model validation

The model-predicted temporal patterns and magnitudes matched measured activations relatively well across the five muscles (Fig. 2). MANCC values were generally excellent in lifting, and strong to excellent in lowering (Fig. 3). The lowering phase had several instances of MANCC below 0.7 in the gluteus muscles and the biceps femoris when performing stoop lifts (Fig. 3). Similar temporal agreement was observed between all eight lifting scenarios. The EMG-measured activations led model-estimated activations by 0.3 ± 3.5 % of the motion for T9, but lagged by 1.0 ± 4.0 % for L1, 2.4 ± 3.9 % for L3, 7.2 ± 3.1 % for the gluteus muscles, and 7.1 ± 5.4 % for the biceps femoris (Fig. 2). Thus, model activations tended to peak earlier than their corresponding measured activations, especially the gluteus muscles and biceps femoris.

Most RMSEs between EMG-measured and model-estimated muscle activation were between 0.05 and 0.10 (Fig. 4). Similar magnitudes were observed between stoop and squat lifts, across crate masses, lifting and lowering portions of the task, and with and without the exosuit for most muscles. Larger RMSEs were seen for the ES L3 spinal level sensor location, and the biceps femoris during stoop lifts. Model-predicted activations were generally lower than EMG-measured values at the ES L3 sensor location, but larger than measured values for the biceps femoris (Fig. 2.).

3.3. Impact and timing of exosuit on muscle forces

Peak model-estimated ES force (Fig. 5) was reduced by the exosuit, increased with the larger box mass in both lifting and lowering phases (all $p < 0.001$), and was slightly larger in squat type than stoop type for

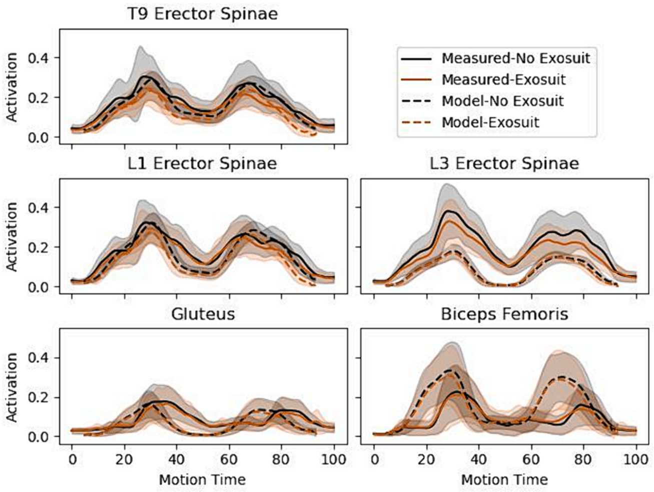

Fig. 2. Average muscle activity waveforms from the five muscles, showing EMG measured (*solid lines*) and model-estimated (*dashed lines*) activity with (*orange*) and without (*black*) exosuit assistance. Shaded areas represent the standard deviations about the average at each time point. All four lifting scenarios had relatively similar waveforms, thus they were combined for the current figure. The lag between model-estimated and measured activity can be clearly seen for the gluteus and biceps femoris muscles. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

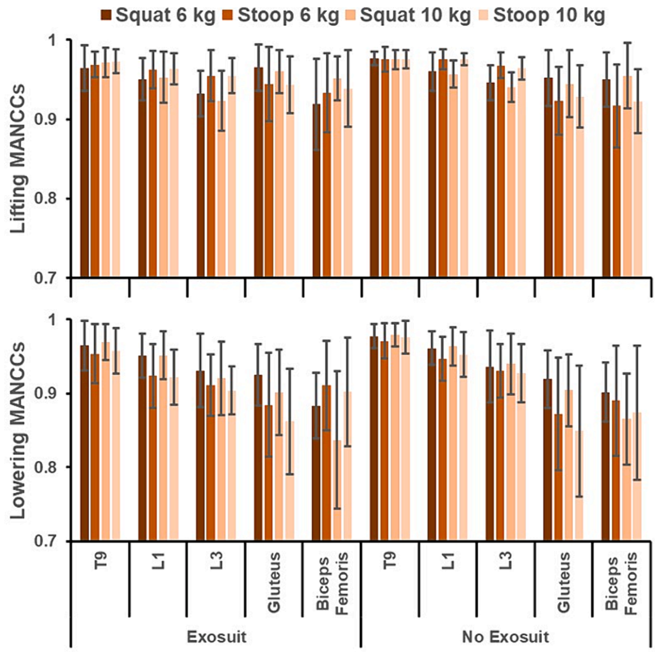

Fig. 3. Cross-correlation coefficients (MANCCs) between measured and model-estimated muscle activity in lifting (*upper panel*) and lowering (*lower panel*) phases of each lift. Data whiskers represent the standard deviation about the mean value.

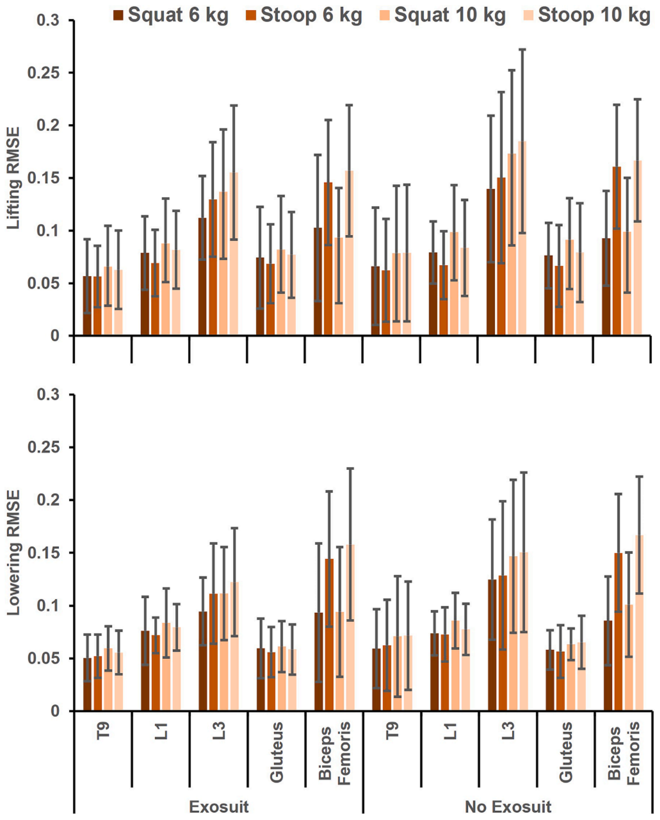

Fig. 4. Root mean square error (RMSE) between measured and model-estimated muscle activity in lifting (*upper panel*) and lowering (*lower panel*) phases of each lift. Data whiskers represent the standard deviation about the mean value.

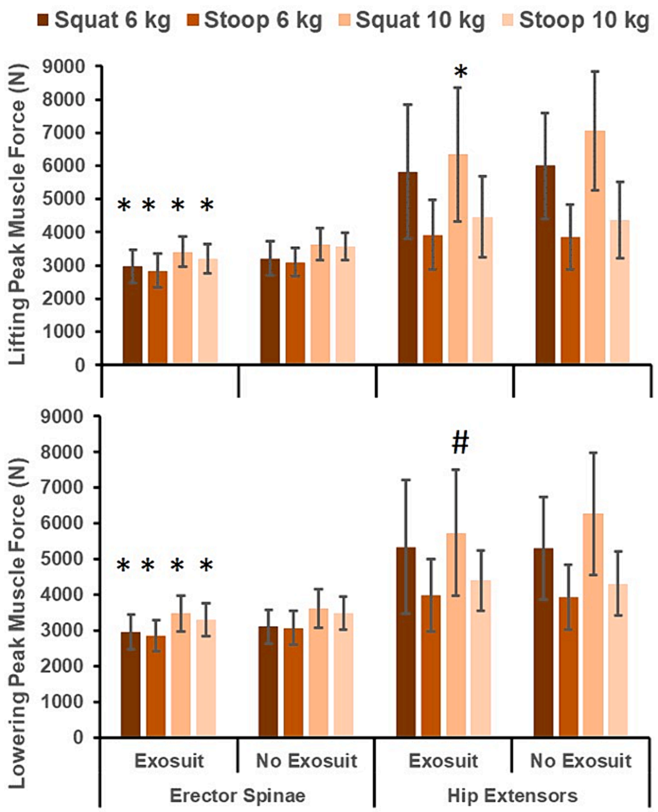

Fig. 5. Model-estimated peak muscle forces of the erector spinae and hip extensor muscles in the lift and lowering phases of each lift. Significant differences from corresponding "No Exosuit" conditions are indicated (* p-value $<$ 0.0125; # p-value $=$ 0.024). Data whiskers represent the standard deviation about the mean value.

lifting (p =.001) but not lowering ($p = 0.167$). There was an interaction of exosuit with lift type in lifting ($p = 0.012$) but not lowering ($p = 0.101$). There was no interaction between exosuit and box mass (lifting $p = 0.088$; lowering $p = 0.602$). Estimated reductions in peak ES forces (Fig. 6) were significant for all lift types in both lifting and lowering phases (all $p < 0.001$). During lifting, peak ES muscle forces were reduced 274 ± 50 N, or ~1.9× the concurrently applied 147 ± 18 N exosuit force. Delivery of peak exosuit force occurred after peak muscle force by approximately 12 ± 4 % of the lift cycle, or 300 ± 100 ms. During lowering, peak ES muscle forces were reduced 173 ± 56 N, or ~1.7× more than the concurrent 101 ± 14 N exosuit force. Peak exosuit force occurred after peak muscle force, by about 29 ± 8 % of the lowering phase, or 725 ± 200 ms.

Peak model-estimated hip extensor force (Fig. 5) was not universally affected by the exosuit (lifting $p = 0.296$; lowering $p = 0.236$) but was larger for squats than stoops and increased with the larger box mass for both lifting and lowering phases (all $p < 0.0125$). No significant interaction of exosuit with box mass was found for lifting ($p = 0.256$) or lowering ($p = 0.179$). An interaction of exosuit with lift type neared significance in lifting ($p = 0.021$), but not lowering ($p = 0.103$). Estimated reductions in peak hip extensor forces for squat with 10 kg were significant during lifting ($p = 0.003$) and approached significance during lowering ($p = 0.024$) (Fig. 5), but there were no differences for squat with 6 kg or stoop lifts (all $p > 0.0125$). During squat lifts with 10 kg, hip extensor force was reduced by 576 ± 385 N, which was ~4.2× more than the concurrent 137 ± 7N exosuit force (Fig. 6). Peak exosuit force during lifting was delivered after peak hip extensor force by 12 ± 6 % of the lifting phase, or 300 ± 150 ms. For squat type with 10 kg during lowering, hip extensor force was reduced by 389 ± 338 N which was ~3.5× more than the concurrent 110 ± 11 N exosuit force. Peak exosuit

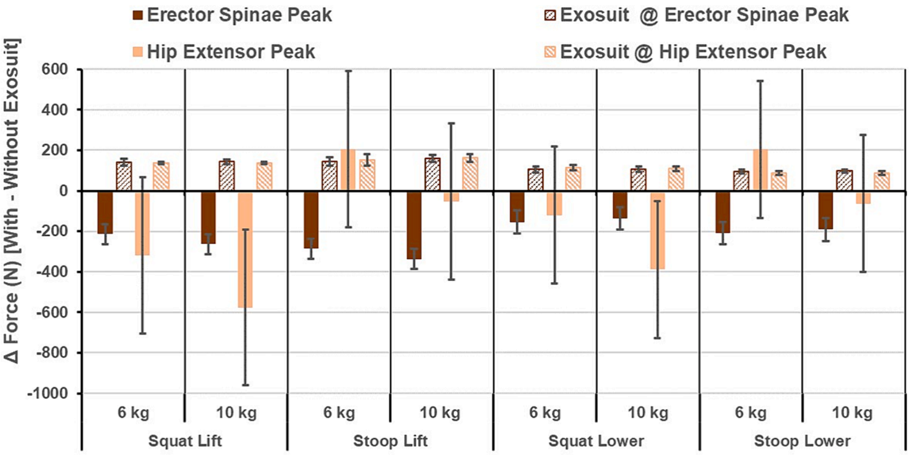

Fig. 6. Point estimate changes in erector spinae and hip extensor peak forces due to exosuit use, calculated from mixed-effects models. Exosuit forces applied at corresponding timepoints are shown for comparison. Data whiskers represent the standard deviation about the mean value.

force during lowering was delivered after peak hip extensor force by 30 ± 7 % of the lowering phase, or 750 ± 175 ms.

4. Discussion

This study validated a musculoskeletal model for assessing the biomechanical impact of a back support exosuit and its effect on muscle forces. The model accurately estimated most back extensor muscle activations during lifting tasks with the exosuit. The exosuit also significantly reduced the estimated peak spinal muscle forces during lifting and lowering phases across all lifting tasks (Fig. 6). For the hip extensors, the model matched temporal patterns of activation but with a time offset, peaking earlier than the measured muscle activations. The exosuit reduced peak muscle forces for the hip extensors only during squat type lifts, particularly with 10 kg.

Overall, the muscle activity magnitudes from the model were well-aligned with measurements, particularly for the ES at T9 and L1 levels, while the magnitudes estimated at the L3 level were less accurate (Figs. 2 and 3), with a systematic underprediction. These results are similar to a prior validation study using this model in older adults (Alemi et al., 2023). The lesser accuracy at L3 may arise from a limitation in the model or in the methods used to combine activity of multiple model muscle fascicles into an EMG-comparable signal, which have not been carefully explored (Alemi et al., 2023). Moreover, the inherent difficulty of accurately normalizing EMG (Burden, 2010; Hug, 2011) could introduce variations between levels, and it should be recognized that the single normalization task used may not have achieved equivalent maximum EMG at all levels measured.

There were notable time lags for EMG-measured compared to model-estimated activity for the gluteus muscles and biceps femoris. When adjusting for this time-lag, magnitude errors of gluteus muscles were comparable to the ES, as were biceps femoris errors during stoop lifting (Fig. 4). The EMG indicates minimal eccentric contraction of the gluteus muscles and biceps femoris during the descent of lifting or lowering, as compared to the more rapid activation of these muscles during ascent (Fig. 2). The model, however, estimated that these muscles were eccentrically active earlier during the descent. The EMG is aligned with prior work reporting that gluteus activation occurs later than erector spinae (Lehman et al., 2004; Leinonen et al., 2000; Van Gelder et al., 2015), and we also observed similar delays in the biceps femoris. Earlier activations of gluteus and biceps femoris in the model may reflect that the static optimization assessment does not account for passive elastic muscle forces, which could significantly contribute to hip extensor forces *in vivo*. This limitation warrants further investigation.

One purpose of a back exosuit is to reduce muscle demands during lifting and lowering tasks (Kermavnar et al., 2021). We previously reported that this exosuit reduces measured muscle activity at the L3 level by 15–19 % in squat and 9–11 % in stoop (Quirk et al., 2023, 2024). The present work expands upon this, estimating a 6–10 % reduction of muscle force when lifting and 4–7 % when lowering (Fig. 6). Comparing changes in muscle force with exosuit forces, we established ES force reductions of ~1.7–1.9-fold larger than the assistance delivered by the exosuit. These results support the theory that parallel assistance back exosuits can reduce compressive spine forces due to a mechanical advantage relative to spinal muscles (Abdoli-Eramaki et al., 2007; Banks et al., 2024; Lamers et al., 2018). Moreover, this exosuit could reduce back muscle fatigue, and thereby injury risk in the back (McGill, 1997; Salvetti et al., 2013). Consistent with a previous study showing peak exosuit force delivery occured after maximal trunk flexion (Chung et al., 2024), maximal exosuit assistance did not coincide with peak muscle force. It is likely that improvements in exosuit control could produce even greater effects.

Hip extensor muscle forces demonstrated a net reduction in muscle force, but only in the 10 kg squat lift. In this scenario the exosuit appeared to reduce hip extensor forces considerably more than the level of applied exosuit force. However, the hip extensor forces were larger and more variable than ES forces and often occurred (over 750 ms) before peak exosuit forces were delivered, making significant exosuit effects indetectable in three of the four lifting scenarios. In general, a lack of significant reductions in model estimated muscle forces align with previous studies that demonstrate this exosuit can consistently reduce hip extensor moments but not muscle activity (Quirk et al., 2023, 2024). Therefore, while this exosuit may reduce back muscle fatigue, it could have a minimal impact on the hip extensor muscles. However, this statement should be confirmed with future modeling work that includes more detailed representations of hip musculature, such as with passive musculotendinous forces and other hip extensors (i.e., semimembranosus, semitendinosus). Such efforts could better optimize the exosuit design and control to produce more meaningful reductions in hip extensor muscle forces.

4.1. *Strengths and limitations*

This study quantified the performance of an advanced musculoskeletal model incorporating a soft active exosuit and provided novel insights into exosuit effects on muscle loading. The model predicted the effect of the exosuit on the back and hip extensor muscles, and the impact of the exosuit during different lifting tasks via muscle force reductions. However, there are several limitations. First, the relatively small sample was limited to young fit adults and did not include all potential anthropometries. Second, men were over-represented in our sample, which may limit the understanding of the results as they apply to women in general. However, previous work from our group has reported EMG reductions can be comparable in a female dominated study (Quirk et al., 2023). Further, it is unlikely that gender differences (e.g. kinematics), even if present, would affect our overall findings due to our within-subject design. Third, the exosuit limited the available space for the EMG sensors which may have impacted the quality of the recordings by influencing placement or contacting the sensors during testing. Moreover, muscle crosstalk, movement artifact, and electromechanical delay can compromise EMG accuracy, and may not represent the intended activations (Hofste et al., 2020). Fourth, the musculoskeletal modeling approach has known limitations, which includes not accounting for passive tissue forces and a tendency to under-predict antagonistic co-activations (Alemi et al., 2023; Banks et al., 2024). Participant muscle strength was not measured to adjust model muscle strength, which could also affect model-determined activations. Finally, participants performed all the trials at a set pace with prescribed techniques, and only sagittal lifting tasks were examined here. The exosuit may perform differently at self-selected paces and in less restricted movements. However, these modeling and methodological limitations should not alter the overall findings which still support the use of musculoskeletal modeling studies to provide further insight into the biomechanical effects of exosuit devices.

5. Conclusions

This study demonstrates that a musculoskeletal model can appropriately incorporate a soft active back support exosuit to predict muscle activations patterns and magnitudes. The model-determined muscle activation in the back and hip extensors aligned with measured data similarly with and without the exosuit, suggesting that modeling the exosuit does not alter model validity. Some temporal offsets were noted between model and measured activations for hip extensor muscles. The model-based estimations of erector spinae and hip extensor muscle forces indicate that the exosuit reduces peak muscle forces to a larger extent than the applied exosuit force. This was consistent for the ES, but for the hip extensors appeared to be lift dependent. Though the underlying source and significance of these differences are not fully understood, peak exosuit force was not temporally aligned with peak muscle force, suggesting a more optimal force application might increase the exosuit effectiveness. Overall, the results support the hypothesis that exosuits can reduce muscle loading, and thereby reduce fatigue and injury risk during manual materials handling tasks.

CRediT authorship contribution statement

Chenxi Yan: Writing – original draft, Software, Formal analysis. **Jacob J. Banks:** Writing – original draft, Software, Formal analysis. **Brett T. Allaire:** Writing – original draft, Software, Data curation. **D. Adam Quirk:** Writing – review & editing, Methodology, Investigation, Data curation, Conceptualization. **Jinwon Chung:** Methodology, Investigation, Data curation, Conceptualization. **Conor J. Walsh:** Writing – review & editing, Supervision, Project administration, Methodology, Funding acquisition, Conceptualization. **Dennis E. Anderson:** Writing – original draft, Supervision, Software, Project administration, Funding acquisition, Formal analysis, Data curation, Conceptualization.

Declaration of competing interest

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

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References

Abdoli-Eramaki, M., Stevenson, J.M., Reid, S.A., Bryant, T.J., 2007. Mathematical and empirical proof of principle for an on-body personal lift augmentation device (PLAD). J. Biomech. 40, 1694–1700. https://doi.org/10.1016/j.jbiomech.2006.09.006.

Akhavanfar, M., Uchida, T.K., Clouthier, A.L., Graham, R.B., 2022. Sharing the load: modeling loads in OpenSim to simulate two-handed lifting. Multibody Syst. Dyn. 54, 213–234. https://doi.org/10.1007/s11044-021-09808-7.

Akhavanfar, M., Mir-Orefice, A., Uchida, T.K., Graham, R.B., 2024. An enhanced spine model validated for simulating dynamic lifting tasks in OpenSim. Ann. Biomed. Eng. 52, 259–269. https://doi.org/10.1007/s10439-023-03368-x.

Alemi, M.M., Burkhart, K.A., Lynch, A.C., Allaire, B.T., Mousavi, S.J., Zhang, C., Bouxsein, M.L., Anderson, D.E., 2021. The influence of kinematic constraints on model performance during inverse kinematics analysis of the thoracolumbar spine. Front. Bioeng. Biotechnol. 9, 688041. https://doi.org/10.3389/fbioe.2021.688041.

Alemi, M.M., Banks, J.J., Lynch, A.C., Allaire, B.T., Bouxsein, M.L., Anderson, D.E., 2023. EMG validation of a subject-specific thoracolumbar spine musculoskeletal model during dynamic activities in older adults. Ann. Biomed. Eng. 51, 2313–2322. https://doi.org/10.1007/s10439-023-03273-3.

Ali, A., Fontanari, V., Schmoelz, W., Agrawal, S.K., 2021. Systematic review of back-support exoskeletons and soft robotic suits. Front. Bioeng. Biotechnol. 9, 765257. https://doi.org/10.3389/fbioe.2021.765257.

Allaire, B.T., Mousavi, S.J., James, J.N., Bouxsein, M.L., Anderson, D.E., 2023. Dependence of trunk muscle size and position on age, height, and weight in a multi-ethnic cohort of middle-aged and older men and women. J. Biomech. 157, 111710. https://doi.org/10.1016/J.JBIOMECH.2023.111710.

Banks, J.J., Quirk, D.A., Chung, J., Cherin, J.M., Walsh, C.J., Anderson, D.E., 2024. The effect of a soft active back support exosuit on trunk motion and thoracolumbar spine loading during squat and stoop lifts. Ergonomics 1–14. https://doi.org/10.1080/00140139.2024.2320355.

Bär, M., Steinhilber, B., Rieger, M.A., Luger, T., 2021. The influence of using exoskeletons during occupational tasks on acute physical stress and strain compared to no exoskeleton – a systematic review and meta-analysis. Appl. Ergon. 94, 103385. https://doi.org/10.1016/j.apergo.2021.103385.

Bruno, A.G., Bouxsein, M.L., Anderson, D.E., 2015. Development and validation of a musculoskeletal model of the fully articulated thoracolumbar spine and rib cage. J. Biomech. Eng. 137, 081003. https://doi.org/10.1115/1.4030408.

Bruno, A.G., Burkhart, K., Allaire, B., Anderson, D.E., Bouxsein, M.L., 2017. Spinal loading patterns from biomechanical modeling explain the high incidence of vertebral fractures in the thoracolumbar region. J. Bone Mineral Res. 32, 1282–1290. https://doi.org/10.1002/jbmr.3113.

Burden, A., 2010. How should we normalize electromyograms obtained from healthy participants? What we have learned from over 25 years of research. J. Electromyogr. Kinesiol. 20, 1023–1035. https://doi.org/10.1016/j.jelekin.2010.07.004.

Burkhart, K., Grindle, D., Bouxsein, M.L., Anderson, D.E., 2020. Between-session reliability of subject-specific musculoskeletal models of the spine derived from optoelectronic motion capture data. J. Biomech. 112, 110044. https://doi.org/10.1016/j.jbiomech.2020.110044.

Chung, J., Quirk, D.A., Applegate, M., Rouleau, M., Degenhardt, N., Galiana, I., Dalton, D., Awad, L.N., Walsh, C.J., 2024. Lightweight active back exosuit reduces muscular effort during an hour-long order picking task. Commun. Eng. 3, 35. https://doi.org/10.1038/s44172-024-00180-w.

Crowninshield, R.D., Brand, R.A., 1981. A physiologically based criterion of muscle force prediction in locomotion. J. Biomech. 14, 793–801. https://doi.org/10.1016/0021-9290(81)90035-X.

Delp, S.L., Anderson, F.C., Arnold, A.S., Loan, P., Habib, A., John, C.T., Guendelman, E., Thelen, D.G., 2007. OpenSim: open-source software to create and analyze dynamic simulations of movement. I.E.E.E. Trans. Biomed. Eng. 54, 1940–1950. https://doi.org/10.1109/TBME.2007.901024.

Ferguson, S.A., Marras, W.S., 2013. Spine kinematics predict symptom and lost time recurrence: How much recovery is enough? J. Occup. Rehabil. 23, 329–335. https://doi.org/10.1007/s10926-012-9413-x.

Hermens, H.J., Freriks, B., Merletti, R., Stegeman, D., Blok, J., Rau, G., Disselhorst-Klug, C., Hägg, G., 1999. European recommendations for surface electromyography - results of the SENIAM project. Roessingh Res. Develop.

Hicks, J.L., Uchida, T.K., Seth, A., Rajagopal, A., Delp, S.L., 2015. Is my model good enough? best practices for verification and validation of musculoskeletal models and simulations of movement. J. Biomech. Eng. 137, 020905. https://doi.org/10.1115/1.4029304.

Hofste, A., Soer, R., Salomons, E., Peuscher, J., Wolff, A., Van Der Hoeven, H., Oosterveld, F., Groen, G., Hermens, H., 2020. Intramuscular EMG Versus Surface EMG of Lumbar Multifidus and Erector Spinae in Healthy Participants. Spine (Phila Pa 1976) 45, E1319–E1325. Doi: 10.1097/BRS.0000000000003624.

Hug, F., 2011. Can muscle coordination be precisely studied by surface electromyography? J. Electromyogr. Kinesiol. 21, 1–12. https://doi.org/10.1016/j.jelekin.2010.08.009.

Huysamen, K., Bosch, T., de Looze, M., Stadler, K.S., Graf, E., O'Sullivan, L.W., 2018. Evaluation of a passive exoskeleton for static upper limb activities. Appl. Ergon. 70, 148–155. https://doi.org/10.1016/j.apergo.2018.02.009.

Kermavnar, T., de Vries, A.W., de Looze, M.P., O'Sullivan, L.W., 2021. Effects of industrial back-support exoskeletons on body loading and user experience: an updated systematic review. Ergonomics 64, 685–711. https://doi.org/10.1080/00140139.2020.1870162.

Lamers, E.P., Yang, A.J., Zelik, K.E., 2018. Feasibility of a biomechanically-assistive garment to reduce low back loading during leaning and lifting. I.E.E.E. Trans. Biomed. Eng. 65, 1674–1680. https://doi.org/10.1109/TBME.2017.2761455.

Lee, L.F., Umberger, B.R., 2016. Generating optimal control simulations of musculoskeletal movement using OpenSim and MATLAB. PeerJ 2016, e1638.

Lehman, G.J., Lennon, D., Tresidder, B., Rayfield, B., Poschar, M., 2004. Muscle recruitment patterns during the prone leg extension. BMC Musculoskeletal Disorder 5, 3. https://doi.org/10.1186/1471-2474-5-3.

Leinonen, V., Kankaanpää, M., Airaksinen, O., Hänninen, O., 2000. Back and hip extensor activities during trunk flexion/extension: effects of low back pain and rehabilitation. Arch. Phys. Med. Rehabil. 81, 32–37. https://doi.org/10.1053/apmr.2000.0810032.

McGill, S.M., 1997. The biomechanics of low back injury: implications on current practice in industry and the clinic. J. Biomech. 30, 465–475. https://doi.org/10.1016/s0021-9290(96)00172-8.

Norman, R., Wells, R., Neumann, P., Frank, J., Shannon, H., Kerr, M., 1998. A comparison of peak vs cumulative physical work exposure risk factors for the reporting of low back pain in the automotive industry. Clin. Biomech. 13, 561–573. https://doi.org/10.1016/s0268-0033(98)00020-5.

Quirk, D.A., Chung, J., Schiller, G., Cherin, J.M., Arens, P., Sherman, D.A., Zeligson, E.R., Dalton, D.M., Awad, L.N., Walsh, C.J., 2023. Reducing back exertion and improving confidence of individuals with low back pain with a back exosuit: a feasibility study for use in BACPAC. Pain Med. 24, S175–S186. https://doi.org/10.1093/pm/pnad003.

Quirk, D.A., Chung, J., Applegate, M., Cherin, J.M., Dalton, D.M., Awad, L.N., Walsh, C. J., 2024. Evaluating adaptiveness of an active back exosuit for dynamic lifting and maximum range of motion. Ergonomics 67, 660–673. https://doi.org/10.1080/00140139.2023.2240044.

Salvetti, M. de G., Pimenta, C.A. de M., Braga, P.E., McGillion, M., 2013. Prevalence of fatigue and associated factors in chronic low back pain patients. Rev. Lat. Am. Enfermagem. 21, 12–19. https://doi.org/10.1590/s0104-11692013000700003.

Van Gelder, L.H., Hoogenboom, B.J., Alonzo, B., Briggs, D., Hatzel, B., 2015. EMG analysis and sagittal plane kinematics of the two-handed and single-handed kettlebell swing. Int. J. Sports Phys. Therapy 10, 811–826.

Winter, D.A., 2009. Biomechanics and motor control of human movement, fourth ed. John Wiley & Sons Inc.

Zander, T., Dreischarf, M., Schmidt, H., 2016. Sensitivity analysis of the position of the intervertebral centres of reaction in upright standing - a musculoskeletal model investigation of the lumbar spine. Med. Eng. Phys. 38, 297–301. https://doi.org/10.1016/j.medengphy.2015.12.003.

Zelik, K.E., Nurse, C.A., Schall, M.C., Sesek, R.F., Marino, M.C., Gallagher, S., 2022. An ergonomic assessment tool for evaluating the effect of back exoskeletons on injury risk. Appl. Ergon. 99, 103619. https://doi.org/10.1016/j.apergo.2021.103619.