# Biomechanical Analysis and Modeling of Back-Support Exoskeletons for Use in Repetitive Lifting Tasks

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ABSTRACT

Low back pain (LBP) remains the most prevalent and costly work-related disability worldwide and is directly associated with "physical" risk factors prevalent in manual material handling (MMH) tasks. Back-support exoskeletons (BSEs) are a promising ergonomic intervention to mitigate LBP risk, by reducing muscular exertion and spine loading. The purpose of this work was to help better understand both the "intended" and "unintended" consequences of BSE use on physical risk factors for LBP, as an essential prerequisite for the safe and effective implementation of this technology in actual workplaces.

The first study assessed the effects of using two BSEs on objective and subjective responses during repetitive lifting involving symmetric and asymmetric postures. Wearing both BSEs significantly reduced peak levels of trunk extensor muscle activity and reduced energy expenditure. Such reductions, though, were more pronounced in the symmetric conditions and differed between the two BSEs tested.

The second study quantified the assistive torque profiles of two passive BSEs using a computerized dynamometer, with both human subjects and a mannequin. Clear differences in torque magnitudes were evident between the BSEs, though both generated more assistive torques during flexion than extension.

The third study estimated the effects of BSE use on lumbosacral compressive and shear forces during repetitive lifting using an optimization-based model. Using both BSEs reduced peak compression and anteroposterior shear forces, but these effects differed between tasks and BSE designs. Reductions in composite measures of trunk muscle activity did not correspond consistently with changes in spine forces when using a BSE.

The fourth study quantified the effects of two passive BSEs on trunk stability and movement coordination during repetitive lifting. Some adverse effects on stability were evident for pelvis and thorax movements and coupling of these body segments, suggesting that caution is needed in selecting a BSE for a given MMH task.

Overall, we found that the efficacy of BSEs is design- and task-specific. Important safety features of the exoskeletons were also identified, providing insights on their performance boundaries. Overall, the BSEs tested were more effective and safer in tasks closer to the mid-sagittal plane and with moderate degrees of trunk flexion.

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GENERAL AUDIENCE ABSTRACT

Low back pain (LBP) remains the most prevalent and costly work-related disability worldwide, and the risk of LBP is related to "physical" risk factors common in manual material handling (MMH) tasks. Back-support exoskeletons (BSEs) are a new ergonomic intervention that may reduce the risk of occupational LBP, by reducing muscular efforts and loads on the spine. For the safe use of BSEs, though, it is critical to better understand both the "intended" and "unintended" consequences of this emerging technology. In this dissertation, such consequences of BSE use were evaluated in the context of repetitive lifting tasks.

The first study assessed the efficacy of two BSEs in terms of physical demands during repetitive lifting tasks involving a range of torso bending and twisting. Wearing both BSEs reduced the physical demands on back muscles and decreased energy consumption. Larger reductions, though, were observed in forward bending and such reductions differed between the two BSEs tested.

The second study measured the amount of support provided by two BSEs using a new measurement method, which was examined for both human subjects and a mannequin. Clear differences in the BSE support were evident between the BSEs, and both devices generated more support during torso forward bending than returning upright.

The third study estimated the effects of BSE use on low back loadings during repetitive lifting using a computational model. Using both BSEs reduced loads on the low back region, though such reductions were task-specific and depended on the BSE design.

The fourth study quantified the effects of the BSE use on torso stability and movement patterns during repetitive lifting. Some adverse effects on stability were evident for lower and upper torso, suggesting that caution is needed in selecting a BSE for a given MMH task.

Findings from this work show the potential benefits of BSEs for use in MMH tasks, yet such benefits can depend on the BSE design and the MMH task they are used for. Further, BSE use can lead to adverse effects, especially with tasks involving extreme working postures.

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# 1 Chapter 1

## 1.1 Introduction

The U.S. workforce continues to experience high rates of work-related musculoskeletal disorders (WMSDs) due to overexertion and bodily reaction (BLS, 2019). Among all WMSDs, the back was most affected, accounting for ~40% of such cases and resulting in a median of seven lost workdays (BLS, 2019). This high burden of low-back WMSDs is attributed to physical risk factors prevalent in manual material handling (MMH) tasks, including forceful exertions, repetitive lifting and bending, and sustained/prolonged non-neutral postures (Hoogendoorn et al., 2000; da Costa & Vieira, 2010). While the mechanization or automation of work processes have reduced exposures to these risk factors in some cases, these risk factors are less-easily addressed for jobs that require the flexibility and adaptability of human workers Hunter, 2001; Alzuheri et al., 2010. For the latter, diverse intervention approaches have been explored to reduce the physical requirements involved. Examples of these approaches include training in work methods (Daltroy et al., 1997; Burke et al., 2006), modifying workstations (Silverstein & Clark, 2004), re-designing work processes (Haight & Belwal, 2006; Madinei et al., 2018), and using mechanical aids such as cranes and power-lift tables (Nussbaum et al., 1999; Lavender et al., 2013). While these approaches have potential advantages, concerns have been raised over their efficacy, cost-effectiveness, and ease of adoption (Hignett, 2003; Clemes et al., 2009).

Exoskeletons are wearable devices that augment, enable, assist, and/or enhance motion, posture, and physical activity (Lowe et al., 2019). In recent years, the occupational application of such technologies has received increasing attention as an additional and promising workplace intervention to improve the balance between task demands and worker capacity, while retaining the mobility that manual work requires. Specifically, back-support exoskeletons (BSEs) assist the low-back region by contributing to the spine extensor moments needed to counteract flexor moments due to gravity and inertia on the upper body and from handled loads. BSEs provide assistive torques either passively, by means of springs, dampers, and/or elastic materials, or actively, through powered actuators and associated power supplies (Lee et al., 2012). At present, passive BSEs are predominant in the commercial market and are being tested and adopted in several occupational settings due to their cost-efficiency and ease of implementation (De Looze et al., 2016; Hensel & Keil, 2019).

Earlier work has shown that using a passive BSE can reduce physical demands. Use of a BSE reduced trunk extensor muscle activity during static trunk bending, by up to 57% during symmetric tasks (Barrett & Fathallah, 2001; Graham et al., 2009; Ulrey & Fathallah, 2013; Bosch et al., 2016; Lamers et al., 2018; Koopman et al., 2019), and by up to 37% during asymmetric tasks (Madinei et al., 2020b). In the context of repetitive lifting, reductions of up to 54% in trunk muscle activity (or trunk external moment) were found during repetitive, symmetric lifting (Abdoli-e et al., 2006; Abdoli-e & Stevenson, 2008; Frost et al., 2009; Godwin et al., 2009; Lotz et al., 2009; Wehner et al., 2009; Lamers et al., 2018; Näf et al., 2018; Alemi et al., 2019), and by up to 30% during asymmetric lifting (Abdoli-e & Stevenson, 2008; Alemi et al., 2019; Alemi et al., 2020). Earlier work has also shown that the use of a passive BSE can provide metabolic savings of up to 17% during symmetric lifting (Alemi, 2019; Baltrusch et al., 2019) and up to 6% during asymmetric lifting tasks (Alemi et al., 2020). This wide range, though, suggests that BSE benefits are dependent on the design approach and the specific tasks for which they are used.

Moreover, using a BSE can impose unintended or adverse consequences on the user. (Here, we use the term “unintended” to indicate effects that are not likely to have been either drivers of exoskeleton design or desired outcomes of using an exoskeleton.) Though available evidence is limited, in some cases these consequences may be important. For example, wearing a BSE can increase discomfort at the chest (Bosch et al., 2016; Hensel & Keil, 2019) and thighs (Amandels et al., 2018), restrict trunk range-of-motion (Abdoli-e et al., 2007; Sadler et al., 2011), or decrease trunk angular velocity (Koopman et al., 2020b). BSE use can also cause the wearer to adopt a more stooped lifting style, likely to increase the device's supportive forces that occur with larger lumbar flexion angle (Frost et al., 2009). Further, one study found that wearing a personal lift-assistive device can decrease spine stability during static trunk flexion (Agnew & Stevenson, 2008), whereas another indicated that the same device increased local dynamic stability during a repetitive lifting task (Graham et al., 2011). Given the divergence of findings and methodologies employed in earlier work, there is a clear need for comparative investigations of potential limitations and other adverse effects of different BSEs when used for MMH tasks.

In summary, the use of BSEs offers a new solution to control exposures to LBP physical risk factors during MMH tasks. Yet, given the rapid market introduction of BSEs, and the divergence of methodologies employed in earlier studies, evidence of the effectiveness of different BSEs for diverse MMH tasks is lacking. Furthermore, the potential adverse consequences of BSE use are not sufficiently understood to support or ensure their safe widespread adoption. Also lacking at present are comprehensive evaluations of alternative BSEs in terms of their impacts on spine loading and stability; both of the latter are important LBP risk factors that can increase the stress/strain within trunk tissues to levels that exceed the thresholds of trunk nociceptors and lead to tissue damage (McGill et al., 2003). As such, the ability of BSEs to mitigate spine loadings while maintaining/enhancing spine stability is essential for their effective and safe implementation. Figure 1.1 provides a high-level overview of the diverse potential impacts of BSEs (and other occupational exoskeletons, in terms of benefits, unintended effects, and challenges. As described subsequently, the main goals in this dissertation were the former two aspects.

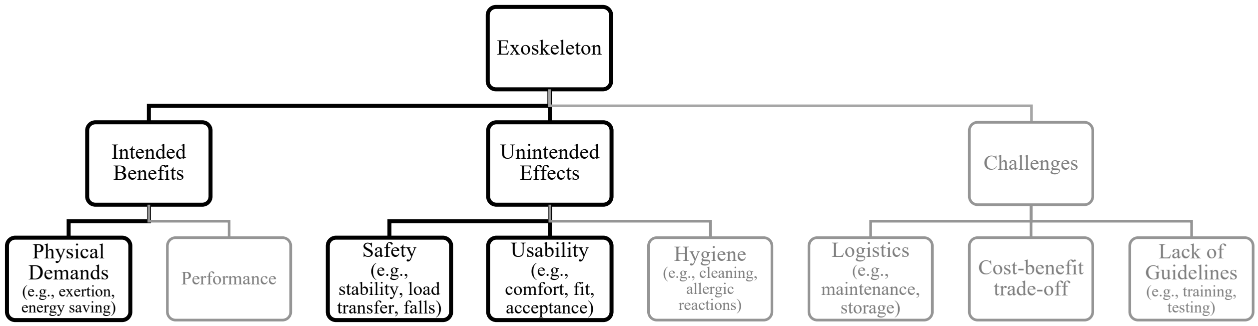


Figure 1.1. A schematic of intended benefits, unintended effects, and challenges for adopting exoskeletons. Solid black lines and boxes indicate areas of focus in the current dissertation.

## 1.2 Research Needs

As emphasized above, promoting the adoption and use of BSEs in MMH tasks would be premature based solely on existing evidence, since critical gaps remain regarding the effectiveness, efficacy, and safety of these devices. This dissertation is comprised of four chapters, each addressing different aspects of intended and unintended consequences of BSE use.

1) **Assessing the effects of BSE use on objective and subjective responses during repetitive lifting**

   The majority of reported research on passive BSEs has been specific to a particular brand or application, but with little ability to generalize findings beyond the particular BSE design or condition(s) tested. Thus, there are clear needs for more systematic, comparative evaluations of different BSE designs under diverse working conditions. Such comparisons are especially needed to help guide the selection and application of these devices, especially for tasks involving asymmetric postures since such postures are associated with an increased risk of LBP (Punnett et al., 1991; Norman et al., 1998; Punnett et al., 2005). Recently, we examined the efficacy of two passive BSEs during repetitive lifting tasks (Alemi et al., 2020), finding that both BSEs resulted in significant reductions in energy expenditure by (4-13%) and peak activity of the trunk extensor muscles (by 10-28%). Such benefits, though, were substantially task-dependent and differed between the devices tested. Furthermore, the lifting conditions simulated in that study considered the functionality of the BSEs near their extreme operating regions. As such, further exploration is needed to assess the efficacy of these devices in more moderate lifting postures. Also lacking in the literature is a comparison of BSE effects on working postures when performing diverse MMH tasks. Addressing these critical gaps will improve our understanding of the beneficial and adverse effects of different BSE designs and help guide the selection and application of BSEs by maximizing their benefits and minimizing/avoiding their unintended/preventable side effects.

2) **Quantify the assistive torque profiles of passive BSEs:**

Understanding the mechanical behavior of the torque generation mechanisms embedded in passive BSEs is essential to determine the efficacy of these devices in both static and dynamic tasks. These mechanical behaviors are also critical in biomechanical modeling to predict the effects of BSE use. Given the viscoelastic behavior of the spring gas cylinders incorporated in the passive BSEs examined in this work, assistive torques may have a complex dependency on flexion/extension angle and velocity. Exoskeleton torque profile data, though, are typically proprietary and not publicly available. An earlier study quantified the torque profile generated by Laevo™ using a force transducer (Koopman et al., 2020b) and found that the torque output was not dependent upon flexion/extension speed. They also reported larger torques during flexion than extension. The approach used by (Koopman et al., 2020b) is seemingly straightforward, yet it limits control over trunk flexion/extension speeds since the participants arbitrarily flex/extend their torso. Further, it is unclear whether the assistive torques for the Laevo™ were the pure torques generated by the device. We are also unaware of any reported torque profile information for BackX™. As such, we introduced a novel approach to quantify the assistive torque profiles of both Laevo™ and BackX™ using a computerized dynamometer.

3) **Quantify changes in spine loads resulting from BSE use**:

While existing evidence supports the potential for passive BSEs to reduce low-back demands and associated injury risks, many earlier studies examined the beneficial effects of BSEs only in terms of reductions in superficial back muscle activity. Loads on the human spine, however, are influenced not only by the activity of surface trunk muscles, but also by deep trunk muscles and passive tissues (Bazrgari & Shirazi-Adl, 2007), as well as segmental kinematics and external loads (i.e., due to gravity and inertia). Consequently, and especially if lifting behaviors change when using a BSE, reductions in back muscle activity, measured via electromyography (EMG), may not necessarily imply a reduction in spine loading. As such, we expected that using a BSE would reduce spine loads, yet that the magnitude of this reduction would not always correspond directly with measured reductions in back muscle activity, given that the latter has only a partial role contributing to resultant spine loads in some postures. Practically, this would imply that invalid conclusions would be reached if investigations rely on surface EMG measures to assess the impacts of BSEs on spine loads.

4) **Determine whether BSE use affects trunk dynamic stability and movement coordination**:

   A major concern when using a BSE is the ability to maintain spine stability, specifically to satisfy local and global equilibrium in the presence of neuromuscular control errors and/or mechanical perturbations. (Graham et al., 2011). Contemporary evidence suggests that compromised spine stability could lead to sub-optimal transmission of compressive and shear forces along the spinal column and thereby contribute to the risk of low back injuries. (Cholewicki & McGill, 1996; Panjabi, 2003). Control of the spine is achieved by the central nervous system and involves forces generated by active and passive tissues in the torso (Hoffman & Gabel, 2013). Wearing a BSE, by supplying additional support, can reduce the activation levels of the trunk extensor musculature, ultimately decreasing the compressive loads incurred in the lumbar spine (Koopman et al., 2020b). A decrease in muscle activation, however, could also compromise the contribution of active muscle stiffness to spine stability, since there is a direct relationship between muscle stiffness and activation level (Stokes & Gardner-Morse, 2003; Agnew & Stevenson, 2008). It is presently unknown, though, whether such changes in trunk neuromuscular control parameters affect the ability to maintain stable movement behavior (i.e., dynamic stability) when performing various occupational tasks. For the safe adoption of BSEs in an actual workplace, it is thus important to determine the extent to which different BSEs alter trunk dynamic stability when performing MMH tasks.

This work is innovative in being the first to systematically evaluate the effects of different BSEs on physical demands, usability, and safety across different task types. This comprehensive evaluation is especially needed to understand appropriate applications and safety considerations of these technologies, and to help effectively adopt and integrate them into the workplace. Second, the current dissertation is also the first to determine the dual effects of different BSE designs on spine loading and stability. Evidence regarding these important effects can facilitate the proper selection of BSEs, by identifying appropriate work tasks that could benefit from a specific BSE and by informing users of task characteristics or BSE designs that present concerns about spine stability. This work is also the first to employ an optimization-based model of the spine, integrating BSE mechanics and the mechanical interaction of a BSE with the body. In particular, a mechanical interface for each BSE design was generated in a commercial computational platform, and a novel approach was used to quantify the mechanical behavior of each BSE (i.e., torque vs. angle relationship). This innovative method will be useful for future enhancements of the design characteristics of passive BSEs. As such, this research is unique in implementing an effective suite of assessment tools, and the results will contribute substantially to the future development of ergonomic and safety guidelines, ultimately supporting more effective and safer adoption and use of occupational exoskeletons.

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# 2 Chapter 2. Biomechanical Assessment of Two Back-Support Exoskeletons in Symmetric and Asymmetric Repetitive Lifting with Moderate Postural Demands[1]

## 2.1 Abstract

Two passive back-support exoskeleton (BSE) designs were assessed in terms of muscular activity, energy expenditure, joint kinematics, and subjective responses. Eighteen participants (gender-balanced) completed repetitive lifting tasks in nine different conditions, involving symmetric and asymmetric postures and using two BSEs (along with no BSE as a control condition). Wearing both BSEs significantly reduced peak levels of trunk extensor muscle activity (by ~9-20%) and reduced energy expenditure (by ~8-14%). Such reductions, though, were more pronounced in the symmetric conditions and differed between the two BSEs tested. Participants reported lower perceived exertion using either BSE yet raised concerns regarding localized discomfort. Minimal changes in lifting behaviors were evident when using either BSE, and use of both BSEs led to generally positive usability ratings. While these results are promising regarding the occupational use of BSEs, future work is recommended to consider inter-individual differences to accommodate diverse user needs and preferences.

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## 2.2 Introduction

Low back disorders (LBDs) continue as the leading cause of work-related disability, accounting for ~40% of all work-related musculoskeletal disorders (WMSDs) and ~38% of cases in the U.S. involving days away from work (BLS, 2019). The development of LBDs has been directly associated with “physical” risk factors including overexertion, repetitive lifting, bending, and prolonged/sustained non-neutral trunk postures (Punnett et al., 2005; da Costa & Vieira, 2010; Griffith et al., 2012). Engineering controls that re-design tools or workspace (Silverstein & Clark, 2004; Lavender et al., 2013), use of mechanical aids (Westgaard & Winkel, 1997; Nussbaum et al., 1999), and/or worker training (Daltroy et al., 1997; Burke et al., 2006) have each been proposed as potential ergonomic interventions to minimize the risks of work-related LBDs. However, evidence of their efficacy, sustainability, and/or usability remains limited in practice (Hignett, 2003; Clemes et al., 2009), and in some working scenarios one or more of these intervention approaches can be infeasible, impractical, or excessively costly.

Back-support exoskeletons/exosuits (BSEs) are designed to support, augment, and/or assist with the back and hip muscles, by producing restorative torques, passively by the means of springs and/or elastic materials, or actively through use of powered actuators (Lee et al., 2012). BSEs have emerged recently as an alternative and promising intervention to reduce physical demands on the spine, while retaining the mobility that manual work requires. While the development of active BSEs is still in progress (Toxiri et al., 2017; Huysamen et al., 2018; Toxiri et al., 2019), multiple passive devices are already commercially available and are being tested and adopted in the workplace due to their cost-efficiency and ease of implementation (De Looze et al., 2016; Hensel & Keil, 2019).

Previous studies have evaluated the efficacy of passive BSEs, specifically during static trunk bending and repetitive lifting. In many of these studies, the potential benefits of passive BSEs were quantified in terms of a reduction in back muscle demands. Passive BSEs were found to reduce trunk extensor muscle activity during static trunk bending, by up to 57% during symmetric tasks (Barrett & Fathallah, 2001; Graham et al., 2009; Ulrey & Fathallah, 2013; Bosch et al., 2016; Lamers et al., 2018; Koopman et al., 2019), and by up to 37% during asymmetric tasks (Madinei et al., 2020b). Reductions of up to 54% in trunk muscle activity (or trunk external moment) were found during repetitive, symmetric lifting (Abdoli-e et al., 2006; Abdoli-e & Stevenson, 2008; Frost et al., 2009; Godwin et al., 2009; Lotz et al., 2009; Wehner et al., 2009; Lamers et al., 2018; Näf et al., 2018; Alemi et al., 2019), and by up to 30% during asymmetric lifting (Abdoli-e & Stevenson, 2008; Alemi et al., 2019; Alemi et al., 2020). Earlier work has also shown that the use of a passive BSE can reduce metabolic demands by up to 17% during symmetric lifting (Alemi, 2019; Baltrusch et al., 2019) and by up to 6% during asymmetric lifting (Alemi et al., 2020). Despite such potential benefits, use of passive BSEs may also impose unintended or adverse consequences on users, such as increased discomfort at the chest (Bosch et al., 2016; Hensel & Keil, 2019) and thighs (Amandels et al., 2018), or adoption of riskier working postures to exploit the device support, such as increased lumbar flexion and knee extension (Frost et al., 2009; Sadler et al., 2011; Bosch et al., 2016; Koopman et al., 2020b).

While there is a growing body of literature on the efficacy of passive BSEs in repetitive lifting tasks, most earlier work is limited in three important aspects. First, previous studies often considered only one particular BSE; thus, it is unclear whether there is a preferred BSE design under various working conditions. Additional comparative evaluations of different BSE designs are needed to help guide the selection and application of these devices under diverse conditions, given that the functionality of a BSE appears to be task-specific and dependent on the design approach (Alemi et al., 2020). Such comparisons are especially needed for tasks involving asymmetric postures since these are associated with an increased risk of LBDs (e.g., Punnett et al., 1991; Norman et al., 1998; Punnett et al., 2005). Recently, we examined the efficacy of two passive BSEs during repetitive lifting tasks (Alemi et al., 2020), finding that both BSEs resulted in significant reductions in energy expenditure (by 4-13%) and peak activity of trunk extensor muscles (by 10-28%). Such benefits, though, were task-dependent and differed between the devices tested. Furthermore, the lifting conditions simulated in that study considered the functionality of the BSEs near their extreme operating regions, and thus those findings may not generalize to more moderate working postures. As such, the efficacy of these devices in a range of moderate lifting postures needs further exploration. Second, prior investigations were often focused on trunk muscle activities, however conclusions regarding the actual benefits of using a BSE would be premature without knowledge of trunk kinematics (Koopman et al., 2020b). Third, evidence on the efficacy of passive BSEs is rather limited in asymmetric working postures as compared to symmetric postures.

To address these limitations, this exploratory study evaluated the efficacy of two passive BSE designs during repetitive symmetric and asymmetric lifting and lowering involving a range of lifting heights to simulate diverse tasks. The two BSEs tested were both commercially-available: BackX™ model AC (SuitX™, www.suitx.com), and Laevo™ V2.5 (www.laevo.nl). Both of these devices incorporate passive torque generation mechanisms about the hip that are intended to augment the trunk extensor muscles, yet they have distinct design features. The BackX™ AC includes a structural frame that pulls the torso backwards by distributing the pressure to the shoulder straps and chest pad, whereas the Laevo™ transfers the load through pushing against the chest only. Further, support levels and modes can be adjusted easily with the BackX™, while the Laevo™ only provides different engagement angles. We included muscle activity, energy expenditure, and joint kinematics as objective outcome measures, and supplemented these with several subjective assessments including perceived exertion, discomfort, and usability. Based on earlier findings as summarized earlier, we hypothesized that using these BSEs during repetitive lifting activities would reduce muscle activity in the lower back muscles and provide metabolic savings, but that the magnitude of these benefits would vary between the two BSEs and across task conditions.

## 2.3 Methods

### 2.3.1 Participants

A convenience sample of 18 gender-balanced participants, recruited from the local student and community populations, completed the study. Respective means (SD) of age, body mass, stature, and body mass index were 26.8 (3.9) years, 178.4 (4.4) cm, 80.9 (5.0) kg, and 25.5 (2.2) kg/m$^2$ among the males, and 25.1 (3.1) years, 165.8 (4.3) cm, 62.5 (5.7) kg, and 22.7 (1.5) kg/m$^2$ among the females. Participants reported no current or recent (i.e., past 12 months) musculoskeletal disorders or injuries. Additional inclusion criteria, related to anthropometry, were adopted from BSE user instructions to ensure a proper fit to the BSEs. The research reported herein complied with the tenets of the Declaration of Helsinki and was approved by the Institutional Review Board at Virginia Tech. Informed consent was obtained from each participant prior to any data collection, and compensation was provided at $10/hr.

### 2.3.2 Experimental tasks and procedures

A repetitive material handling task was simulated in a laboratory environment to examine the effects of two BSEs under several conditions that varied in task symmetry and height. Participants performed trials of repetitive lowering/lifting using a wooden box (40 × 25 × 23 cm, with 4 cm handle clearance diameter), the mass of which was set to 10% of individual participant body mass. Participants were instructed to repetitively lower/lift the box for 4 minutes in different conditions (see below), at a pace of 10 lower/lift cycles per minute (i.e., 40 lower/lift cycles in each trial). Lifting pace was controlled by a digital metronome. The lifting pace and duration were chosen, similar to earlier work (Graham et al., 2011), to ensure a continuous lifting pattern while minimizing the development of muscular fatigue, and the load mass was standardized to account for differences in body sizes among the participants.

Participants performed the lowering/lifting trials in three conditions, involving different levels of *Height* and *Symmetry* (Figure 2.1), which were intended to reflect a range of working postures frequent in the performance of manual material handling tasks, such as in warehouses and distribution centers (Kuorinka et al., 1994). Symmetric lowering/lifting was done to/from two heights set based on individual anthropometry: mid-shank and knee level (respectively referred to as Sym_Ground and Sym_Knee hereinafter). Asymmetric lowering/lifting was done to/from a location 90º to the right of the mid-sagittal plane (referred to as Asy_Knee hereinafter). This latter task was only done at knee height, since reaching to/from mid-shank height was found to be challenging for many participants in pilot work. Note that the initial box location for all lowering/lifting conditions was set at individual waist height (i.e., anterior superior iliac spine), and the mid height of the shank was determined as the mid-point between the patella (knee height) and the lateral malleolus (ankle height). Target locations for placing the box were marked using wooden blocks, and the horizontal distances between the centers of the target locations on the table and on the floor were controlled (symmetric: 25 cm, asymmetric: 50 cm). Each lowering/lifting cycle involved the following in sequence: 1) participant standing in the upright posture facing the box; 2) grasping the box and lowering it down to the target location; 3) lifting the box back to the initial location; and, 4) returning to the original, upright posture. Participants were allowed to freely choose their lifting style and feet location while maintaining a consistent feet location for the entire trial.

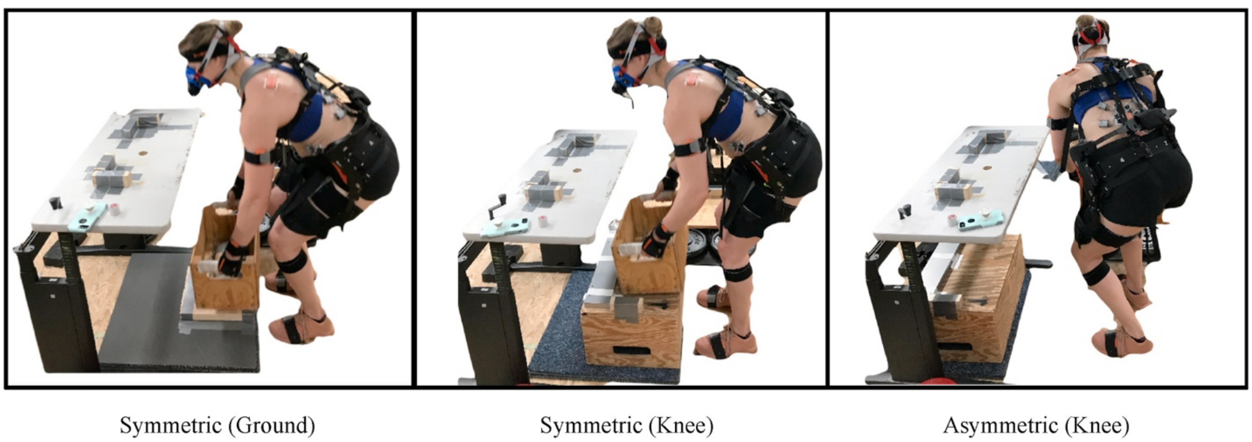


Figure 2.1. Illustrations of the repetitive lowering/lifting task in each of the three experimental conditions.

Each participant completed a training session (~1 hour) followed by an experimental session (~4.5 hours) on separate days. In the training session, participants were familiarized with the support levels and functionality of each BSE following manufacturers' manuals. Participants then practiced the lowering/lifting task in each of the three experimental conditions, during which they were also asked to determine their preferred lifting style, as well as exoskeleton support levels (for BackX™) and engagement angles (for Laevo™). Note that the BackX™ has four combinations of support, consisting of two modes (instant vs. standard) and two support levels (low vs. high). The instant mode provides assistive torque immediately after the wearer bends forward, while in the standard mode supportive torque is provided when trunk flexion reaches 30-45º. The Laevo™ allows for an adjustment of the engagement angle with respect to the upright posture. Distributions of preferred support levels for BackX™ and engagement angles for Laevo™ are provided in Appendix A. Participants were familiarized with providing ratings of perceived exertion (RPE), by holding a 1-kg weight in their outstretched arm (i.e., fully extended arm with elbow locked and shoulder flexed at 90º) to the maximum of their endurance and providing intermittent RPE ratings using the Borg (2004) CR-10 scale.

In the experimental session, participants completed a total of nine lowering/lifting trials, involving the factorial combination of three *Interventions* (i.e., BackX™, Laevo™, and Control = no BSE) and the three *Task Conditions.* Presentation orders of *Intervention* and *Task Condition* were each counter-balanced using 3 × 3 Latin squares, and a minimum of 5-min of rest was provided between lifting trials.

### 2.3.3 Instrumentation and Data Processing

Surface electromyography (EMG) was recorded bilaterally from five muscle groups in the lower lumbar, thoracic, abdominal, and shoulder regions, with electrode placements based on previous guidelines (Hermens et al., 1996; Cram, 2010). Specific muscle groups were: thoracic erector spinae (TES), iliocostalis lumborum (ILL), rectus abdominis (RA), external oblique (EO), and anterior deltoid (AD). After appropriate skin preparation (abrasion and cleaning with alcohol), pairs of pre-gelled, bipolar Ag/AgCl electrodes with a 2.5 cm inter-electrode spacing were placed over the noted muscle groups. These muscle groups were selected based on relevance to the simulated lifting tasks and accessibility across the devices used (e.g., different exoskeleton contact areas with the body, due to straps, pads, and rod locations).

Following electrode placement, participants completed a series of maximum voluntary isometric contractions (MVICs) for each muscle group, using a commercial dynamometer (Biodex System 3 Pro, Biodex Medical Systems Inc., NY, USA) and a custom frame to isolate the pelvis and lower extremities. For the trunk extensor (TES and ILL) and flexor (RA and EO) muscles, participants stood in the frame (positioned next to the dynamometer) with their trunk flexed 20º and performed trunk flexion, extension, and bidirectional axial rotation (Marras & Mirka, 1993a; Jia et al., 2011; Madinei et al., 2018). For the AD muscle, arm flexion was performed with the shoulder flexed at 90º (Boettcher et al., 2008). MVICs were replicated twice for each muscle group, during which non-threatening verbal encouragement was provided. Rest breaks of 30 seconds or longer were provided between MVICs.

Raw EMG signals were sampled at 1.5 kHz during MVICs and the lowering/lifting trials, using a telemetered system (TeleMyo Desktop DTS, Noraxon, AZ, USA). These signals were band-pass filtered (20-450 Hz, 4$^{th}$-order Butterworth, bidirectional) and subsequently low-pass filtered (3 Hz cut-off, 4$^{th}$-order Butterworth, bidirectional) to create linear envelopes. Processed EMG signals were normalized (nEMG) to maximum values collected during MVICs. As in earlier work (Potvin et al., 1990; Frost et al., 2009; Graham et al., 2009; Madinei et al., 2020b), four separate metrics were calculated to represent the level of muscular activity. For the first two metrics, peak (95$^{th}$ percentile) levels of the trunk extensor muscles (TES and ILL) were averaged on each side, yielding $\mathrm{TEM_L} = \frac{\mathrm{TES_L}+\mathrm{ILL_L}}{2}$ and $\mathrm{TEM_R} = \frac{\mathrm{TES_R}+\mathrm{ILL_R}}{2}$..The third metric was obtained as the sum of the peak levels of all bilateral trunk muscles (TES, ILL, RA, and EO), and is subsequently referred to as TTM (total trunk muscle activity). The fourth metric was calculated as the sum of the peak levels of the bilateral AD, and is subsequently referred to as SM (shoulder muscles). Each EMG metric was obtained separately for the lowering and lifting phases (see below for phase determination).

Energy expenditure (metabolic cost) was determined through respiratory data collected using indirect calorimetry (CosMed K5, CosMed, Rome, Italy), the accuracy and precision of which has been reported earlier (Perez-Suarez et al., 2018). Prior to each experimental session, a calibration procedure was completed following the manufacturer's guidelines. Breath-by-breath oxygen and carbon-dioxide uptake rates (mL/min) collected from the calorimeter were smoothed using a $4^{th}$-order, low-pass, bidirectional, Butterworth filter with a cutoff frequency of 0.33 Hz. The Brockway (1987) equation was used to estimate relative energy expenditure rates (kcal/kg•min), and this approach accounted for both participant body mass and BSE mass (if used). Based on pilot work and earlier evidence (Bilzon et al., 2001; Åstrand et al., 2003), steady-state metabolic rate was determined by averaging relative energy expenditure rates over the last 1.5 minutes of each trial.

Segmental body kinematics were monitored at 60 Hz using a wearable inertial motion capture system (MVN Awinda, Xsens Technologies B.V., Netherlands). The standard rotation sequence recommended by ISB (ZXY) was used to analyze kinematic data (Wu et al., 2005). Kinematic data were filtered using a $4^{th}$-order, low-pass, bidirectional Butterworth filter with a cut-off frequency of 5 Hz. For each trial, peak ($95^{th}$ percentile) triaxial angular velocities – axial rotation (AR), lateral bending (LB), and flexion/extension (FE) – were determined for the trunk and lumbar spine (thorax vs. pelvis). Angular velocities were separated for the lowering and lifting phases. Triaxial ranges-of-motions (ROMs) were also obtained, and mean values were calculated across lowering/lifting cycles. Maximum and minimum trunk inclination angles (with respect to the upright posture) were used to identify the beginning and ending of each lowering/lifting phase. Note that angular velocities and ROMs were used here as outcome measures to capture potential differences in the lifting methods employed between *Interventions*.

After completing each trial, participants reported ratings of perceived discomfort (RPDs) resulting from exoskeleton contact with their body – specifically at the chest, waist, and thighs – using 7-point Likert scales (Kuijt-Evers et al., 2007), where 0 = no discomfort and 6 = extreme discomfort (Appendix B). Subsequently, they reported RPEs for the shoulders, lower back, abdominal region, and legs, using the Borg CR-10 scale. For both RPD and RPE ratings, participants provided a single overall value for bilateral body parts. After completing all three trials with a given BSE, participants provided an overall usability score for that BSE on a continuous scale from 0 (not helpful at all) to 100 (absolutely helpful), and rated exoskeleton fit, comfort, and movement hinderance using separate 7-point Likert scales (Appendix B). Finally, they were asked to select their preferred BSE after completing all lifting trials.

#### 2.3.4 Statistical Analyses

Separate three-way, mixed-factor analyses of variance (ANOVAs) were used to assess the effects of *Intervention*, *Task Condition*, and *Gender* on each of outcome measures (EMG metrics, energy expenditure, joint angular velocities and ROMs, RPEs, and RPDs). The presentation orders of *Task Condition* and *Intervention* were included as a blocking effect. For EMG metrics and angular velocities, these ANOVAs were performed separately for results obtained in the lowering and lifting phases. Two-way ANOVAs were used for usability ratings, with independent variables of *Intervention* and *Gender*. Parametric model assumptions were assessed, and in several cases transformations were needed to achieve normally-distributed residuals. All subjective measures and usability ratings were analyzed using parametric analysis, the robustness of which was reported earlier (Rickards et al., 2012; Mircioiu & Atkinson, 2017). Summary outcomes were back-transformed and are reported as least squares means (95% CIs), and effect sizes are reported using eta-squared ($\eta^2$). Significant interaction effects were explored using simple-effects testing, and *post hoc* paired comparisons were completed using the Tukey-Kramer procedure where relevant. Given the study goals, the subsequent presentation of results and the discussion emphasizes the main and interaction effects of *Intervention*. All statistical analyses were performed using JMP Pro 14 (SAS, Cary, NC), using the restricted maximum likelihood (REML) method, with statistical significance concluded when $p < 0.05$.

## 2.4 Results

### 2.4.1 Muscle Activity

A complete summary of ANOVA results for each outcome measure is provided in Appendix C. ANOVA results for nEMG measures are summarized in Table 2.1 (see Appendix C). *Intervention* main effects were significant for $TEM_L$, $TEM_R$, and TTM during both the lowering and lifting phases. Using the BackX™ significantly reduced the levels of $TEM_L$, $TEM_R$, and TTM, respectively by 20.0, 18.3, and 17.3% during the lowering phase, and by 11.9, 11.9, and 10.4% during the lifting phase (Figure 2.2). Laevo™ use also led to significant reductions in $TEM_L$, $TEM_R$, and TTM, but only during the lowering phase, with respective magnitudes of 9.3, 8.7, and 7.8% (Figure 2.2). Further, a significant *Intervention* × *Gender* interaction effect was found for $TEM_L$ during lowering (Figure 2.3). All simple effects were significant, with *Intervention* significant for both *Genders* ($p \leq 0.0006$), and *Gender* significant for all *Interventions* ($p \leq 0.0037$). Females experienced a significant reduction when using either BSE (by 22.4% with BackX™ and

11.5% with Laevo™), while males exhibited a significant reduction only when using the BackX™ (by 16.5%).

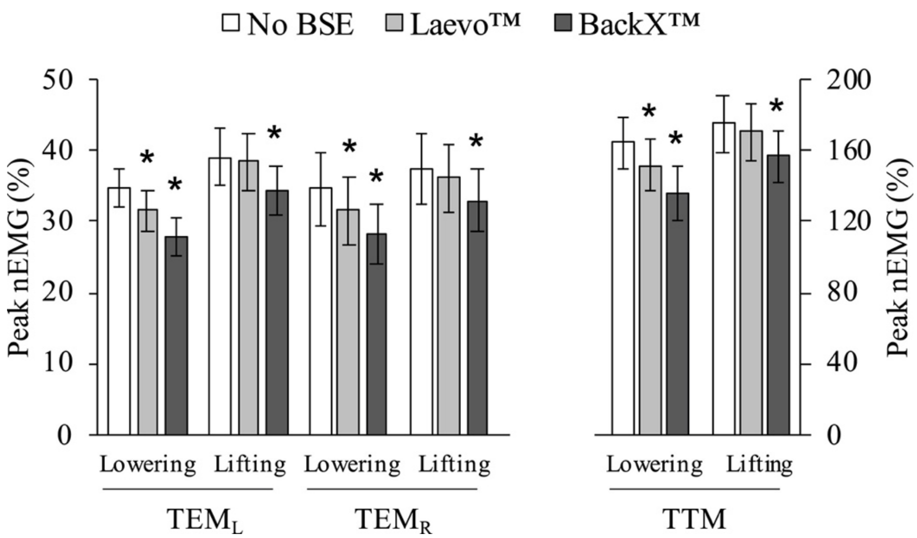


Figure 2.2. *Intervention* effects on metrics of peak normalized muscle activity (nEMG). Note that * denotes significant differences from the control condition (i.e., no BSE), and error bars indicate 95% confidence intervals.

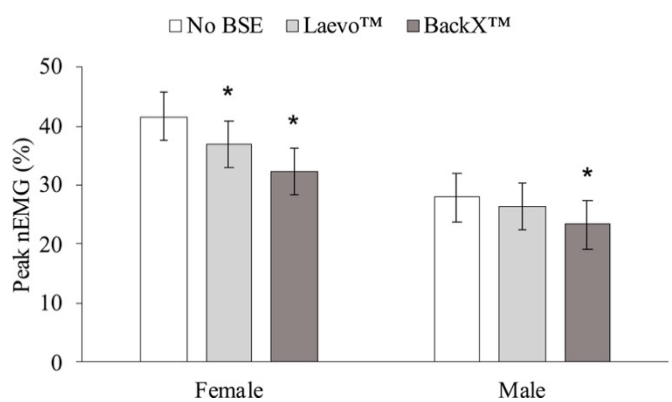


Figure 2.3. *Intervention* × *Gender* interaction effect on peak normalized activity (nEMG) of the left trunk extensor muscles (TEML). Note that * denotes significant differences from the control condition (i.e., no BSE), and error bars indicate 95% confidence intervals.

## 2.4.2 Energy Expenditure

There were both significant main and interaction effects of *Intervention* on energy expenditure (Table 2.2 Appendix C). Regarding the *Intervention* × *Gender* interaction effect (Figure 2.4), *Intervention* was significant for both *Genders* ($p < 0.0091$), while *Gender* was only significant for BackX™ ($p < 0.0001$). Use of either BSE resulted in a significant reduction in energy expenditure for females (by 8.9% with Laevo™ and 13.2% with BackX™), while males experienced a significant reduction only when using the Laevo™ (by 6.4%). Regarding the *Intervention* × *Task Condition* interaction effect (Figure 5), all simple effects were significant, excepting the effect of *Intervention* in the asymmetric condition ($p = 0.56$). Use of either BSE led to a significant reduction in energy expenditure in both symmetric conditions. Specifically, in the Sym_Ground condition, the reductions were 9.5% with Laevo™ and 13.6% with BackX™; while in the Sym_Knee condition these reductions were 10.2% with Laevo™ and 8.1% with BackX™. In contrast, no significant reductions were observed in the asymmetric condition.

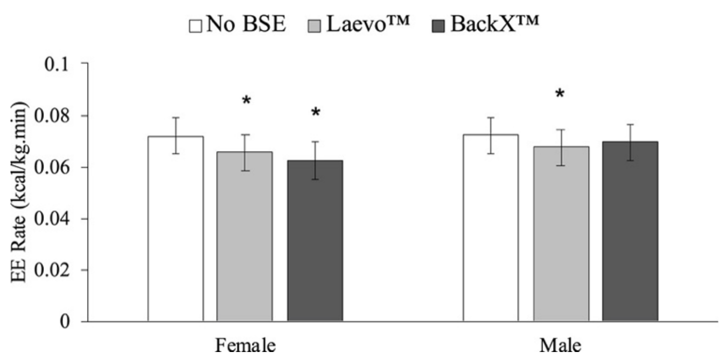


Figure 2.4. *Intervention* × *Gender* interaction effects on relative energy expenditure rate (EE-rate). Note that * denotes significant differences from the control condition (i.e., no BSE), and error bars indicate 95% confidence intervals.

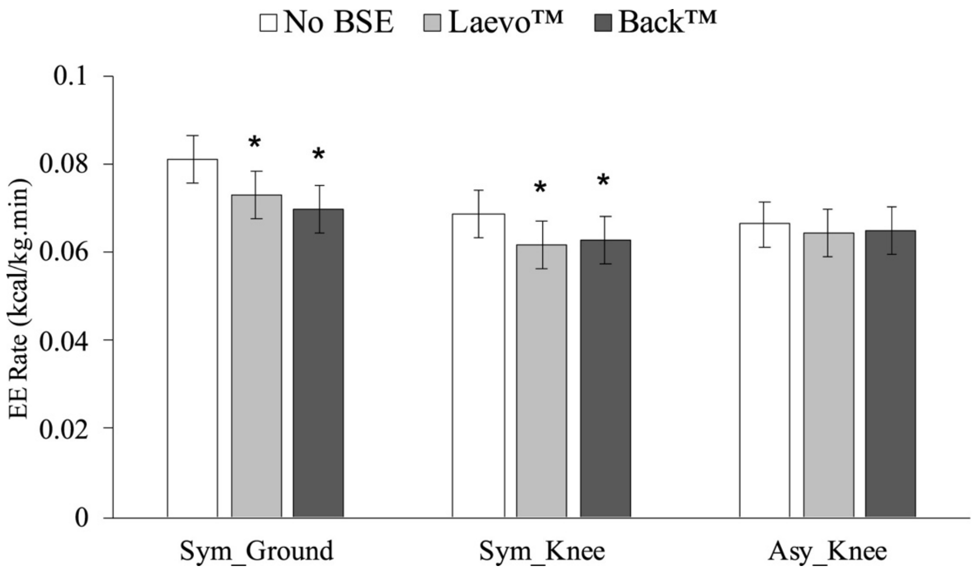


Figure 2.5. *Intervention* × *Task Condition* interaction effects on relative energy expenditure rate (EE-rate). Note that * denotes significant differences from the control condition (i.e., no BSE), and error bars indicate 95% confidence intervals.

### 2.4.3 ROM

A summary of ANOVA results for kinematic measures is presented in Table 2.3 (Appendix C). The *Intervention* × *Task Condition* interaction effect was significant for $Trunk_{FE}$, $Lumbar_{AR}$, and $Lumbar_{FE}$. For $Trunk_{FE}$, the effect of *Task Condition* was significant for all *Interventions* ($p <$ 0.0001). In contrast, *Intervention* was not significant for any *Task Condition* ($p > 0.0513$), and differences between *Interventions* within a given *Task Condition* were relatively small (less than ~5º). Similarly, for $Lumbar_{AR}$ the effect of *Task Condition* was significant for all *Interventions* ($p$ < 0.0001), while *Intervention* was only significant in the asymmetric condition ($p = 0.0001$). Regarding $Lumbar_{FE}$, the effect of *Task Condition* was significant for all *Interventions* ($p <$ 0.0001), while *Intervention* was only significant in the Sym_Knee condition ($p = 0.0304$). Further analyses revealed that neither BSE significantly influenced $Lumbar_{AR}$ or $Lumbar_{FE}$.

### 2.4.4 Angular Velocity

The main effect of *Intervention* was significant for $Trunk_{FE}$ while lowering (Table 2.4 – Appendix 3). Using the Laevo™ significantly decreased $Trunk_{FE}$ while lowering, by 7.1% compared to the no BSE condition [Laevo™ = 52.6 (48.9, 56.5) °/s vs. no BSE = 56.6 (52.7, 60.5) °/s]. The *Intervention* × *Task Condition* interaction effect was significant for both $Trunk_{FE}$ and $Lumbar_{FE}$ during lifting. Regarding the former, all simple effects were significant, excepting the effect of *Intervention* in the Sym_Knee condition ($p = 0.346$). Using the Laevo™ significantly reduced $Trunk_{FE}$ when lifting in the Sym_Ground condition compared to no BSE [Laevo™ = 74.3 (68.4, 80.5) °/s vs. no BSE =84.6 (78.3, 91.2) °/s]. In the Asy_Knee condition, however, neither BSE significantly influence $Trunk_{FE}$ during lifting. For $Lumbar_{FE}$ during lifting, all simple effects were significant, excepting the effect of *Intervention* in the Sym_Ground condition ($p = 0.12$). Again, no significant effects of BSE use were found for $Lumbar_{FE}$ during lifting.

Finally, there was a significant *Intervention* × *Task Condition* × *Gender* interaction effect on $Trunk_{LB}$ during lowering. Regarding this interaction effect, there were significant effects of *Intervention* × *Task Condition* for both *Genders* ($p < .0001$), and the *Gender* × *Task Condition* interaction was significant for all three *Interventions* ($p < .0001$), but the *Intervention* × *Gender* interaction was not significant for any *Task Condition* ($p > 0.092$). Post hoc analysis, however, revealed no significant effects of BSE use on $Trunk_{LB}$ during lowering.

### 2.4.5 Subjective Ratings

Ratings of perceived discomfort (RPD) at the chest, waist, and thighs were significantly affected by *Intervention* main effects and *Intervention* × *Gender* interaction effects (Table 2.5 - Appendix C, and Figure 2.6). Regarding the interaction effect on chest RPD, all simple effects were significant, excepting the effect of *Gender* using the BackX™ ($p = 0.948$). Both genders reported significantly higher RPDs at the chest when using the Laevo vs. BackX™. Simple effects analyses of waist RPDs indicated that *Intervention* was only significant for females ($p < 0.0001$), and *Gender* was only significant when using the BackX™ ($p = 0.036$). Females reported significantly higher RPDs at the waist when using the BackX™. Finally, simple effects analyses for thigh RPDs revealed that *Intervention* was only significant for females ($p = 0.0015$), and *Gender* was not significant for any *Intervention* ($p > 0.303$). Females reported significantly higher RPDs at the thighs when using the BackX™ vs. Laevo™.

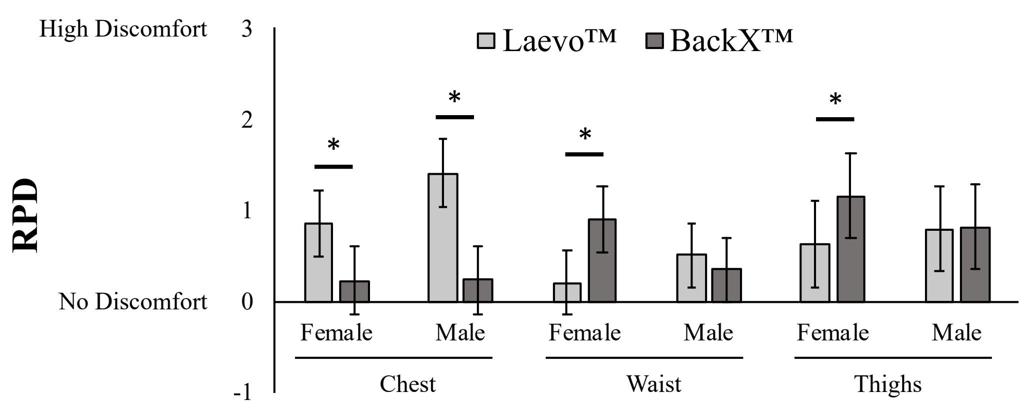


Figure 2.6. *Intervention* × *Gender* effects on ratings of perceived discomfort (RPD) at the chest, waist, and thighs. Note that * denotes significant differences from the control condition (i.e., no BSE), and error bars indicate 95% confidence intervals.

There were significant main effects of *Intervention* (Table 2.5 - Appendix C, Figure 2.7) on all RPE scores. Using the BackX™ significantly reduced RPE scores in the shoulders, lower back, legs, arms, and abdominal region compared to using no BSE, by 29.8, 40.2, 24.5, 17.6, and 34.9%, respectively. Laevo™ use also resulted in significantly reduced RPEs, by 22.8, 32.7, 24.9, 23.5, and 36.9%, respectively.

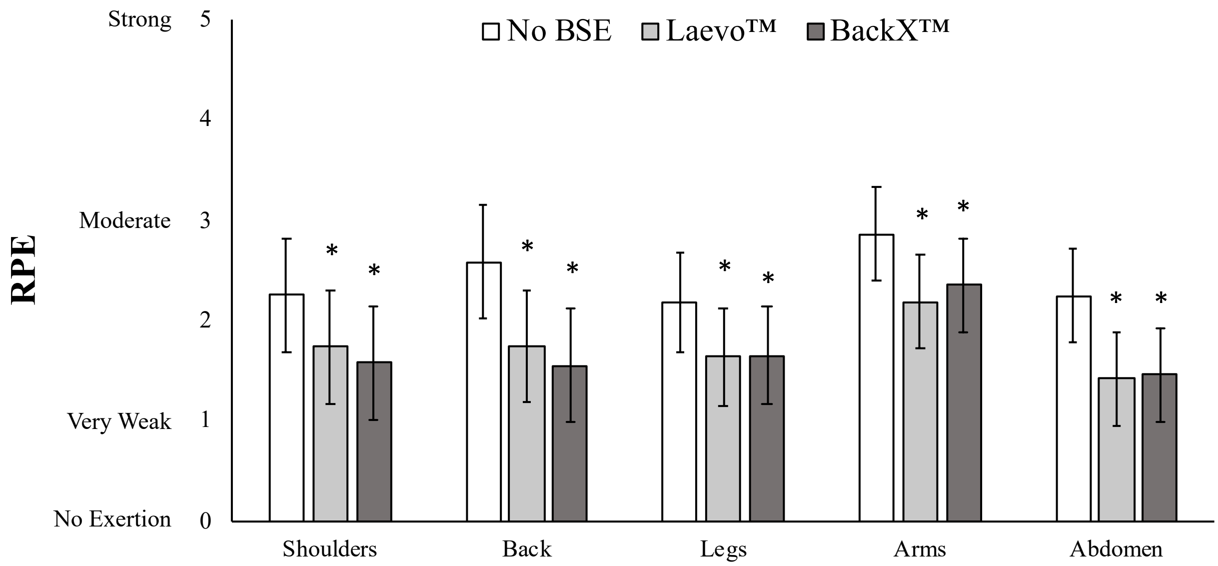


Figure 2.7. *Intervention* effects on ratings of perceived exertion (RPE) at the shoulders, back, legs, arms, and abdominal region. Note that * denotes significant differences from the control condition (i.e., no BSE), and error bars indicate 95% confidence interval

### 2.4.6 Usability Ratings

Figure 2.8 summarizes overall usability scores reported for both BSEs. Using either BSE was, on average, "moderately" to "very helpful" for both genders. There were no significant main or interaction effects of *Intervention* or *Gender* on overall usability scores ($p > 0.14$; Table 2.6 - Appendix C). A significant main effect of *Intervention* was found regarding perceived fit ($p < 0.003$); participants indicated a better fit using the BackX™ (5.0 (1.1)) than the Laevo™ (3.4 (1.4)) (Figure 9). There were also no significant main or interaction effects of *Intervention* or *Gender* on comfort ($p > 0.42$) or body movement restrictions ($p > 0.44$), with respective overall responses of 3.9 (1.3) and 3.4 (1.3). Finally, the BackX™ was slightly more preferred, with 10 participants (6 males and 4 females) preferred that BSE, and 8 participants (3 males and 5 females) preferring the Laevo™.

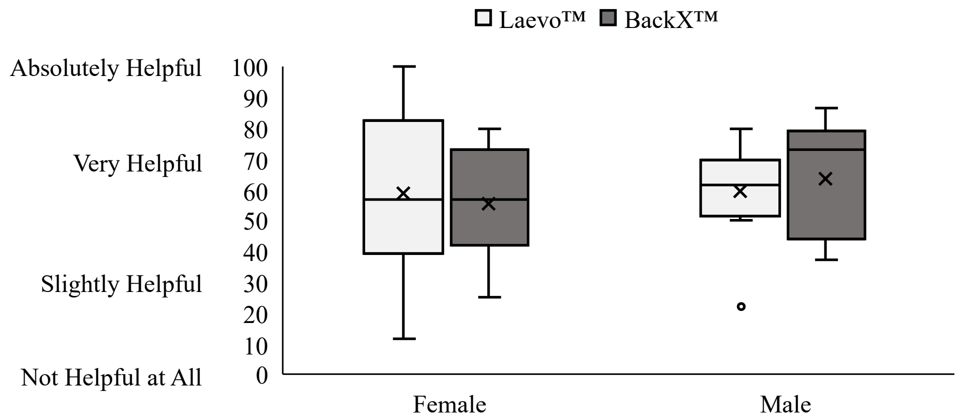


Figure 2.8. Responses to the question: “Overall, how helpful do you think the device was during the task?”, separated by gender and BSE. The symbol “×” indicates mean responses.

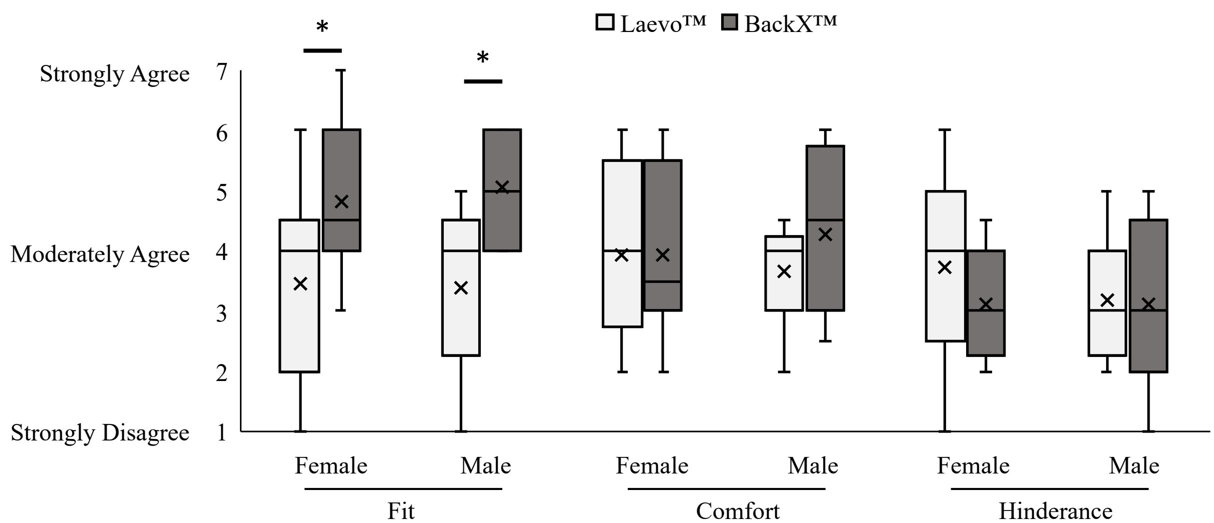


Figure 2.9. Responses to usability questions regarding overall fit, comfort, and body movement hinderance, separated by gender and BSE. The symbol “×” indicates mean responses, and * denotes significant effects.

## 2.5 Discussion

### 2.5.1 Trunk Muscle Activity

Both BSEs yielded significant reductions in trunk muscle activity, yet the magnitudes of these reductions were dependent both on the specific BSE and the lifting phase. BackX™ use significantly reduced $TEM_L$, $TEM_R$, and TTM, respectively by 20% (7.0% of MVIC), 18.3% (6.3% of MVIC), and 17.3% (28.5% of MVIC) during the lowering phase, and by 9.3% (3.2% of MVIC), 8.7% (3.0% of MVIC), 7.8% (12.8% of MVIC) during the lifting phase. Laevo™ use led to significant reductions only during the lowering phase, with respective reductions of 9.3% (3.2% of MVIC), 8.7% (3.0% of MVIC), 7.8% (12.8% of MVIC). The larger reductions in trunk muscle activity observed with the BackX™ vs Laevo™ might have resulted from participants selecting higher support settings with the former. For example (see Appendix A), most participants (6-8 males and 5-8 females, depending on the task condition) selected the highest level of support when using the BackX™ (i.e., instant mode, high level), but relatively "moderate" support settings when using the Laevo™ (i.e., 5-15° of cam angle). These differences in the choices of support settings might stem from the distinct design features providing supportive torque (i.e., BackX™ pulls the torso backwards while Laevo™ pushes against the chest), differences in the magnitude of assistive torque between the two BSEs, and/or differences in device fit.

In a secondary analysis, we examined peak nEMG of the abdominal muscles, which showed no significant main or interaction effects of *Intervention* on peak activity levels ($p$-values $> 0.11$), and with the magnitudes in the range of 2.7-9.8%, 2.9-10.3%, and 2.8-9.3% for the BackX™, Laevo™, and Control conditions, respectively.

Interestingly, larger reductions in trunk muscle activity were observed when using the BSEs during the lowering vs. lifting phases (Figure 2.2). An increased benefit of both BSEs while lowering may have resulted from a hysteresis effect present in the viscoelastic torque generation mechanisms. For example, Koopman et al., 2020b) reported that the assistive torque generated by the Laevo™ is considerably higher during the lowering phase compared to lifting. Future BSE designs might thus consider adapting the torque generation mechanisms, such as by minimizing this hysteresis effect or by providing consistent torque profiles during both trunk flexion and extension.

Both BSEs appeared to be more effective for females than males in some aspects (e.g., significant *Intervention* × *Gender* interaction effect on $TEM_L$, Figure 3), though the associated effect size was rather small ($\eta^2 \sim 0.01$). Specifically, both BSEs tested here reduced $TEM_L$ for females (by 22.4%, or ~9% of MVIC, using BackX™; and by 11.5%, or ~5% of MVIC, using Laevo™), while males experienced significant reductions only when using the BackX™ (by 16.5%, or ~5% of MVIC). We believe the larger reductions experienced by females may be due to their lighter torso mass, considering that preferred support settings were comparable across genders (Appendix A). Further research, however, is needed to determine if such gender-related differences arise primarily from anthropometric differences, or if instead there are inadequacies in BSE design approaches to accommodate both genders (e.g., device fit, torque magnitude).

TEM reductions found here are comparable to results from previous work. Specifically, Koopman et al. (2020b) found a significant reduction (by ~10%) in peak activity of the trunk extensor muscles when using the Laevo™ during "far-body" symmetric lifting; however, they found no significant reduction during "near-body" symmetric lifting. Comparable conditions in the current study were Sym_Knee and Sym_Ground, in which $TEM_L$ and $TEM_R$ significantly decreased (by ~9%), but only during the lowering phase. In a recent study, using BackX™ and Laevo™ reduced TEM activity by 24 and 17% during symmetric lifting, respectively, while the respective reductions were 13 and 15% during asymmetric lifting (Alemi et al., 2020). The only comparable condition in the current study was Sym_Ground, in which BackX™ use also yielded a larger reduction in TEM compared to Laevo™ (by 26% vs. 14% during lowering; 11% vs. 2% during lifting). The larger reductions observed by Alemi et al. (2020), however, might be due to the differences in task duration, external load magnitude, or the choice of support settings with each BSE.

Finally, no significant effects of *Intervention* or *Intervention*-related interaction effects were found for peak muscle activity in the shoulder (AD), which suggests that neither BSEs imposed adverse effects on upper extremity muscle groups during the lowering/lifting tasks. This finding is consistent with the results from previous work, wherein only minor changes in shoulder muscle activity were reported during symmetric and asymmetric lifting tasks with the BackX™ and Laevo™ (Alemi et al., 2020). However, it should be emphasized that only a single shoulder muscle was monitored here.

### 2.5.2 Kinematic Measures

Neither BSE substantially influenced triaxial trunk or lumbar ROMs. These results are in agreement with earlier work (Baltrusch et al., 2019; Koopman et al., 2020b), in that minor changes in $Lumbar_{FE}$ and $Trunk_{FE}$ ROM ($\leq 6^{o}$) were reported when using the Laevo™ during symmetric lifting from ankle or knee height. Similarly, no postural changes were reported when using a personal lift-assistive device (Abdoli-e et al., 2006; Abdoli-e & Stevenson, 2008), a biomechanically-assistive garment (Lamers et al., 2018), or a bending non-demand return device (Ulrey & Fathallah, 2013) during symmetric and/or asymmetric lifting tasks. Our results, along with this earlier evidence, suggests that no (or only minor) changes in postures occur when using a BSE. However, it is unclear if this is a beneficial outcome: increased trunk motion (i.e., flexion) during lifting tasks could serve to enhance the support provided by a BSE, but would also likely change the loads imposed on the spine.

In contrast to ROMs, we found a significant reduction in $\text{Trunk}_{\text{FE}}$ velocity (by ~7%) when using the Laevo™ during the lowering phase [Laevo™ = 52.6 (48.9, 56.5) °/s vs. no BSE = 56.6 (52.7, 60.5) °/s]. Wearing the Laevo™ also reduced $\text{Trunk}_{\text{FE}}$ velocity during the lifting phase in the Sym_Ground condition, by ~12% [Laevo™ = 74.3 (68.4, 80.5) vs. no BSE =84.6 (78.3, 91.2) °/s]. While the magnitudes of these reductions are rather small, we believe these changes might stem from potentially inadequate assistive torque provided by the Laevo™. As shown above, when using the Laevo™ participants increased their trunk muscle activity (in comparison to BackX™ use), perhaps to compensate for inadequate assistive torque generated by this device. As a result, they might have attenuated their lifting speed to mitigate mechanical demands on the spine. These outcomes are consistent with earlier evidence, in which up to an 18% reduction in peak $\text{Trunk}_{\text{FE}}$ velocity was found during symmetric lifting with the Laevo™ (Koopman et al., 2020b). The larger reduction in $\text{Trunk}_{\text{FE}}$ velocity reported by Koopman et al. (2020b) vs. here, though, might be due to differences in the simulated lifting conditions or external load magnitude, or the fact that their study controlled the support settings (at low and high cam angles).

### 2.5.3 Energy Expenditure

We found a significant *Intervention* × *Task Condition* interaction effect, in which using either BSE significantly reduced energy expenditure but only in the two *symmetric* conditions. Specifically, reductions in the Sym_Ground condition were 9.5 and 13.6% with the Laevo™ and BackX™, respectively; respective reductions in the Sym_Knee condition were 10.2 and 8.1%.

Contrarily, no significant reductions in energy expenditure were observed in the *asymmetric* condition. The inability of either BSE to reduce energy expenditure in the asymmetric condition might stem from the torque generation mechanisms in these exoskeletons, which appear to be inefficient beyond moderately symmetric conditions. Earlier work also reported no significant effects of Laevo™ and BackX™ use in reducing the energy expenditure during standing and asymmetric lifting tasks (Alemi et al., 2020).

Females here experienced a greater reduction in energy expenditure when using either BSE. Wearing the BackX™ and Laevo™ respectively resulted in 13.2 and 8.9% reductions in energy expenditure among females, while males experienced a significant reduction only when using the Laevo™ (by 6.4%). This gender difference might be due to the lower torso weight among females, again given that preferred support settings were comparable across genders (Appendix A). Such gender-related differences could also arise from a difference in the adequacy of support settings to provide an "optimum" level of resistive torque, such as based on core strength. Similar to the suggestion above regarding muscle activity, future work seems needed to differentiate whether gender-related differences in the effects of BSE use on energy expenditure are secondary to anthropometric differences or reflect design limitations to accommodate both genders.

Energy expenditure reductions found here with BSE use are in agreement with recent work (Alemi et al., 2020) that found a significant reduction during 5-min of symmetric lifting of a 6.8-kg box from the ground (12.6% with BackX™ and 8.9% with Laevo™), while reductions in the asymmetric lifting were not significant. Baltrusch et al. (2019) further reported a non-significant ~8% reduction and a significant ~17% reduction for the Low-cam and High-cam settings, respectively, when using the Laevo™ during 5-min of symmetric lifting of a 10-kg box from knee or ankle heights. A recent study of the VT-Lowe’s exoskeleton showed a significantly reduced metabolic cost, by up to 7.9% during 12-min of symmetric lifting of a box weighing 20% of body weight (Alemi, 2019). An earlier study, however, found no significant effect of a personal lift-assistive device on oxygen consumption during 15-min of symmetric lifting of a box requiring 10% of maximum back strength (Whitfield et al., 2014). This discrepancy might have resulted from differences in task duration, BSE design features, and/or lifting techniques employed. Future work is suggested to establish a standardized methodology for investigating the dependence of energy expenditure on specific BSE design approaches (e.g., torque-angle relationship) during diverse lifting tasks.

### 2.5.4 Perceived Discomfort and Exertion

Participants overall reported low-to-medium levels of discomfort at the chest, waist, and thighs (Figure 7). Both genders, though, reported significantly higher discomfort at the chest when using the Laevo™. Additional verbal feedback from the participants revealed that the relatively higher chest discomfort experienced when using the Laevo™ was likely due to its chest plate, which can pivot during movements causing the plate to rub against the chest during trunk movements, especially when twisting. Females also experienced higher waist and thigh discomfort when using the BackX™, which we believe resulted from the waist belt and leg pads of BackX™ not sufficiently accommodating them (vs. males). Specifically, females may have experienced more pressure from the waist belt and thigh pads, resulting in a higher discomfort, because of differences in hip width.

Perceived exertion at the shoulders, lower back, legs, arms, and abdominal region all significantly decreased when using either BSE (Figure 2.6). RPE reductions at the lower back were anticipated, since both BSEs are designed to offset external flexor moments on the torso by providing a counterbalance torque. Our observations regarding the lower back agree with earlier evidence of reduced levels of perceived exertion in this region when using BackX™ or Laevo™ (Alemi et al., 2020), or a personal lift-assistive device (Lotz et al., 2009), in manual lifting tasks, and are also consistent with the reductions in trunk muscle activity discussed earlier. Reductions in perceived exertion found here at other body regions suggest further that the external load transferred from the torso did not adversely affect other body regions, a conclusion supported by results regarding shoulder muscle activity (which indicated no adverse effect of BSE use on the anterior deltoid muscle).

### 2.5.5 Usability Ratings and User Feedback

Participants perceived using either BSE to be moderately to very helpful (Figure 8), and these ratings were comparable across genders. Ratings of fit showed that the BackX™ was superior for both genders (Figure 2.9). Participants moderately agreed with the question regarding overall comfort of the BSEs to wear at work, and the results were again comparable across genders. While responses regarding body movement hinderance when wearing the BSEs were also comparable across both genders and BSEs (Figure 2.9), females expressed concerns regarding movement hinderance when using the Laevo™ (mainly due to shifting and moving of the chest and thigh pads, and the fit of the waist pad and buttock belts). Males also commented on the contact between the torso rods and their ribcage while twisting. Finally, there was no clear indication of a preferred BSE, with slightly more participants preferring the BackX™ (10 participants out of 18) over the Laevo™ (8 participants), which is generally consistent with the other usability outcomes observed.

### 2.5.6 Limitation

A few limitations of the present study need to be noted. First, the sample only included young healthy adults (20-35 yrs.), so caution should be taken in generalizing the findings for an older population. Second, participants were familiarized with each BSE and practiced the lifting tasks only during an initial training session. Whether this training was sufficient for participants to benefit fully from the BSEs remains unknown. Third, we focused here on relatively short-term effects of different BSEs (i.e., 4 min. of repetitive lowering/lifting), and it is unclear if the BSE effects reported here can be generalized for more prolonged and/or frequent use of a BSE. Fourth, the current lifting tasks were simulations, performed in a controlled laboratory environment, and thus the relevance of our results to actual work settings, especially with suboptimal working conditions (uneven ground surfaces, restricted working space, etc.), warrants further investigation.

## 2.6 Conclusions

Occupational tasks involving repetitive lifting/lowering can be challenging to eliminate or modify in practice, and alternative interventions such as assistive devices are promising. We evaluated the effects of two commercially-available back-support exoskeletons on peak muscle activity, joint kinematics, energy expenditure, perceived discomfort and exertion (at shoulder, lower back, and leg), and usability, during symmetric and asymmetric lifting tasks. Using both BSEs reduced peak trunk extensor muscle activity (by ~9-20%, or 3-7% of maximum) and reduced energy expenditure (by ~8-14%). No substantial changes in trunk or lumbar kinematics were observed when using either BSE, which suggests that neither BSEs substantially influenced lifting methods. Use of both BSEs generally had positive impacts on subjective ratings. Our results further suggest that the beneficial effects of both BSEs are more pronounced in symmetric vs.

asymmetric lifting and lowering tasks, and that within these tasks there are differences between BSE designs. Future work is recommended to better characterize this task specificity and to determine the generalizability of BSE effects on objective and subjective outcomes among a wider range of task conditions and users.

## 2.7 Acknowledgment

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Appendix A

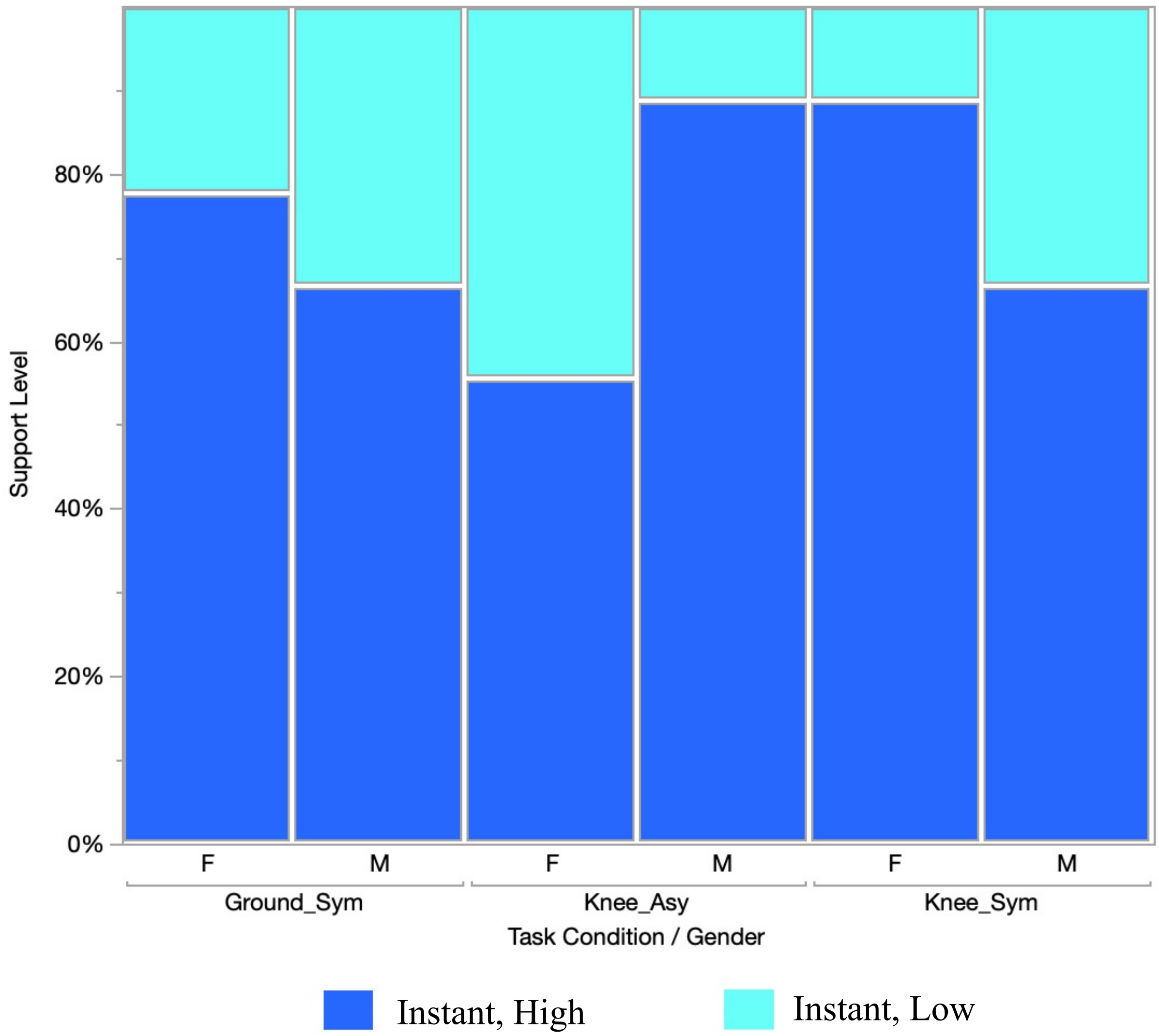


Figure 2.10. Preferred BackX™ support modes and levels

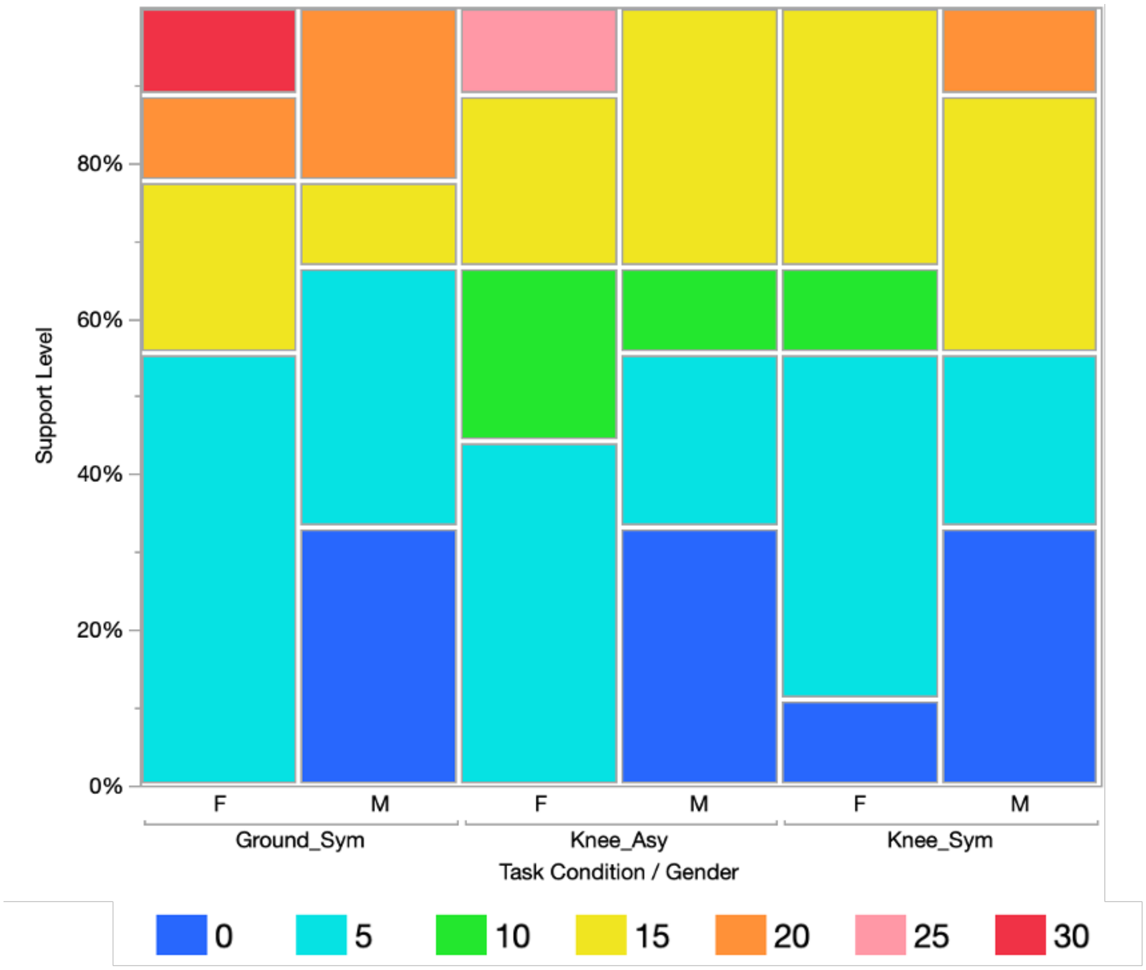


Figure 2.11. Preferred Laevo™ cam angles (0º: highest support, 30º: lowest support)

Appendix B

*Usability questions and associated scales:*

1. **Rating of perceived exertion**

| No exertion at all | Very, very light | Very light | Light | Moderate | Somewhat hard | Strong | | Very strong | | | Extremely Strong |
|---|---|---|---|---|---|---|---|---|---|---|---|
| 0 | 0.5 | 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 | 10 |

**2. Ratings of perceived discomfort**

| | **No discomfort at all** | | | **High discomfort** | | | **Extreme discomfort** |
|---|---|---|---|---|---|---|---|
| Chest (under and around the chest plate) | 0 | 1 | 2 | 3 | 4 | 5 | 6 |
| Waist (under and around the belt | 0 | 1 | 2 | 3 | 4 | 5 | 6 |
| Thigh (under and around the pad) | 0 | 1 | 2 | 3 | 4 | 5 | 6 |

**3. Usability questions**

Overall, to what extent do you agree with the following statements?

| | Strongly Disagree | | | Moderately Agree | | | Strongly Agree | **BackX™** | **Laevo™** |
|---|---|---|---|---|---|---|---|---|---|
| It fits well throughout all trials | 1 | 2 | 3 | 4 | 5 | 6 | 7 | | |
| It is comfortable to wear to work | 1 | 2 | 3 | 4 | 5 | 6 | 7 | | |
| It limits my movement (e.g., bending the back, moving an arm, walking) | 1 | 2 | 3 | 4 | 5 | 6 | 7 | | |

Overall, how helpful do you think the device was during the task?

| Not helpful at all | | | Slightly helpful | | Moderately helpful | Very helpful | | | | Absolutely helpful | **BackX™** | **Laevo™** |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| 0 | 10 | 20 | 30 | 40 | 50 | 60 | 70 | 80 | 90 | 100 | | |

## Appendix C

Table 2.1. Summary of ANOVA results regarding the main and interaction effects of *Gender*, *Intervention*, and *Task Condition* on $TEM_L$, $TEM_R$, TTM, and SM during the lowering and lifting phases. Each cell provides the *F* value, along with the associated *p* value and $\eta^2$ in parentheses. Note that significant effects are highlighted in bold font.

| Muscle | Phase | Trans. | Gender (G) | Intervention (I) | Task Condition (TC) | I × G | I × TC | G × TC | I × G × TC |
|---|---|---|---|---|---|---|---|---|---|
| $TEM_L$ | Lowering | | **18.77 (0.001, 0.368)** | **36.41 (<.0001, 0.096)** | **11.47 (<.0001, 0.030)** | **4.08 (0.019, 0.011)** | 1.65 (0.166, 0.009) | 2.86 (0.061, 0.008) | 0.05 (0.995, 0.0003) |
| | Lifting | Log | **12.99 (0.002, 0.343)** | **19.41 (<.0001, 0.037)** | **36.26 (<.0001, 0.070)** | 1.08 (0.345, 0.002) | 0.42 (0.797, 0.002) | 0.85 (0.431, 0.002) | 0.90 (0.468, 0.003) |
| $TEM_R$ | Lowering | Log | 3.58 (0.077, 0.139) | **27.71 (<.0001, 0.052)** | **29.64 (<.0001, 0.056)** | 0.43 (0.654, 0.001) | 1.95 (0.106, 0.007) | 0.85 (0.43, 0.002) | 0.20 (0.936, 0.001) |
| | Lifting | Log | **4.63 (0.047, 0.146)** | **9.68 (0.0001, 0.024)** | **56.59 (<.0001, 0.142)** | 0.19 (0.827, 0.001) | 0.09 (0.985, 0.001) | 1.20 (0.305, 0.003) | 0.03 (0.998, 0.0002) |
| TTM | Lowering | | **12.71 (0.003, 0.320)** | **29.11 (<.0001, 0.073)** | **10.61 (<.0001, 0.027)** | 2.60 (0.078, 0.007) | 1.68 (0.159, 0.008) | 0.83 (0.441, 0.002) | 0.11 (0.979, 0.001) |
| | Lifting | Log | **13.905 (0.002, 0.370)** | **15.0 (<.0001, 0.031)** | **16.04 (<.0001, 0.034)** | 0.29 (0.751, 0.001) | 0.30 (0.881, 0.001) | 0.18 (0.84, 0.0003) | 0.39 (0.815, 0.002) |
| SM | Lowering | | **6.90 (0.018, 0.258)** | 0.25 (0.781, 0.001) | 1.49 (0.229, 0.003) | 0.34 (0.711, 0.001) | 0.52 (0.723, 0.002) | 1.08 (0.343, 0.002) | 0.44 (0.782, 0.002) |
| | Lifting | | 2.20 (0.157, 0.084) | 2.79 (0.065, 0.008) | **35.72 (<.0001, 0.102)** | 0.16 (0.85, 0.001) | 0.60 (0.667, 0.003) | 2.30 (0.105, 0.007) | 0.96 (0.43, 0.006) |

Table 2.2. Summary of ANOVA results for the main and interaction effects of Gender, Intervention, and Task Condition on normalized energy expenditure. Entries are F values (with *p* values, $\eta^2$ in parenthesis), and significant effects are in highlighted using bold font.

| Effect | Trans | Gender (G) | Intervention (I) | Task Condition (TC) | I × G | I × TC | G × TC | I × G × TC |
|---|---|---|---|---|---|---|---|---|
| Energy Expenditure (kcal/kg•min) | | 0.43 (0.521, 0.016) | **20.60 (<.0001, 0.052)** | **59.83 (<.0001, 0.150)** | **5.19 (0.007, 0.013)** | **3.99 (0.005, 0.02)** | 2.33 (0.102, 0.006) | 0.76 (0.553, 0.004) |

Table 2.3. Summary of ANOVA results regarding the main and interaction effects of *Gender*, *Intervention*, and *Task Condition* on 3D Trunk and Lumbar ROMs. Each cell provides the $F$ value, along with the associated $p$ value and $\eta^2$ in parentheses. Note that significant effects are highlighted in bold font. AR: axial rotation; LB: lateral bending; FE: flexion/extension.

| | Direction | Trans. | Gender (G) | Intervention (I) | Task Condition (TC) | I × G | I × TC | G × TC | I × G × TC |
|---|---|---|---|---|---|---|---|---|---|
| Trunk | AR | Sqrt | 1.27 (0.276, 0.002) | 0.74 (0.479, 0.0003) | **2050.49 (<.0001, 0.935)** | 0.56 (0.571, 0.0003) | 0.20 (0.937, 0.0002) | **4.36 (0.015, 0.002)** | 0.99 (0.417, 0.0009) |
| Trunk | LB | Sqrt | 3.47 (0.081, 0.002) | 0.26 (0.77, 0.0001) | **1687.36 (<.0001, 0.949)** | 0.49 (0.615, 0.0003) | 0.81 (0.521, 0.0009) | 0.30 (0.743, 0.0002) | 1.85 (0.125, 0.002) |
| Trunk | FE | | 2.59 (0.127, 0.023) | 2.34 (0.101, 0.003) | **652.62 (<.0001, 0.744)** | 0.55 (0.576, 0.0006) | **2.92 (0.024, 0.007)** | **3.92 (0.022, 0.005)** | 0.70 (0.595, 0.002) |
| Lumbar | AR | Sqrt | 1.52 (0.236, 0.007) | 0.91 (0.406, 0.0009) | **868.81 (<.0001, 0.833)** | 0.004 (0.996, <.0001) | **5.83 (0.0003, 0.011)** | **12.09 (<.0001, 0.012)** | 0.24 (0.916, 0.0005) |
| Lumbar | LB | Log | 0.47 (0.501, 0.002) | 1.99 (0.141, 0.003) | **548.43 (<.0001, 0.820)** | 1.22 (0.299, 0.002) | 0.68 (0.609, 0.002) | **4.39 (0.015, 0.007)** | 0.58 (0.678, 0.002) |
| Lumbar | FE | Sqrt | 0.43 (0.521, 0.011) | 2.48 (0.088, 0.006) | **160.01 (<.0001, 0.40)** | 0.37 (0.69, 0.0009) | **2.68 (0.035, 0.013)** | 1.25 (0.291, 0.003) | 0.70 (0.591, 0.004) |

Table 2.4. Summary of ANOVA results regarding the main and interaction effects of *Gender*, *Intervention*, and *Task Condition* on 3D Trunk and Lumbar Velocities during the lowering and lifting phases. Each cell provides the $F$ value, along with the associated $p$ value and $\eta^2$ in parentheses. Note that significant effects are highlighted in bold font. AR: axial rotation; LB: lateral bending; FE: flexion/extension.

| | Direction | Phase | Trans. | Gender (G) | Intervention (I) | Task Condition (TC) | I × G | I × TC | G × TC | I × G × TC |
|---|---|---|---|---|---|---|---|---|---|---|
| Trunk | AR | Lowering | Sqrt | 0.72 (0.41, 0.0004) | 0.24 (0.786, 0.0001) | **3978.17 (<.0001, 0.973)** | 1.43 (0.244, 0.0003) | 0.17 (0.952, 0.0001) | 1.84 (0.163, 0.0005) | 1.47 (0.215, 0.0007) |
| | | Lifting | Sqrt | 0.89 (0.361, 0.0004) | 0.12 (0.887, <0.0001) | **3739.84 (<.0001, 0.976)** | 1.16 (0.316, 0.0003) | 0.15 (0.962, 0.0001) | 0.31 (0.733, 0.0001) | 1.17 (0.327, 0.0006) |
| | LB | Lowering | Sqrt | 1.15 (0.301, 0.0007) | 0.21 (0.813, 0.0001) | **2054.58 (<.0001, 0.956)** | 0.24 (0.787, 0.0001) | 0.79 (0.537, 0.0007) | 1.56 (0.214, 0.0007) | 2.69 (0.034, 0.003) |
| | | Lifting | Sqrt | 1.65 (0.218, 0.0008) | 0.01 (0.991, <0.0001) | **2368.17 (<.0001, 0.963)** | 0.16 (0.85, 0.0001) | 0.66 (0.624, 0.0005) | 1.74 (0.18, 0.0007) | 1.79 (0.136, 0.002) |
| | FE | Lowering | Sqrt | **5.70 (0.03, 0.033)** | **3.86 (0.024, 0.006)** | **528.47 (<.0001, 0.76)** | 0.10 (0.907, 0.0001) | 2.29 (0.064, 0.007) | 0.79 (0.456, 0.001) | 1.27 (0.286, 0.004) |
| | | Lifting | Sqrt | **5.04 (0.039, 0.032)** | 2.82 (0.064, 0.003) | **752.67 (<.0001, 0.779)** | 0.33 (0.719, 0.0003) | **3.99 (0.005, 0.008)** | **3.60 (0.03, 0.004)** | 0.93 (0.45, 0.002) |
| Lumbar | AR | Lowering | Log | 1.52 (0.235, 0.012) | 0.36 (0.698, 0.001) | **571.51 (<.0001, 0.761)** | 1.52 (0.224, 0.002) | 2.29 (0.063, 0.006) | 1.64 (0.198, 0.002) | 0.40 (0.806, 0.001) |
| | | Lifting | Log | 1.62 (0.221, 0.010) | 0.06 (0.944, 0.0001) | **846.08 (<.0001, 0.824)** | 1.47 (0.234, 0.001) | 2.21 (0.072, 0.004) | 1.60 (0.205, 0.002) | 0.08 (0.989, 0.0001) |
| | LB | Lowering | Log | 1.06 (0.319, 0.004) | 2.57 (0.081, 0.003) | **721.91 (<.0001, 0.844)** | 1.17 (0.314, 0.001) | 0.61 (0.66, 0.001) | **6.41 (0.002, 0.008)** | 0.81 (0.523, 0.002) |
| | | Lifting | Log | 0.98 (0.336, 0.004) | 1.82 (0.167, 0.002) | **797.92 (<.0001, 0.858)** | 0.45 (0.642, 0.0005) | 0.84 (0.5, 0.002) | **6.68 (0.002, 0.007)** | 0.78 (0.539, 0.002) |
| | FE | Lowering | Log | 0.29 (0.6, 0.009) | 0.80 (0.454, 0.002) | **101.94 (<.0001, 0.284)** | 0.41 (0.667, 0.001) | 2.38 (0.056, 0.013) | 1.13 (0.328, 0.003) | 0.33 (0.856, 0.002) |
| | | Lifting | Log | 0.32 (0.581, 0.010) | 0.78 (0.46, 0.002) | **118.59 (<.0001, 0.289)** | 0.26 (0.769, 0.0006) | **3.89 (0.005, 0.019)** | 2.51 (0.085, 0.006) | 1.06 (0.38, 0.005) |

Table 2.5. Summary of ANOVA results regarding the main and interaction effects of *Gender*, *Intervention*, and *Task Condition* on ratings of perceived discomfort (RPDs) and ratings of perceived exertion (RPEs). Each cell provides the $F$ value, along with the associated $p$ value and $\eta^2$ in parentheses. Note that significant effects are highlighted in bold font.

| | Body Part | Trans. | Gender (G) | Intervention (I) | Task Condition (TC) | I × G | I × TC | G × TC | I × G × TC |
|---|---|---|---|---|---|---|---|---|---|
| RPD | Chest | | 1.47 (0.243, 0.024) | **61.09 (<.0001, 0.2707)** | 1.29 (0.2815, 0.011) | **5.47 (0.022, 0.024)** | 2.78 (0.069, 0.025) | 0.89 (0.415, 0.0079) | 0.29 (0.749, 0.003) |
| | Waist | | 0.42 (0.525, 0.009) | **5.64 (0.02, 0.036)** | 0.37 (0.6903, 0.005) | **14.08 (0.0004, 0.090)** | 0.20 (0.818, 0.003) | 0.61 (0.547, 0.008) | 0.63 (0.538, 0.008) |
| | Thighs | | 0.10 (0.754, 0.003) | **6.72 (0.012, 0.030)** | 1.48 (0.2356, 0.013) | **4.82 (0.031, 0.022)** | 0.23 (0.794, 0.002) | 1.23 (0.3, 0.011) | 1.15 (0.322, 0.010) |
| RPE | Shoulders | | 0.0001 (0.991, 0.0001) | **9.68 (0.0001, 0.047)** | 0.008 (0.9918, <0.0001) | 0.94 (0.395, 0.005) | 1.67 (0.193, 0.008) | 0.05 (0.9954, 0.0005) | 0.77 (0.545, 0.007) |
| | Back | | 0.07 (0.801, 0.002) | **21.16 (<.0001, 0.105)** | 0.09 (0.9174, 0.0004) | 0.57 (0.567, 0.003) | 0.28 (0.758, 0.001) | 0.59 (0.674, 0.006) | 0.156 (0.96, 0.002) |
| | Legs | | 0.22 (0.649, 0.008) | **9.34 (0.0002, 0.045)** | 0.51 (0.6007, 0.003) | 2.47 (0.089, 0.012) | **3.46 (0.035, 0.017)** | 1.83 (0.127, 0.018) | 1.31 (0.272, 0.013) |
| | Arms | | 0.23 (0.639, 0.008) | **10.38 (<.0001, 0.061)** | 1.59 (0.2091, 0.009) | 0.28 (0.758, 0.002) | 1.55 (0.217, 0.009) | 0.26 (0.906, 0.003) | 0.12 (0.976, 0.001) |
| | Abdomen | | 0.20 (0.662, 0.006) | **19.65 (<.0001, 0.104)** | 0.63 (0.5365, 0.003) | 0.15 (0.865, 0.0008) | 0.32 (0.727, 0.002) | 0.47 (0.755, 0.005) | 0.37 (0.831, 0.004) |

Table 2.6. Summary of ANOVA results regarding the main and interaction effects of *Gender*, and *Intervention* on fit, comfort, body movement hinderance, and overall usability ratings. Each cell provides the $F$ value, along with the associated $p$ value and $\eta^2$ in parentheses. Note that significant effects are highlighted in bold font.

| | Trans. | Gender (G) | Intervention (I) | G × I |
|---|---|---|---|---|
| Fit | | 0.035 (0.855, 0.0008) | **12.247 (0.003, 0.2708)** | 0.101 (0.7545, 0.0022) |
| Comfort | | 0.003 (0.956, 0.0001) | 0.691 (0.4179, 0.0148) | 0.691 (0.418, 0.0148) |
| Hinderance | | 0.423 (0.5248, 0.0128) | 0.626 (0.44, 0.0184) | 0.435 (0.519, 0.0128) |
| Usability | | 0.219 (0.646, 0.0101) | 0.108 (0.743, 0.0003) | 2.7 (0.1039, 0.0074) |

# 3 Chapter 3: A Novel Approach to Quantify the Assistive Torques Generated by Passive Back-Support Exoskeletons

## 3.1 Abstract

Industrial exoskeletons are a promising ergonomic intervention to reduce the risk of work-related musculoskeletal disorders by providing external physical support to workers. Passive exoskeletons, having no power supplies, are of particular interest given their predominance in the commercial market. Understanding the mechanical behavior of the torque generation mechanisms embedded in passive exoskeletons is, however, essential to determine the efficacy of these devices in reducing the physical loads in manual material handling tasks. We introduced a novel approach to quantify the assistive torque profiles of two passive back-support exoskeletons (BSEs) using a computerized dynamometer. The feasibility of this approach was examined for both human subjects and a mannequin. Clear differences in assistive torque magnitudes were evident between the BSEs, and both devices generated more assistive torques during flexion than extension. The assistive torques obtained from the human subjects were often within similar ranges to those from the mannequin, though values were more comparable over a narrow range of flexion/extension angles. Characterizing exoskeleton assistive torque profiles can help in better understanding how to select a torque profile for given task requirements and user anthropometry, and assist in predicting the potential impacts of exoskeleton use by incorporating measured torque profiles in a musculoskeletal modeling system. Future work is recommended to assess this approach for other occupational exoskeletons.

## 3.2 Introduction

Occupational exoskeletons are becoming a new frontier of human-machine research; by supporting/augmenting the work capacity of a worker, these devices have the potential to reduce physical demands and prevent work-related musculoskeletal disorders. Most commercially-available occupational exoskeletons are passive systems, which include compliant elements (e.g., springs or other elastic materials) to provide external torques about a joint of interest (e.g., back or shoulder). These external torques are generated as function of the included angles between proximal and distal segments comprising the joint of interest. Compared to active/powered systems, passive exoskeletons are technologically more mature, and have been considered for a range of applications, including automotive manufacturing (Hensel & Keil, 2019; Ferreira et al., 2020; Kim et al., 2021), agriculture (Upasani et al., 2019; Thamsuwan et al., 2020), and construction (Kim et al., 2019). To promote the safe adoption and use of an exoskeleton, it is critical to determine the effects of using an exoskeleton in diverse task scenarios. Indeed, numerous lab- and field-based studies have quantified the biomechanical effects of different exoskeletons on the user (e.g., Lamers et al., 2018; Baltrusch et al., 2019; Koopman et al., 2019; Alemi et al., 2020; Koopman et al., 2020b; Madinei et al., 2020a; Kim et al., 2021). Given the resource-intensive nature of lab or field testing, however, and the increasing availability of different exoskeleton designs, an effective approach is needed to simulate or predict the impacts of exoskeleton use for potential occupational applications (Nussbaum et al., 2019).

Musculoskeletal modeling software could facilitate assessing exoskeleton use under various work scenarios. Indeed, there are several reports of the impacts of exoskeleton use or different exoskeleton designs (e.g., support torque profiles), in terms of estimated muscle activities and joint forces/torques, using the AnyBody Modeling System (Agarwal et al., 2016; Jensen et al., 2018; Fritzsche et al., 2021) or OpenSim (de Kruif et al., 2017; Khamar et al., 2019; Zhou & Chen, 2021). When simulating human-exoskeleton interactions, however, mechanical aspects of an exoskeleton need to be defined in the modeling software environment. These aspects include component inertial properties, available degrees-of-freedom, and supportive torque profiles (i.e., torque-angle relationships). The latter is of particular importance, since exoskeletons have distinct torque-generating mechanisms. Exoskeleton manufacturers, though, typically treat these torque profiles as proprietary, likely because the profiles can substantially affect the effectiveness of an exoskeleton and a user's perception. Furthermore, some mechanisms may not be purely elastic, instead exhibiting viscoelastic behaviors (e.g., speed dependency and hysteresis).

Though some exoskeleton torque profiles have been reported using simulated data (Bartel & Davy, 2006; Hyun et al., 2019), there are only few reports of results using direct measures. Koopman et al. (Koopman et al., 2019; Koopman et al., 2020b) measured the torque profile of a particular back-support exoskeleton (BSE; Laevo™ V2.4, Delft, Netherlands). Specifically, they placed a force transducer under the chest pad of the exoskeleton, likely because this pad is connected directly to the torque-generation mechanism at the hip though a structural beam, and monitored the kinematics of this mechanism using three LED markers attached to the beam. They reported no effect of flexion/extension speed on torque profiles, but that they there was hysteresis present (i.e., supportive torques were larger during trunk flexion than extension). This approach is seemingly straightforward and easy to implement, but it permits only a limited control over exoskeleton angular velocity since a participant needs to voluntarily flex/extend the trunk. Further, placing a force transducer under the chest pad of a BSE could present a practical challenge depending on the design of the pad (e.g., have a large surface area), or if the torque-generating mechanism at the hip is structurally connected to the backside of a user's trunk.

Thus, we describe here a novel, alternative approach to quantify torque profile(s) of an exoskeleton. This approach uses a computerized isokinetic dynamometer, which permits accurate control of trunk flexion/extension angles and angular velocities. With the proposed approach, torque profiles were assessed for two different BSEs in different support settings over a range of flexion/extension kinematics. We implemented this approach using both human subjects and a mannequin, to determine whether both might be feasible.

## 3.3 Methods

### 3.3.1 Back-support exoskeletons

We used the BackX™ Model S and Laevo™ V2.5, which incorporate passive torque-generating mechanisms about the hip that are intended to augment the trunk extensor muscles, yet which also have distinct design features. BackX™ support settings can be adjusted. Specifically, there are four combinations of support, consisting of two *modes* (instant vs. standard) and two *support levels* (low vs. high). The instant mode provides assistive torque immediately after the wearer bends forward, while in the standard mode supportive torque is provided when trunk flexion reaches ~35º. In contrast, the Laevo™ allows for an adjustment of the onset angle (0-35º flexion relative to the upright posture) at which support is initiated. The BackX™ was tested at low and high support levels in the instant mode, and the Laevo was tested at low (35º) and high (0º) onset angles; an additional condition was included for both BSEs, in which supportive torques were turned off.

### 3.3.2 Experimental design and procedures

To measure torque profiles (i.e., torque vs. angle/speed relationship), we utilized both the isometric and continuous passive motion (CPM) modes of a computerized isokinetic dynamometer (HUMAC NORM, CSMi, Stoughton, MA, USA). The former was to obtain torque profiles under static conditions (i.e., constant joint angle), and the latter to obtain torque profiles during motion (i.e., flexing/extending the joint of interest).

For a given BSE, separate torque profiles were measured in each of three supportive torque settings (i.e., $BSE_{OFF}$, $BSE_{LOW}$, and $BSE_{HIGH}$) under the static and dynamic conditions. These torque profiles were obtained using both an articulated mannequin (MZ-HM01, RoxyDisplay, East Brunswick, NJ) and human subjects. The mannequin had a hinge joint at the hip allowing for pure hip flexion/extension, and (obviously) avoided potential voluntary or reflexive muscle activation in dynamic conditions. Conversely, we considered that human subjects would involve more realistic motions of a BSE, since the mannequin lacked soft tissues and had a body shape not representative of actual populations.

<u>Use of the mannequin:</u> To measure torque profiles, the mannequin was properly fitted with a BSE, then positioned and secured on the dynamometer bed such that the rotational center of the BSE torque generation mechanism (i.e., left hip joint center of the mannequin) was aligned to the rotational center of the dynamometer motor. Then, the left thigh of the mannequin was connected to the hip adapter of the dynamometer (Figure 3.1). For the static condition, the isometric mode was configured to passively move the left thigh of the mannequin to 13 different hip joint angles (0, 10, 20, …, 120º), then each angle was maintained for 10 seconds. For the dynamic conditions, the CPM mode was configured to operate over a range from 0º (neutral hip angle) to 120º (hip flexion) at five different angular speeds (20, 40, 60, 80, and 100º/sec.) intended to capture typical

occupational task demands (e.g., Marras et al., 1993; Lavender et al., 2012). At each speed, the left thigh of the mannequin was driven passively and isokinetically over the set angular range 14 times, mimicking repetitive hip flexion and extension. In both static and dynamic conditions, dynamometer torques and angles were recorded at 100 Hz.

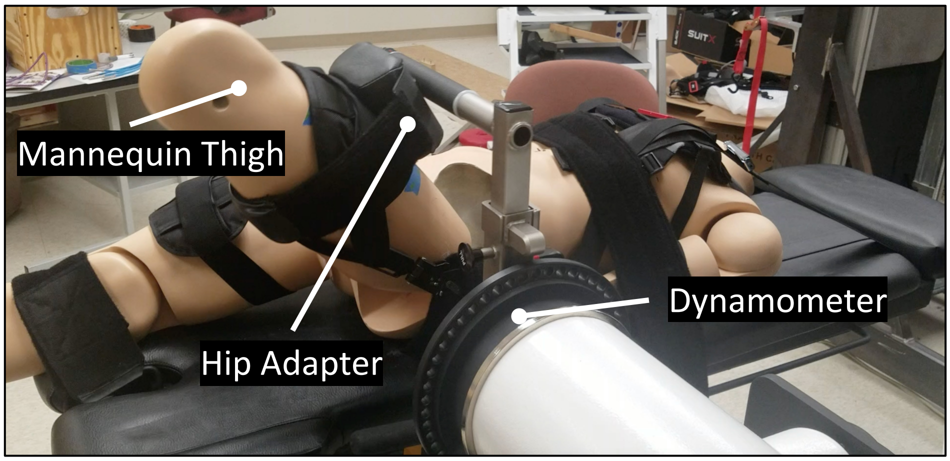


Figure 3.1. Illustration of the experimental setup using a mannequin lying prone on the bed of a dynamometer. The mannequin is "wearing" the BackX™, and the left "thigh" was positioned or moved using a hip adapter connected to the dynamometer.

Use of human subjects: Two participants with the respective mean (SD) age, height, and weight of 26.5 (3.5) yrs, 182.0 (5.7), and 70.0 (7.1) completed testing. The research procedures were approved by the Institutional Review Board at Virginia Tech, and informed consent was obtained from the participants prior to any data collection. Participants were fitted with a BSE and asked to stand in the trunk modular component (TMC) of the HUMAC NORM. The TMC foot plate on which participants stood was then vertically adjusted so that the rotational center of the BSE torque-generating mechanism was aligned to the rotational center of the TMC, and participants were secured to the TMC following the manufacturer's recommendations (Figure 3.2). For the static condition, as when the mannequin was used, the isometric mode was configured to passively move the trunk of participants to each of 10 trunk flexion angles (0, 10, 20, …, 90º), and measured isometric torques for 10 seconds. During dynamic trials, the CPM mode was configured to operate over a range of 0º (standing upright) to 90º (trunk flexion) at the same five joint speeds as when the mannequin was used. Note that we used 90º instead of 120º as the maximum trunk flexion angle due to a mechanical limitation of the TMC and to accommodate participant range-of-motion and comfort.

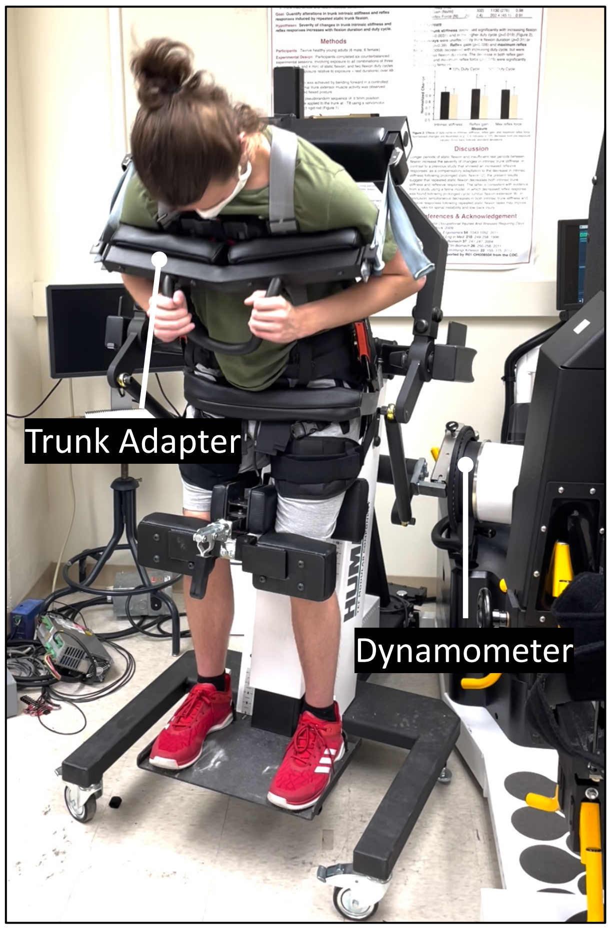


Figure 3.2. Demonstration of the experimental setup using human subjects. A participant is wearing the BackX™, and the trunk is positioned or moved using the trunk adapter of the dynamometer.

For each angular speed, participants first completed 10 training trials, during which they were repeatedly asked to limit abdominal muscle activity to minimize confounding effects on measured torques. This training was accomplished using real-time visual feedback of bilateral rectus abdominus (RA) activity, via normalized surface electromyography (EMG). Electrode placements were performed based on previous guidelines (Criswell, 2010). Participants completed initial maximum voluntary isometric contractions (MVICs), using the dynamometer to isolate the pelvis and lower extremities. Participants were secured to the trunk adapter with their trunk flexed 20° and performed active trunk flexion (Marras & Mirka, 1993b). MVICs were replicated twice, during which non-threatening verbal encouragement was provided. Raw EMG signals were recorded at 2 kHz using a telemetered system (TeleMyo Desktop DTS, Noraxon, AZ, USA), and rest breaks of 30s or longer were provided between MVICs. A rest break of 1 minute or longer was provided after the training trials. EMG signals were band-pass filtered (20–450 Hz, 4th-order Butterworth, bidirectional) and subsequently low-pass filtered (3 Hz cut-off, 4th-order Butterworth, bidirectional) to create linear envelopes. During dynamic trials, the trunk of participants was driven passively and isokinetically over the set range-of-motion 14 times. Processed EMG signals during the dynamic trials were normalized (nEMG) to maximum values collected during MVICs; nEMGs were displayed during the noted training trials via a monitor at roughly the participant's waist level.

### 3.3.3 Data reduction and outcome measures

Torque data were low-pass filtered ($2^{nd}$-order, bidirectional Butterworth filter, cut-off frequency = 9 Hz). Torque and angle data were resampled at 100 points per angle (e.g., 120º x 100 = 12,000 data points for the mannequin) for subsequent analysis. These data were then separated into flexion and extension phases, and mean values were obtained across the 14 trials. Dynamic torques reported hereafter are for the low speed only (20º/sec), at higher speeds there were more substantial effects of the dynamometer acceleration/deceleration (see Results). Exoskeleton-generated torque profiles were obtained by subtracting angle-specific torques in the $BSE_{OFF}$ condition from each respective support conditions ($BSE_{HIGH}$ and $BSE_{LOW}$). Assistive torque profiles are reported unilaterally (per torque-generating mechanism). Peak (95th percentile) nEMG values for the bilateral RA were obtained to characterize abdominal muscle activity.

## 3.4 Results

No substantial effects of speed in dynamic conditions were apparent for the assistive torques of either BSE during the flexion and extension phases over the range of 20-100º/sec (Figure 3.4 – Appendix D). Different torque profiles were recorded towards the beginning and ending phases of movement, likely due to the acceleration/deceleration of the hip adapter. Both BSEs generated more assistive torque during flexion than extension (Figures 3.3A-D). The maximum torques generated by BackX™ were respectively 24.8, 19.2, and 17.9 Nm for the flexion, extension, and static conditions at high support, and 14.7, 11.3, and 11.3 Nm at low support. The respective values for the Laevo™ were 9.7, 6.4, and 8.5 Nm at high support, and 7.9, 6.4, and 5.9 Nm at the low support condition. Note that these values are torque outputs per unilateral mechanism.

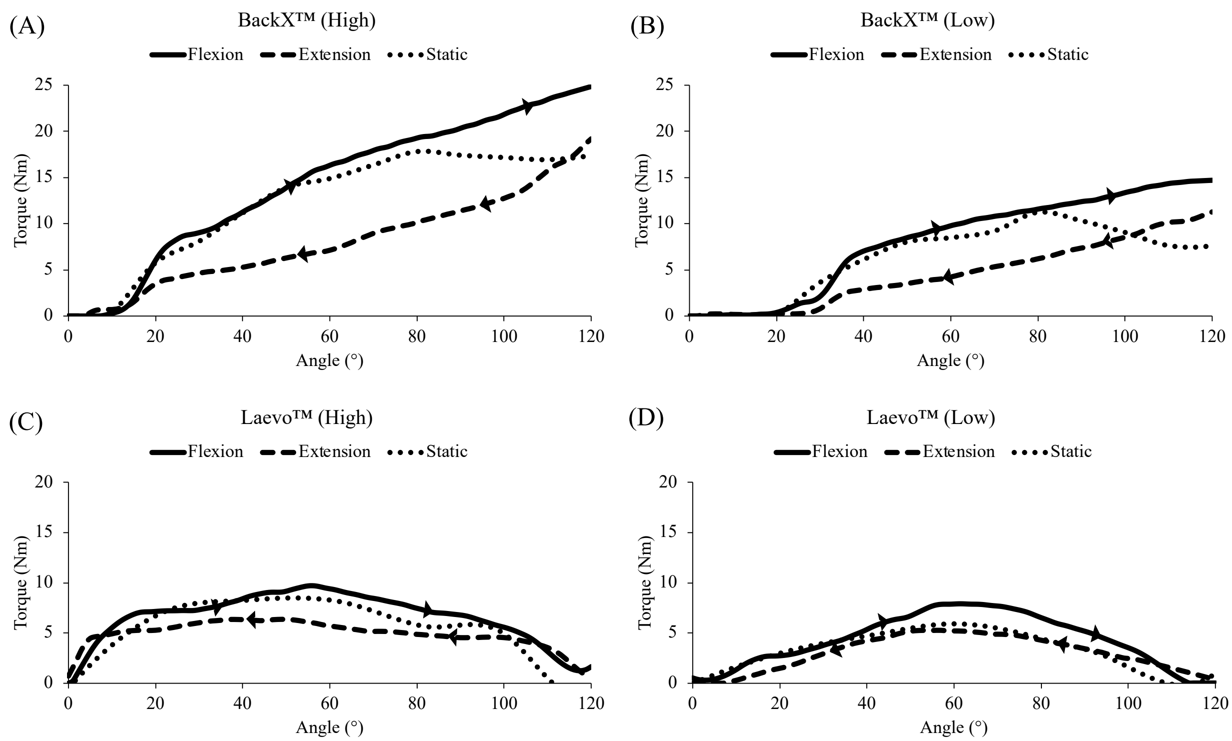


Figure 3.3. Demonstration of the torque profiles obtained from the mannequin with the BackX™ and Laevo™ during flexion, extension, and static phases, separated by support settings (High and Low). Torque profiles are for each (unilateral) mechanism, with dynamic data shown for the 20º/sec condition, and angles indicate hip flexion.

The assistive torques obtained from the human subjects were often within similar ranges to those from the mannequin, though values were more comparable over a narrow range of flexion/extension angles (i.e., 30º-60º; Figure 3.5 – Appendix D). Within this range, torque differences were typically less than ~5 Nm (~14-60% difference). Further, peak RA activity was relatively low overall (<~8 %MVIC) and remained consistent when using either BSE at the different flexion/extension speeds (Table 1 – Appendix D).

## 3.5 Discussion

We developed and applied an approach to measure the assistive torque profiles of a BSE using a computerized dynamometer and characterized torque profiles of two BSEs in different support settings (high and low) in both static and dynamic conditions. With this approach, we obtained torque profiles by manipulating hip flexion/extension angles of a mannequin, and trunk flexion/extension angles of a human subject. For both inanimate and animate subjects, comparable assistive torques were obtained, yet use of the latter produced effective torque profiles over a relatively narrower range of joint angles due to several practical limitations as discussed below.

In the dynamic condition (i.e., flexion/extension speed: 20-100º/sec.), larger assistive torques were generated during the flexion vs. extension phases (Figure 3.3), consistent with the finding of Koopman et al. (2020; 2019). They also reported no effect of flexion/extension speed on torque profiles and noted that the difference between torques during the flexion and extension phases is due to friction in the torque generation mechanism. We similarly found no obvious effect of controlled flexion/extension speeds (Figure 3.4 – Appendix D). Both the BackX™ and Laevo™ utilize a gas-spring mechanism for torque generation (Kazerooni et al., 2019; Panero et al., 2021). As is typical of a gas spring, friction forces are added or subtracted to static forces generated respectively when being compressed (i.e., flexion phase) or extended (i.e., extension phase). The BackX™ (vs. Laevo™) had a larger difference in assistive torques between the flexion and extension phases, indicating that friction forces in the torque generation mechanism may be higher for the BackX™ (Figure 3). While the use of gas springs might be anticipated to yield velocity-dependent effects, it is possible that such effects are only observable outside the kinematic range tested here. Interestingly, some assistive torques were generated even when the torque-generating mechanism was not engaged (i.e., $BSE_{OFF}$), with peak values on the order of 5-10 Nm. These torques were higher for the Laevo™ vs. BackX™ (Figure 3.6 – Appendix D), and might have caused by different sources from the BSEs (e.g., mass, moment of inertia, strap tension, intrinsic resistance of the torque generation mechanism).

The assistive torque profiles in the static condition obtained for Laevo™ here were comparable with the results reported by Koopman et al. (2019). Specifically, the Laevo™ was estimated here to generate total torques (i.e., combined bilateral mechanisms) of up to ~18 Nm depending on the support setting. Similarly, the torques reported by Koopman et al. (2019) were mainly lower than 20 Nm at different bending angles (see Figure 4 in Koopman et al., 2019). For the dynamic condition, however, there were some inconsistencies between our results and Koopman et al. (2020b). Specifically, the Laevo™ was estimated here to generate total torques (i.e., combined bilateral mechanisms) of up to ~20 Nm depending on the support setting and bending direction, whereas torques reported by Koopman et al. (2020b) reached ~30 Nm within the same range of motion. Further, the Laevo™ torque profile here was an inverse parabolic shape with maximum torque at ~50–60º. This pattern was comparable to those of Koopman et al. (2020b) only for the high support setting. We believe this inconsistency stems from differences in methods used for defining the assistive torque of a BSE. Specifically, we derived assistive torques by subtracting angle-specific torques in the $BSE_{OFF}$ condition from each respective support condition ($BSE_{HIGH}$ and $BSE_{LOW}$), while the torques reported by Koopman et al. (2020b) appear to incorporate the combined $BSE_{OFF}$ and $BSE_{ON}$ torques. This combined effect is further evident from Figure 3.6 (Appendix D), which shows a similar pattern and torque magnitude compared to those of Koopman et al. (2020b); note, though, that torques presented here need to be multiplied by two for direct comparison. We believe the torques obtained from the $BSE_{OFF}$ condition incorporate the effects from the mechanical properties of the BSEs and the adapter (e.g., mass, moment of inertia, strap tension), and such effects were extracted from the torque measurements, given the study goals in the following chapter. Future research, however, is needed to develop a unified approach in measuring and presenting the assistive torques of exoskeletons.

Measuring supportive torque profiles while a human subject wears a BSE may provide more realistic torque profiles, by more properly reflecting relative motions between the BSE and the human user. With the current approach, though, we found it challenging to measure torque profiles using human subjects, especially during the dynamic condition. BSE torque profiles obtained using human subjects were comparable to those obtained using the mannequin over a narrow range of flexion/extension angles (~30º-60º; Figure 3.6 – Appendix D). In contrast, torque profiles in the static condition were quite comparable over the entire range of flexion/extension angles examined (Figure 3.6 – Appendix D). Given that peak RA muscle activity was rather small (suggesting minimal active torque generation), we posit that the noted challenges in measuring assistive torques using human subjects under a dynamic condition arise from some combination of several sources:

- When accelerating to reach a target flexion/extension angular velocity, or decelerating to stop, rotational moments due to the inertia of a participant's trunk and the trunk adapter occur in the opposite direction of movement. This effect mainly results because the continuous passive mode of the dynamometer was used, so that the trunk of a participant was passively moved by the dynamometer rather than the participant voluntarily flexing/extending the trunk. In addition, effects of wobbling masses in the human trunk can affect measured torques during angular acceleration and deceleration (e.g., Bazrgari et al., 2011). A more gradual angular acceleration/deceleration could help, though would still limit the angular range over which isokinetic data could be obtained.
- There is a limit on the feasible trunk range-of-motion using the TMC of the dynamometer when a participant is inside and wearing a BSE. Participants reported considerable discomfort beyond 90º of trunk flexion while in the TMC. Of note, the supine position used

for the mannequin allowed a large range-of-motion (0º-120º). Yet, our preliminary testing indicated that position still presented difficulties with human subjects, such as discomfort with hip flexed exceeded 90º –100º and lack of control over knee movements.

In summary, we captured torque profiles of BSEs using a computerized dynamometer in both static and dynamic conditions and with different BSE support settings. This approach permits control over joint kinematics and appears to be more effective using a mannequin vs. human subjects. Future work is recommended to assess this approach for other occupational exoskeletons, such as soft devices (exosuits) and for arm-support exoskeletons. Characterizing exoskeleton assistive torque profiles can help in better understanding how to select a torque profile for given task requirements and user anthropometry, and assist in predicting the potential impacts of exoskeleton use by incorporating measured torque profiles in a musculoskeletal modeling system.

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Appendix D

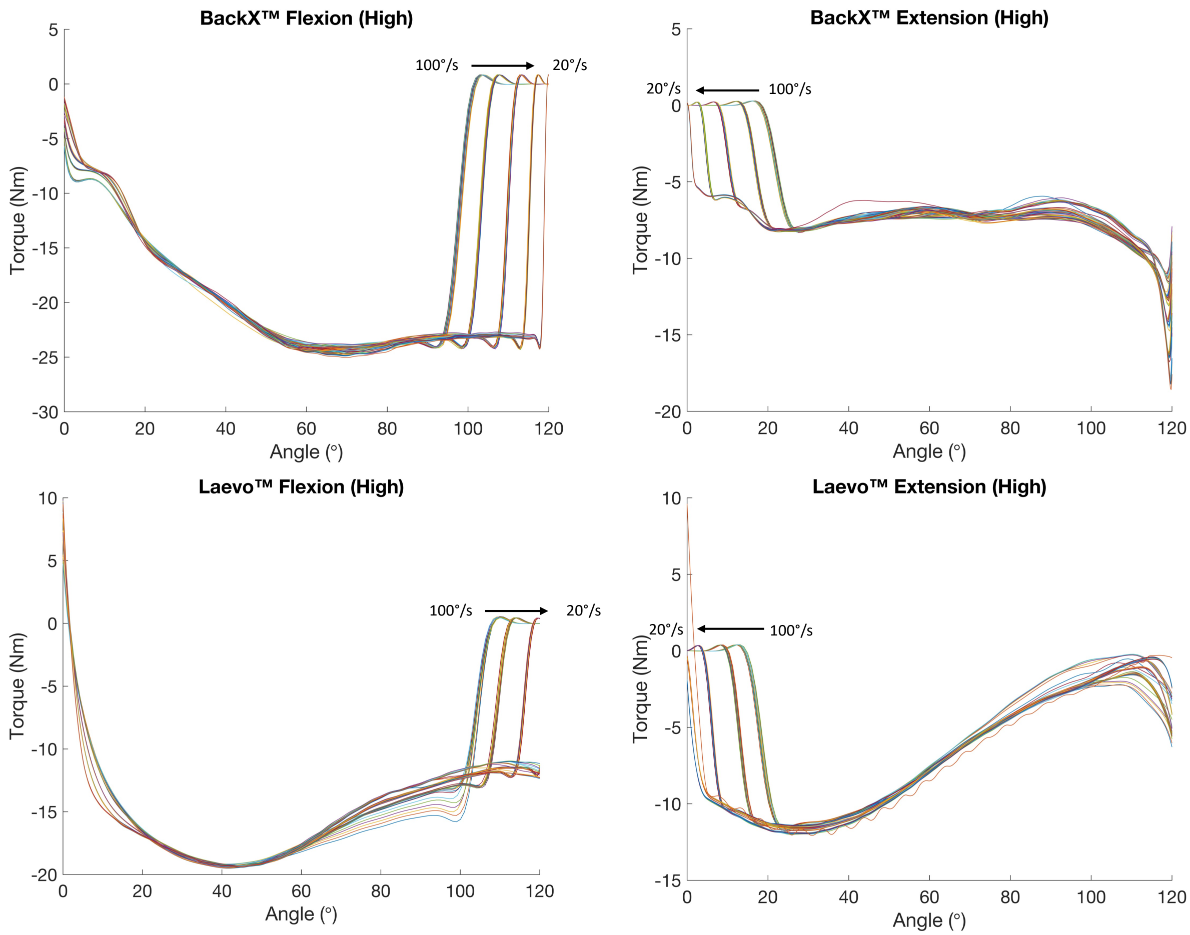


Figure 3.4. Examples of the effects of flexion/extension speed on torque profiles for the mannequin "wearing" the BackX™ and Laevo™ at high support levels. Angles indicate hip flexion for the mannequin. Illustrations at each of the five angular velocities include 14 replications. Torque profiles are similar across angular velocities, except for the beginning and ending phases of movement (likely due to the acceleration/deceleration of the hip adapter).

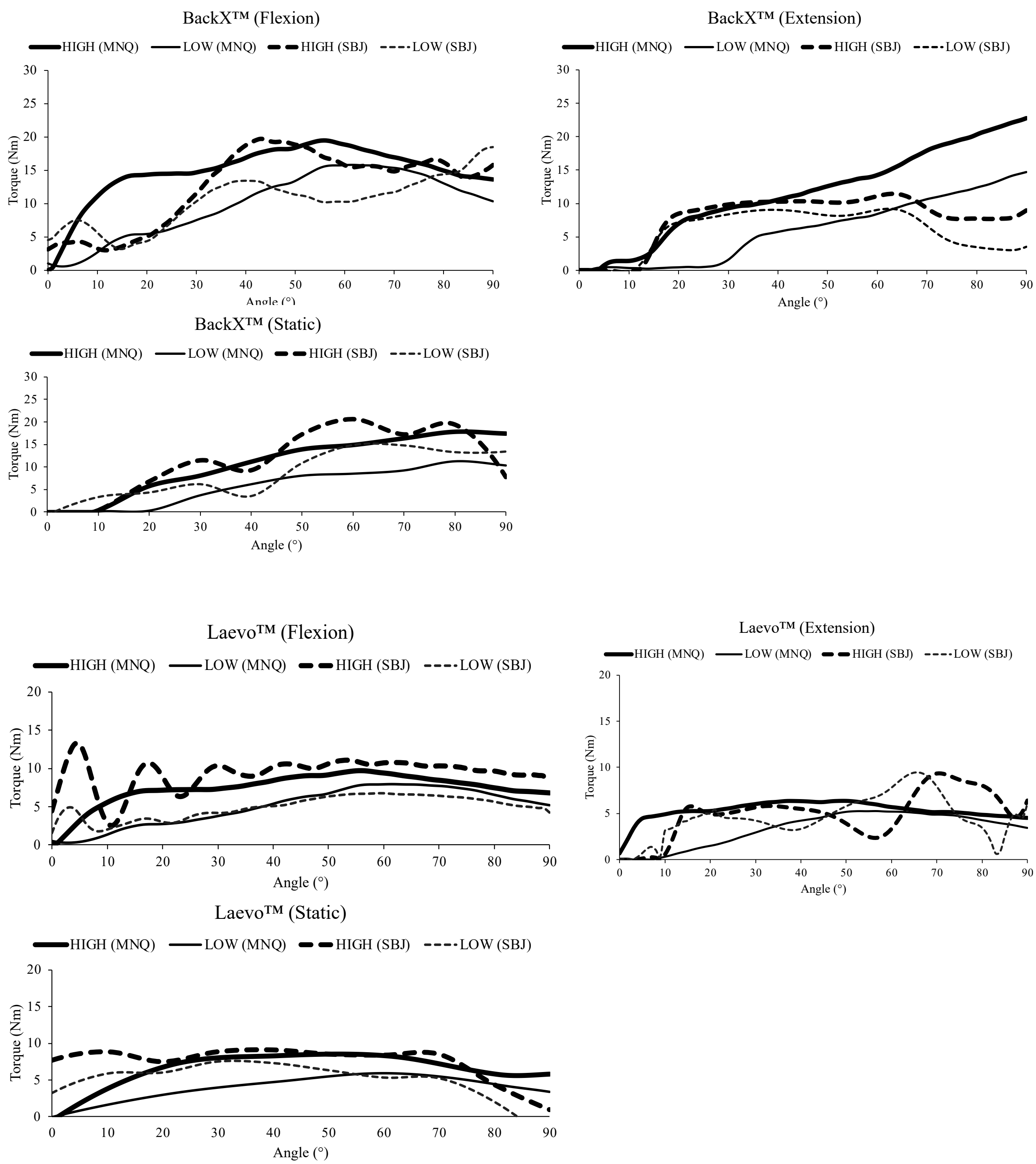


Figure 3.5. Torque profiles obtained from the mannequin (MNQ) and human subjects (SBJ) for BackX™ and Laevo™ in two support conditions (LOW and HIGH), separated by flexion (FLX) and extension (EXT) phases. Torque profiles are for each (unilateral) mechanism, with dynamic data shown for the 20º/sec condition.

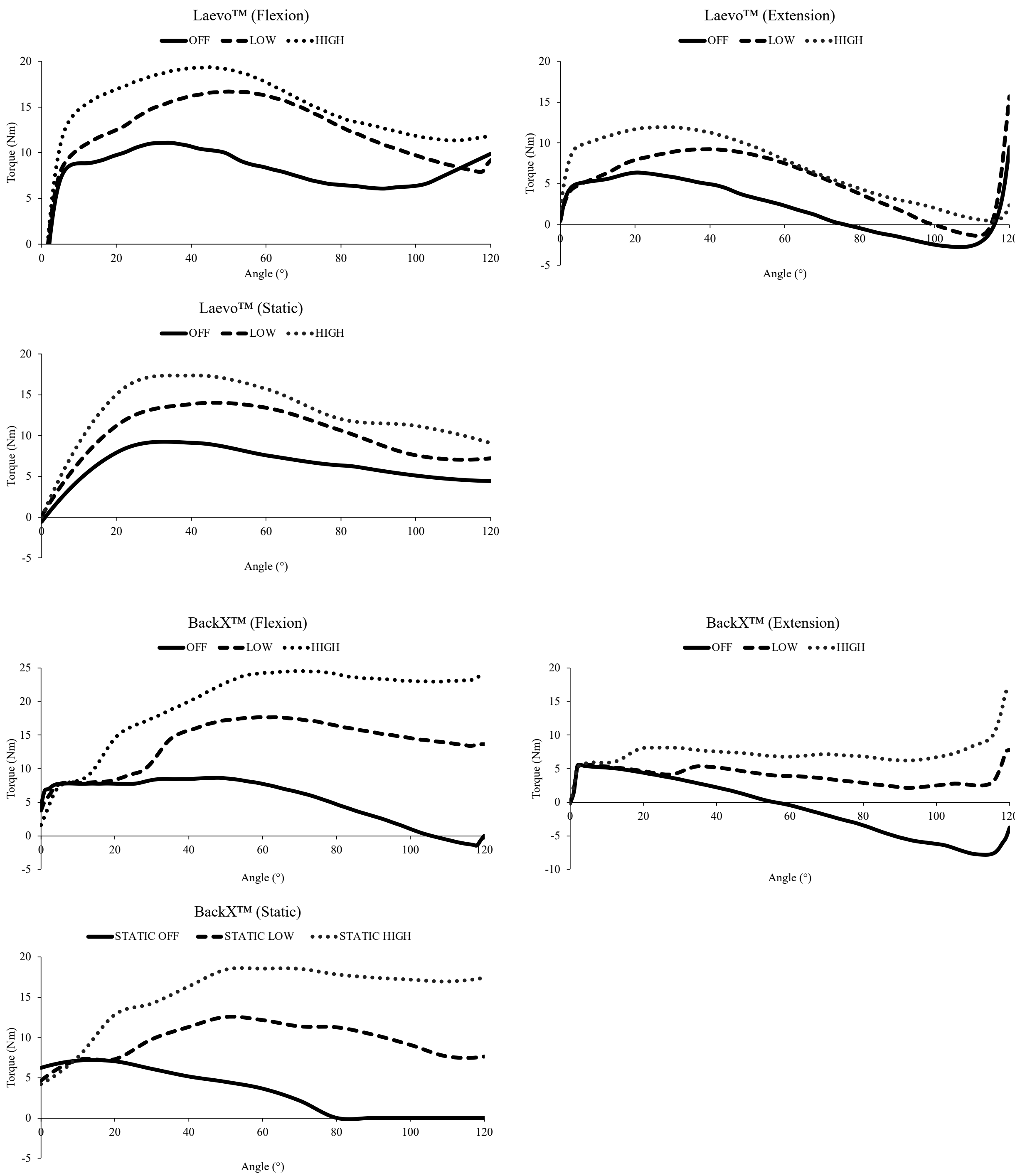


Figure 3.6. Demonstration of the torque profiles obtained from the mannequin for BackX™ and Laevo™ in three support conditions (OFF, LOW, and HIGH), shown separately for static testing and separately for the flexion and extension phases in dynamic testing. Torque profiles are for each (unilateral) mechanism, with dynamic data shown for the 20º/sec condition.

Table 3.1. Peak (95%ile) activation levels of the bilateral rectus abdominus muscle, separated by support level (high, low, off) and angular velocity (25, 50, 75, 100º/sec). Cell entries are means (SDs) and values are percent of MVICs.

| Speed (º/sec) | Support Setting | BackX™ | | Laevo™ | |
|---|---|---|---|---|---|
| | | Left | Right | Left | Right |
| 25 | Off | 6.3 (0.28) | 5.37 (2.48) | 6.12 (3.25) | 6.84 (1.7) |
| | Low | 5.72 (1.29) | 5.61 (3.37) | 6.54 (3.72) | 6.26 (3.73) |
| | High | 5.75 (0.79) | 5.14 (2.74) | 6.11 (2.9) | 6.04 (2.46) |
| 50 | Off | 5.65 (1.25) | 5.53 (3.12) | 5.94 (3.55) | 7.25 (0.84) |
| | Low | 5.27 (1.86) | 5.01 (3.39) | 6.32 (3.19) | 5.53 (3.47) |
| | High | 5.94 (0.95) | 5.16 (3.29) | 6.04 (3.02) | 6.42 (2.44) |
| 75 | Off | 5.74 (1.32) | 5.17 (2.55) | 6.32 (3.73) | 7.18 (1.86) |
| | Low | 5.09 (1.44) | 4.83 (3.5) | 6.24 (3.18) | 5.58 (3.59) |
| | High | 5.67 (1.45) | 4.92 (3.54) | 6.26 (3.02) | 6 (2.7) |
| 100 | Off | 5.44 (1) | 5.15 (2.65) | 5.92 (3.03) | 7.35 (2.3) |
| | Low | 5.37 (1.56) | 5.25 (3.92) | 6.46 (3.73) | 5.72 (3.63) |
| | High | 5.33 (1.08) | 4.69 (2.98) | 6.47 (3.52) | 6.21 (3.24) |

# 4 Chapter 4: Estimating Lumbar Spine Loading When Using Back-Support Exoskeletons in Lifting Tasks

## 4.1 Abstract

Low-back pain (LBP) continues as the leading cause of work-related musculoskeletal disorders, and the high LBP burden is attributed largely to physical risk factors prevalent in manual material handling tasks. Industrial back-support exoskeletons (BSEs) are a promising ergonomic intervention to help control/prevent exposures to such risk factors. While earlier research has demonstrated beneficial effects of BSEs in terms of reductions in superficial back muscle activity, limited evidence is available regarding the impacts of these devices on the spine loads. We evaluated the impacts of two passive BSEs (BackX™ AC and Laevo™ V2.5) on lumbosacral compressive and shear forces during repetitive lifting using an optimization-based model. Eighteen participants (gender-balanced) completed four minutes of repetitive lifting in nine different conditions, involving symmetric and asymmetric postures when using the BSEs (along with no BSE as a control condition). Using both BSEs reduced estimated peak compression and anteroposterior shear forces (by ~8-15%). Such reductions, however, were task-specific and depended on the BSE design. Laevo™ use also mediolateral shear forces during asymmetric lifting (by ~35%). We also found that reductions in composite measures of trunk muscle activity may not correspond with changes in spine forces when using a BSE. These results can help guide the proper selection and application of BSEs during repetitive lifting tasks. Future work is recommended to explore the viability of different biomechanical models to assess changes in spine mechanical loads due to BSE use, and whether reasonable estimates would be obtained using such models.

## 4.2 Introduction:

The U.S. workforce continues to experience high rates of work-related musculoskeletal disorders (WMSDs) due to overexertion and bodily reaction (BLS, 2019). Among all WMSDs, the back was most often affected, accounting for ~40% of such cases and resulting in a median of seven lost workdays (BLS, 2019). This high burden of low-back WMSDs is largely attributed to physical risk factors prevalent in manual material handling (MMH) tasks, including forceful exertions, repetitive lifting and bending, and sustained/prolonged non-neutral postures (Hoogendoorn et al., 2000; da Costa & Vieira, 2010). To help control/prevent exposures to these physical risk factors, back-support exoskeletons (BSEs) have been introduced as a new ergonomic intervention to reduce physical demands on the spine (De Looze et al., 2016). Passive BSEs, requiring no actuators or power supply, are of specific interest, due to their cost-efficiency, ease of implementation, and predominance in the commercial market.

While existing evidence supports the potential for passive BSEs to reduce low-back demands and associated injury risks, many earlier studies examined the beneficial effects of BSEs only in terms of reductions in superficial back muscle activity (e.g., Lamers et al., 2018; Näf et al., 2018; Alemi et al., 2019; Madinei et al., 2020a). Loads on the human spine, however, are influenced not only by the activity of these superficial muscles, but also by deep trunk muscles and passive tissues (Bazrgari & Shirazi-Adl, 2007), as well as segmental kinematics and external loads (i.e., due to gravity and inertia). Consequently, and especially if lifting behaviors change when using a BSE, reductions in back muscle activity, via electromyography (EMG), may not necessarily imply a reduction in spine loading.

A few studies investigated the impact of BSE use on spine compression forces during static holding and dynamic lifting tasks, using EMG-assisted biomechanical models. Using one BSE

(Laevo™) significantly reduced spine compression forces, by 5-10% when lifting a 10-kg box from a far distance (60 cm) (Koopman et al., 2020b). Using another BSE (SPEXOR) also significantly reduced spinal compression forces, by 14% during lifting and by 13-21% during bending tasks (Koopman et al., 2020a). In a different study, the effects of using the Laevo™, Robo-Mate™, and SPEXOR BSEs on spine compression forces were tested by Kingma et al. (2020), during repetitive lifting of boxes with masses of 10, 15, and 10 kg, respectively. They found minor changes using the first exoskeleton, and 17% and 14% reductions were found for the second and third BSEs. However, the use of detailed biomechanical models in such evaluations can be challenging, not only due to the additional experimental effort required (e.g., for model calibration and EMG normalization), but also because of the interference of BSE structures with the EMG sensors that can compromise data quality. A recent study also examined the efficacy of using a newly introduced passive BSE (Paexo™, Ottobock) during a repetitive lifting/lowering of a 10-kg box from ground onto a table using an optimization-based approach in the AnyBody™ Modeling System (Schmalz et al., 2021). They found that Paexo™ reduced peak L5/S1 compression forces by 20%.

We reported earlier on a detailed assessment of two passive BSEs during symmetric and asymmetric repetitive lifting with moderate postural demands (Madinei et al., 2020a), finding 9-20% reductions in trunk extensor muscle activities, and with minimal changes in lifting behavior observed, when using either BSE. It is unknown, however, if such reductions in muscle activity correspond to similar changes in spine compression forces. To assess this correspondence, we used an optimization-based musculoskeletal model in the AnyBody Modeling System (AMS) to estimate the mechanical loads on the lumbar spine when using the BSEs during the simulated lifting tasks. In contrast to the studies cited earlier, an EMG-assisted model was not used here.

With the BSEs tested, we could not easily measure the activity of the muscles of interest, since the structure and padding of the BSEs interfered with surface EMG sensors and could compromise the quality of the EMG data. We hypothesized that using both BSEs would reduce spine loads (i.e., lumbosacral compression and shear), though to differing extents across lifting conditions. We also expected that magnitude of such reductions would not necessarily or consistently correspond with measured reductions in superficial back muscle activity, given that the latter has only a partial role contributing to resultant spine loads in some postures.

## 4.3 Methods

### 4.3.1 Experimental design and procedures:

We completed a secondary analysis of data obtained from a study described earlier (Madinei et al., 2020a). In that study, participants (9 Females and 9 Males) performed free-style repetitive lifting in several combinations of *Intervention* and *Lifting Condition*. *Intervention* included three levels: a control (unassisted) condition and two BSEs (BackX™ AC and Laevo™ V2.5). The three *Lifting Conditions* included symmetric lowering/lifting to/from mid-shank (Sym_Ground) and knee level (Sym_Knee), and asymmetric lowering/lifting to/from knee height located 90º to the right of the mid-sagittal plane (Asy_Knee). During a training session, participants selected their preferred support settings for each BSE and lifting condition. All lowering/lifting conditions were initiated at the standing posture with the box set at individual waist height (i.e., anterior superior iliac spine). Participants completed trials of repetitive lowering/lifting using a wooden box, the mass of which was set to 10% of individual body mass. Each lifting condition lasted for 4 minutes, at a pace of 10 lower/lift cycles per minute, for a total of 40 cycles in each. During the trials, whole-body segmental kinematics were monitored using an inertial motion capture system (MVN Awinda, Xsens Technologies B.V., the Netherlands). Surface electromyography (EMG) was also obtained bilaterally – from the thoracic erector spinae (TES), iliocostalis lumborum (ILL), rectus abdominis (RA), and external oblique (EO) – and subsequently normalized to peak values measured in maximum voluntary isometric exertions. In the current analysis, we set up and applied a biomechanical model to estimate internal spine loads (compression and shear) at the L5/S1 level.

### 4.3.2 Model setup:

An anatomically-detailed spine model in the AnyBody™ Modeling System (AMS: Version 7.3, AnyBody Technology, Aalborg, Denmark) was employed, using the AnyBody Modeling Repository (AMMR V.2.3.3). The spine model consisted of seven rigid segments (pelvis, five lumbar vertebrae, and thorax) with six ball-and-socket lumbar and thoracolumbar joints. This model accounts for nonlinear passive properties of the ligamentous spine, dynamic characteristics of the trunk (i.e., mass, mass moment of inertia, and damping), detailed muscle architecture, wrapping of the global extensor muscles, and satisfaction of equilibrium at all spinal levels and directions (Hansen et al., 2006; De Zee et al., 2007; Han et al., 2012). Several studies have assessed the ability of the AMS spine model to estimate spine loads both during static holding/bending (Rasmussen et al., 2009; Rajaee et al., 2015) and dynamic lifting tasks (Bassani et al., 2017; Larsen et al., 2020). Performance of the model was supported by high correlations between model-based estimates and gold-standard *in vivo* measurements (Wilke et al., 2001) of intradiscal pressure over a moderate range of trunk bending and twisting.

Full body kinematics data collected in Chapter 2 were exported from Xsens MVN Studio as .bvh files and imported into the AMS, in which a linked-segment model with 44 degrees-of-freedom was reconstructed (Larsen et al., 2020). To match the linked-segment model with the musculoskeletal model, we used an established approach involving virtual markers, minimizing linear distances between the virtual markers and the musculoskeletal model (Andersen et al., 2010; Skals et al., 2017b; Karatsidis et al., 2018). Ground reaction forces were predicted using a method evaluated in several previous studies (Fluit et al., 2014; Skals et al., 2017a; Karatsidis et al., 2019). In brief, 25 dynamic contact elements were attached under each foot. Each contact element consisted of five uniaxial force actuators to generate a positive normal force, as well as positive and negative anteroposterior and mediolateral static friction forces. Individual actuation of each contact force actuator was then computed as part of the muscle recruitment problem.

Muscle activations and internal reaction forces were computed through an inverse dynamic analysis, by minimizing the sum of quadratic muscle stress to solve for the problem of kinetic redundancy (Rasmussen et al., 2001; Damsgaard et al., 2006). Note that participant body mass, stature, and other anthropometric measures (e.g., trunk length, shoulder width, hip width, arm span) were used to scale body segment masses and lengths in the AMS. A mechanical interface for each BSE was created in the AMS platform, consisting of a torso frame hinged to two leg frames at the hip joint (Figure 4.1). The length of the chest plate was adjusted based on each participant's xiphoid process and the length of the leg frames was adjusted based on the exoskeleton dimensions. Kinematics of the torso and leg frames were constrained so that they followed the participant's torso and hip motions, respectively.

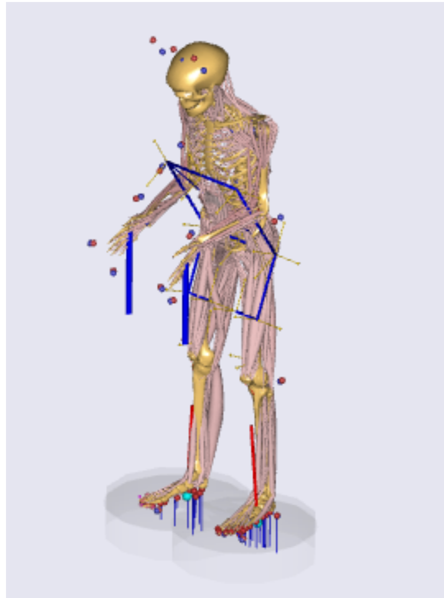

Figure 4.1. Illustration of the BSE interface in the AnyBody™ Modeling System, consisting of a torso frame hinged to two leg frames at the hip joint. External hand forces are indicated by vertical blue lines applied to the palm joints.

### 4.3.3 BSE torque profiles

The mechanical behavior of each BSE (i.e., torque vs. angle relationship) was measured using a computerized isokinetic dynamometer (Humac Norm, CSMi, MA). More details of the procedures are presented elsewhere (Chapter 3). Briefly, a flexible mannequin with hinge joints at the hips was used to mimic hip flexion/extension. Each BSE was fitted to the mannequin, the mannequin was positioned and secured to the dynamometer bed, and the left thigh of the mannequin was attached to the hip adapter of the dynamometer. This adapter was then programmed in the continuous passive motion (CPM) mode while setting the BSE support at two levels (instant low and instant high conditions for the BackX™ and low-cam and high-cam conditions for the Laevo™). Hip torque was recorded via the dynamometer software (HUMAC 2016, v.10.7.1) at 100 Hz. BSE torque profiles were derived as a function of angular position and velocity to model the assistive torque provided by the BSEs in the AMS.

### 4.3.4 Data processing and outcome measures

Given that processing all 40 cycles of each condition was excessively time-consuming, we focused our analysis on the 10th and 30th lowering/lifting cycles in each condition. Three-dimensional reaction forces at the L5/S1 intervertebral joint were extracted from the musculoskeletal model, specifically axial compression ($F_{COMP}$), anteroposterior shear ($F_{AP}$), and mediolateral shear ($F_{ML}$). Given symmetry, all $F_{ML}$ data were converted to absolute values. Peak (95th percentile) values were derived as outcome measures. Similar to our earlier work (Madinei et al., 2020a), outcome measures were obtained separately in the lowering and lifting phases.

### 4.3.5 Statistical Analyses

Separate three-way, mixed-factor analyses of variance (ANOVAs) were used to assess the effects of *Intervention*, *Lifting Condition*, and *Gender* on model-estimate peak spine forces. We initially explored whether there were any differences in outcomes between the two noted cycles. Since none were apparent ($p$ values ~0.8-0.9), we removed cycle from the final ANOVA models. Presentation orders of *Task Condition* and *Intervention* were included as blocking factors in the models. Parametric model assumptions were assessed (e.g., normality of residuals and equality of variances) and dependent variables violating these assumptions were transformed using deterministic mathematical functions until all assumptions were satisfied. Statistical significance was concluded when $p < 0.05$, and effect sizes were estimated using eta-squared ($\eta^2$). Significant interaction effects were explored using simple-effects testing, and *post hoc* paired comparisons were completed using the Tukey-Kramer procedure where relevant. Given the study goals, the subsequent presentation of results and the discussion emphasizes the main and interaction effects of *Intervention*.

We further explored the feasibility of using measures of muscle activity to estimate changes in simulated spinal loads when using the BSEs. For this, we first obtained two composite metrics of composite muscle activity as reported in our earlier work (Madinei et al., 2020a). The first metric captured activity of the monitored trunk extensor muscles (TEM), calculated as the sum of the peak (95th percentile) levels of the bilateral iliocostalis lumborum and thoracis erector spinae muscles. Total trunk muscle (TTM) activity was the second metric, and was obtained as the sum of normalized muscle activity levels for the bilateral erector spinae, iliocostalis lumborum, rectus abdominus, and external oblique. We then derived relative changes in these metrics, along with changes in the estimated peak (95th percentile) spine loads, by comparing changes resulting from BSE use relative to the control (no EXO) condition. We initially explored the linear correlations between the metrics of muscle activity and the simulated forces across participants, and separately for each of the 12 combinations of *Intervention*, *Task Condition*, and *Gender*. In a subsequent analysis, we collapsed the data by averaging across participants in each of the 6 *Intervention* x *Task Condition* combinations, then examining correlations across the 6 combinations. Pearson correlation coefficients ($r$) were reported as outcome measures.

## 4.4 Results

### 4.4.1 Spine Reaction Forces

A summary of ANOVA results for each outcome measure is provided in Table 4.2 (Appendix E). *Intervention* main effects were significant for $F_{COMP}$ and $F_{AP}$. Using the BackX™ significantly reduced both of these reaction forces, respectively by 13.3 and 15.1% during the lowering phase, and by 8.2 and 8.5% during the lifting phase (Figure 4.2). Laevo™ use also led to significant reductions in $F_{COMP}$ and $F_{AP}$, respectively by 9.5 and 10.6%, but only during the lowering phase (Figure 4.2).

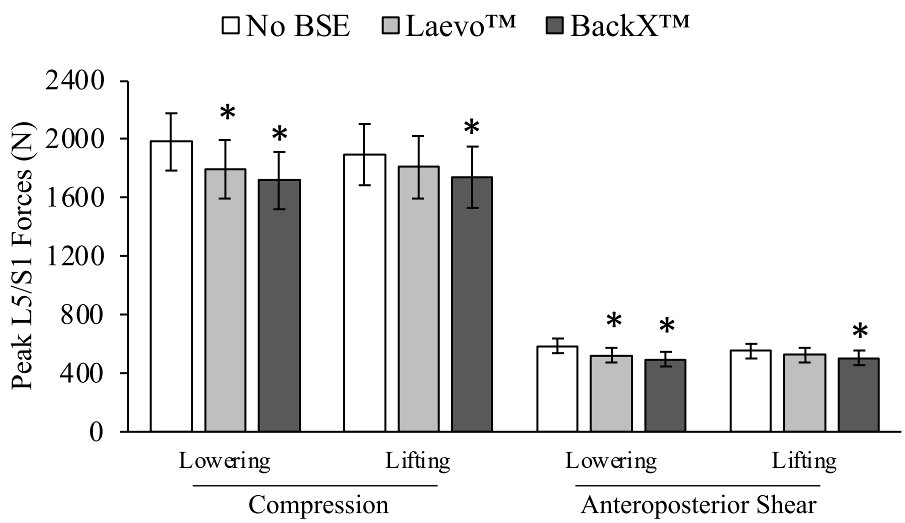


Figure 4.2. *Intervention* effects on peak compression and anteroposterior shear forces at the L5/S1 intervertebral joint. Note that * denotes significant differences from the control condition (i.e., no BSE), and error bars indicate 95% confidence intervals.

There was a significant *Intervention* × *Task Condition* effect on $F_{ML}$ during both the lowering and lifting phases. Specifically, *Task Condition* was significant for all three *Interventions* ($p < 0.0001$), but the effect of *Intervention* was only significant for Asy_Knee ($p < 0.0001$). $F_{ML}$ decreased significantly when using the Laevo™ in the Asy_Knee condition, by 34.5% during the lowering phase and by 30.5% during the lifting phase (Figure 4.3).

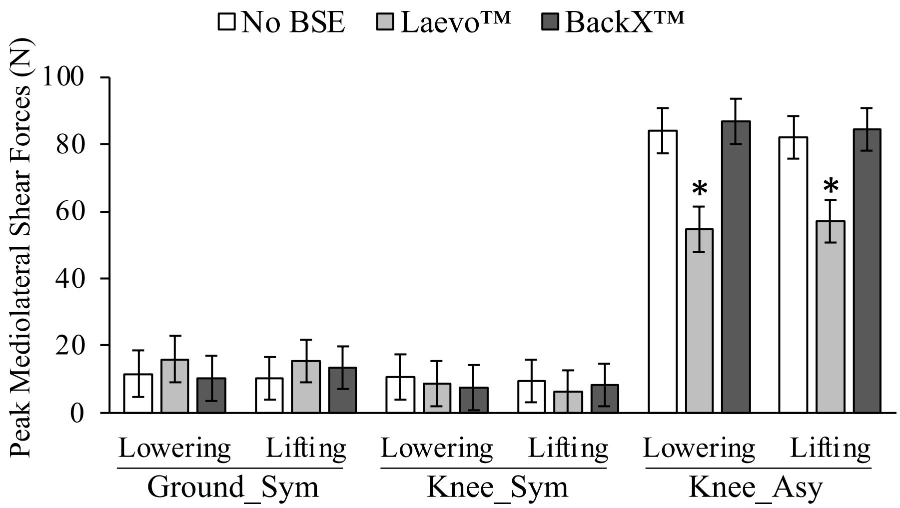


Figure 4.3. *Intervention* × *Task Condition* interaction effects on the peak mediolateral shear forces at the L5/S1 intervertebral joint. Note that * denotes significant differences from the control condition (i.e., no BSE), and error bars indicate 95% confidence intervals.

### 4.4.2 Correlation coefficients

When analyzed separately in each combination of *Intervention* and *Task Condition*, most correlation coefficients were small ($|r|< \sim 0.3$) and not statistically significant (Table 4.3 - Appendix E). Furthermore, visual inspection did not reveal any clear or consistent relationships between the predicted spine forces and metrics of muscle activity. However, after averaging across participants, the correlations were generally higher (Table 4.1). Strong correlations ($r > \sim 0.8$) were found for $F_{COMP}$ among females, which were statistically significant during both the lowering and lifting phases. Moderate correlations ($r > \sim 0.6$) were found among females for $F_{AP}$, although this was only significant during the lowering phase. Small-to-moderate correlations ($\sim 0.2 < |r| < \sim 0.5$) were obtained for males for both $F_{COMP}$ and $F_{AP}$, although none were statistically significant. More details of the outcomes of these correlation analyses are provided in Figure 4.4 (Appendix E).

Table 4.1. Correlation coefficients (r values) between relative changes in simulated spinal loads and the corresponding changes in composite metrics of muscle activities (i.e., TEM= trunk extensor muscles; TTM= total trunk muscles). Bold values indicate statistically significant values.

| | | Composite measures | | | |
|---|---|---|---|---|---|
| Gender | Estimated Forces | TEM | | TTM | |
| | | Lowering | Lifting | Lowering | Lifting |
| Female | $F_{COMP}$ | **0.900** | **0.784** | **0.887** | **0.782** |
| | $F_{AP}$ | **0.897** | 0.589 | **0.878** | 0.601 |
| | $F_{ML}$ | -0.184 | -0.486 | -0.163 | -0.496 |
| Male | $F_{COMP}$ | 0.346 | -0.508 | 0.437 | -0.470 |
| | $F_{AP}$ | 0.456 | -0.225 | 0.546 | -0.181 |
| | $F_{ML}$ | -0.251 | 0.090 | -0.407 | -0.006 |

## 4.5 Discussion

### 4.5.1 Spine reaction forces

We hypothesized that spine loads would be reduced in some lifting conditions using two different BSEs. Estimates obtained from an optimization-based model supported this hypothesis overall, but also indicated that the magnitudes of such reductions were task-specific and dependent on the BSE design. Specifically, BackX™ use led to ~8-13% reductions in $F_{COMP}$ and ~8-15% reductions in $F_{AP}$ depending on the lifting phase. Using Laevo™ also resulted in ~10% reductions in both $F_{COMP}$ and $F_{AP}$ only during the lowering phase. Laevo™ use further reduced $F_{ML}$ in the Asy_Knee condition, by 34.5% during the lowering phase and by 30.5% during the lifting phase.

The larger reductions in spine compression and shear forces observed with BackX™ vs Laevo™ likely resulted from the higher support settings provided by the BackX™ (see Chapter 4 for more details). Specifically, BackX™ provides torques up to 14.7 Nm and 24.8 Nm for the low and high support settings during flexion, and 11.3 Nm and 19.2 Nm for the respective settings during extension. In comparison, respective values for the Laevo™ are 7.9 Nm and 9.7 Nm during flexion, and 5.3 and 6.4 Nm during extension. Further, participants often preferred higher support settings when using the BackX, while it was not the case for the Laevo™. These estimates of spine loads are consistent overall with our earlier findings (Madinei et al., 2020a), wherein we observed larger reductions in trunk extensor muscle activity and total trunk muscle activity when using the BackX™. Our current results also agree with recent studies of using the Laevo™ and Paexo™ during repetitive lifting tasks, which respectively found up to 10 and 20% reductions in spine compression forces (Kingma et al., 2020; Koopman et al., 2020b; Schmalz et al., 2021).

We found larger reductions in spine compression and shear forces during the lowering vs. the lifting phase for either BSE. This outcome is consistent with our earlier report of larger reductions in trunk extensor muscle activity and the sum of trunk muscle activity during lowering vs. lifting (Madinei et al., 2020a). We suggest that the increased benefit of both BSEs while lowering is due to the hysteresis present in the torque generation mechanisms of both BSEs. As shown in Chapter 4, torque outputs generated by both BSEs are greater during trunk flexion than extension (up to 24.8 vs. 19.2 Nm for BackX™; up to 9.7 vs. 6.4 Nm for Laevo™). Koopman et al., 2019 reported a similar effect, in that the assistive torque generated by the Laevo™ was found to be considerably higher during the lowering phase compared to lifting.

We found a significant decrease in $F_{ML}$ during asymmetric lifting when using the Laevo™, which might have resulted from kinematic changes caused by using this BSE. Our earlier findings indicated that Laevo™ use decreased lumbar axial range-of-motion (ROM), by up to ~23%. It also decreased axial angular velocity, by up to ~14% compared to using the BackX™ and ~12% with no BSE (Madinei et al., 2020a). Given that changes in trunk posture and movement speed are directly associated with the mechanical loads on the spine (Davis & Marras, 2000; Lavender et al., 2003), the reductions in trunk axial ROM and velocity likely contributed to the decrease in $F_{ML}$ during asymmetric lifting.

### 4.5.2 Can BSE-induced changes in spine loads be predicted from EMG?

As noted above, we did not use an EMG-based modeling approach to estimate spine loads, since collecting surface EMG was not feasible for some muscles. We thus explored the feasibility of predicting *changes* in simulated spinal loads when using a BSE by *changes* in composite measures of muscle activity. These results build on earlier evidence that simple measures of muscle activity can be used to indirectly estimate spinal loads. For example, Potvin et al. (1990) showed the efficacy of EMG recordings from both thoracic and lumbar portions of the erector spinae muscles in estimating the dynamic compressive forces on the lumbar spine. Notably, Graham et al. (2009) used the same method to assess low back demands when using a wearable assistive device (the "PLAD") during a repetitive lifting task. A similar approach was adopted by Mientjes et al. (1999), who found moderate-to-high correlations between normalized trunk muscle activity and spine compression forces, but only for tasks involving minor axial moments.

Although there were no significant correlations between composite measures of muscle activity and changes in spine forces at the individual level, noticeable correlations were found at the group level. Specifically changes in back muscle activity corresponded reasonably well with the relative changes in spine compression and anteroposterior shear forces for female participants. Female participants experienced larger reductions in trunk muscle activity when using either BSE compared to males (by 22.4% vs. 16.5% with BackX™; and by 11.5% vs. no change with Laevo™). This gender-related effect, however, was not evident in the estimates of the spine compression and anteroposterior shear forces. These results confirm our initial expectation that changes in spine forces may not necessarily correspond with the reductions in metrics of back muscle activity, and the negative results are perhaps not surprising given that the latter has only a partial role contributing to resultant spine loads in some postures. Practically, these results imply that composite measures of muscle activity may not have consistent utility in predicting changes in spine forces resulting from BSE use, and thus, invalid conclusions could be reached if investigators rely only on surface EMG measures to assess the impacts of BSEs on spine loads. Further investigation, however, is needed to compare the composite measures of muscle activity with other biomechanical models (e.g., EMG-assisted or finite element models).

It is worth noting that comparisons of estimated muscle forces/activities with the empirical measures obtained from Chapter 2 was not possible due to lack of one-to-one correspondence of the muscles in each approach. For example, the extensor muscles in AnyBody™ consisted of tens of distinct fascicles, while the muscle activities recorded in Study 1 might involve crosstalk from multiple adjacent muscles, making it challenging to correlate the model-based output values (muscle activity and/or force) with EMG recordings.

## 4.6 Limitations

Some limitations of the current study need to be noted. First, while the anatomical fidelity of the AMS model has been established in earlier work (Hansen et al., 2006; De Zee et al., 2007; Han et al., 2012) the model does rely upon a number of assumptions (e.g., rigid rib cage and thoracic spine, lumbar discs treated as spherical joints) that may have affected estimated of spine loads. Nevertheless, the effects of such assumptions on predicted loading of lower lumbar spine has been suggested to be minimal (Ignasiak et al., 2016). Second, reconstructing motion data from

IMUs may introduce some errors, such as from soft-tissue artifact and discrepancies between the Xsens linked-segment model and the AMS musculoskeletal model (Damsgaard et al., 2006; De Zee et al., 2007). The magnitude of such errors, however, is expected to be <6º when compared to optical motion capture systems (Karatsidis et al., 2018). Furthermore, these errors should not affect our comparisons between conditions since the errors are not likely to vary substantially between conditions. Third, we neglected any effect of BSE mass in the inverse dynamic analyses, as it was unclear how the mass (< 4.5 kg) was distributed over the body. Given that a substantial portion of the BSE mass is carried by the pelvis, though, any effect on spine loading is likely limited. Fourth, forces at the hand-box coupling were not measured, and these forces were instead modelled with additional contact elements. Measurements of these forces, however, are more relevant to improve the accuracy of the kinetic computations *above* the thorax, such as for shoulder and elbow joint reaction forces (Larsen et al., 2020).

## 4.7 Conclusions

We evaluated the impacts of two passive BSEs on lumbosacral compressive and shear forces during symmetric and asymmetric repetitive lifting using an optimization-based model. Using both BSEs reduced peak compression and anteroposterior shear forces (by ~8-15%). Such reductions, however, were task-specific and depended on the BSE design. Laevo™ use also reduced mediolateral shear forces during asymmetric lifting (by ~35%). Our findings further suggest that composite measures of muscle activity may not have consistent utility in predicting changes in spine forces resulting from BSE use. These results can help guide the proper selection and application of BSEs during repetitive lifting tasks. Future work is recommended to explore the feasibility of other biomechanical models to quantify changes in mechanical loads on the spine caused by using a BSE, and if reasonable estimates would be obtained using such models.

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## Appendix E

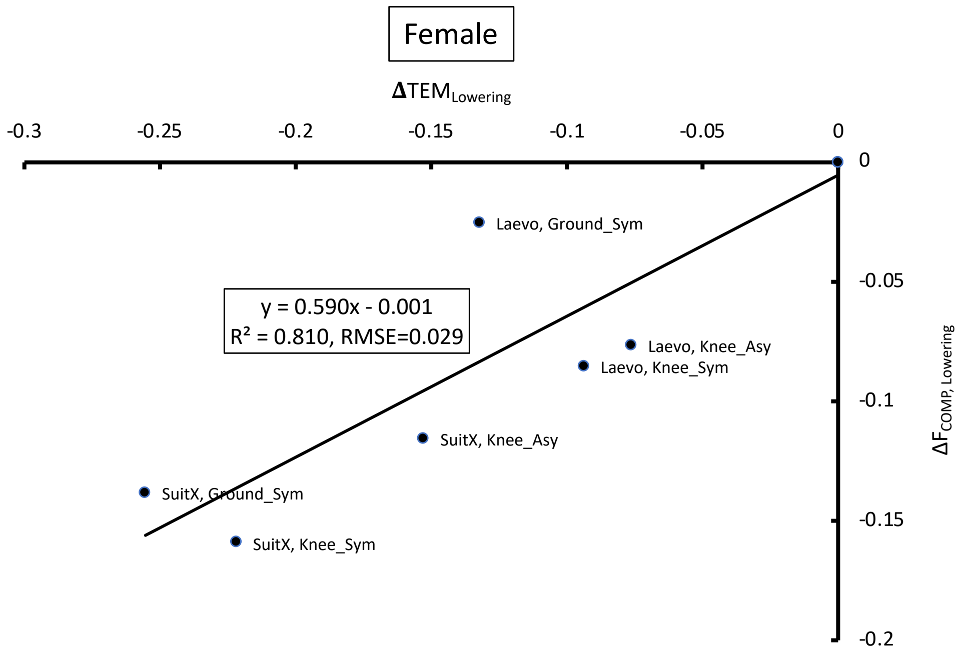
Table 4.2. Summary of ANOVA results regarding the main and interaction effects of Gender, *Intervention*, and *Task Condition* on $F_{COMP}$, $F_{AP}$, and $F_{ML}$ during the lowering and lifting phases. Each cell provides the *F* value, along with the associated *p* value and $\eta^2$ in parentheses. Note that significant effects are highlighted in bold font.

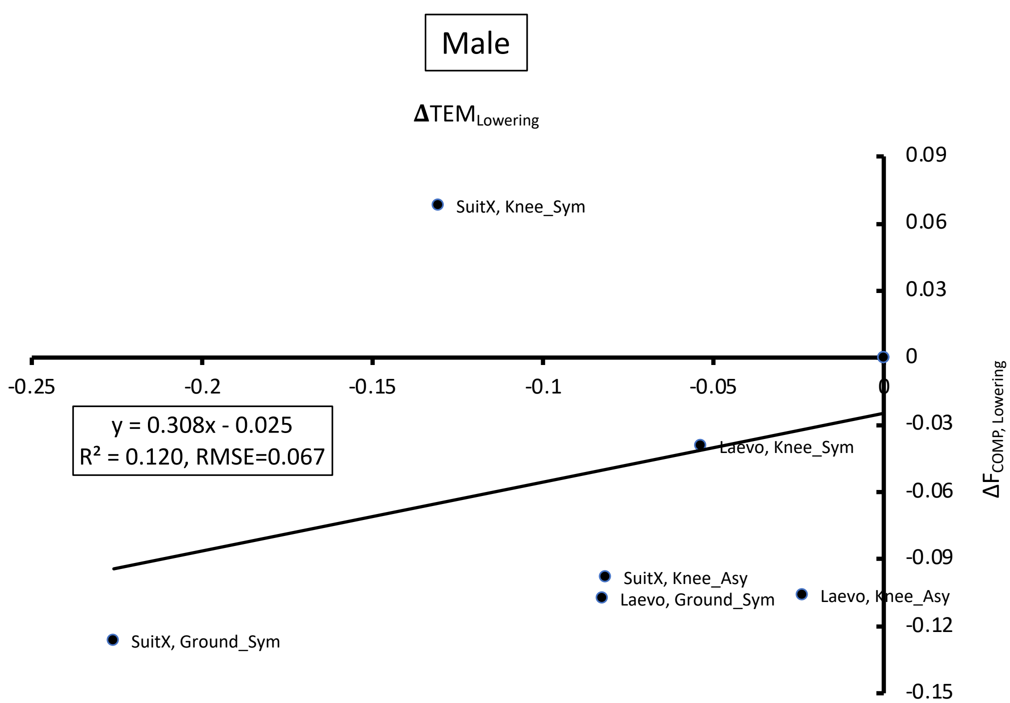
| Muscle | Phase | Transform | Gender (G) | Intervention (I) | Task Condition (TC) | G × I | I × TC | G × TC | G × I × TC |
|---|---|---|---|---|---|---|---|---|---|
| $F_{COMP}$ | Lower | | 1.28 (0.274, 0.12) | **19.62 (<.0001, 0.139)** | **10.75 (<.0001, 0.076)** | 1.14 (0.321, 0.008) | 0.82 (0.511, 0.012) | **5.72 (0.004, 0.041)** | 1.05 (0.381, 0.015) |
| | Lift | | 0.79 (0.388, 0.03) | **7.98 (0.0004, 0.016)** | **11.52 (<.0001, 0.023)** | 0.66 (0.517, 0.001) | 0.17 (0.951, 0.001) | **6.33 (0.002, 0.013)** | 0.33 (0.858, 0.001) |
| $F_{AP}$ | Lower | | 0.58 (0.457, 0.039) | **23.85 (<.0001, 0.169)** | **5.64 (0.004, 0.04)** | 0.56 (0.573, 0.004) | 0.99 (0.414, 0.014) | **4.4 (0.013, 0.031)** | 0.97 (0.425, 0.014) |
| | Lift | | 0.45 (0.514, 0.008) | **7.9 (0.001, 0.004)** | **6.96 (0.001, 0.016)** | 0.32 (0.727, 0.004) | 0.62 (0.647, 0.0004) | **6.36 (0.002, 0.001)** | 0.46 (0.764, 0.003) |
| $F_{ML}$ | Lower | Sqrt | 0.01 (0.906, 0.0001) | **3.44 (0.034, 0.005)** | **428.25 (<.0001, 0.665)** | 0.28 (0.756, 0.0004) | **8.21 (<.0001, 0.025)** | **5.62 (0.004, 0.009)** | 1.54 (0.191, 0.005) |
| | Lift | Sqrt | 0.08 (0.786, 0.0003) | **4.86 (0.008, 0.006)** | **541.08 (<.0001, 0.721)** | 0.13 (0.878, 0.0002) | **6.44 (<.0001, 0.017)** | **4.73 (0.01, 0.006)** | 0.85 0.497, 0.002) |

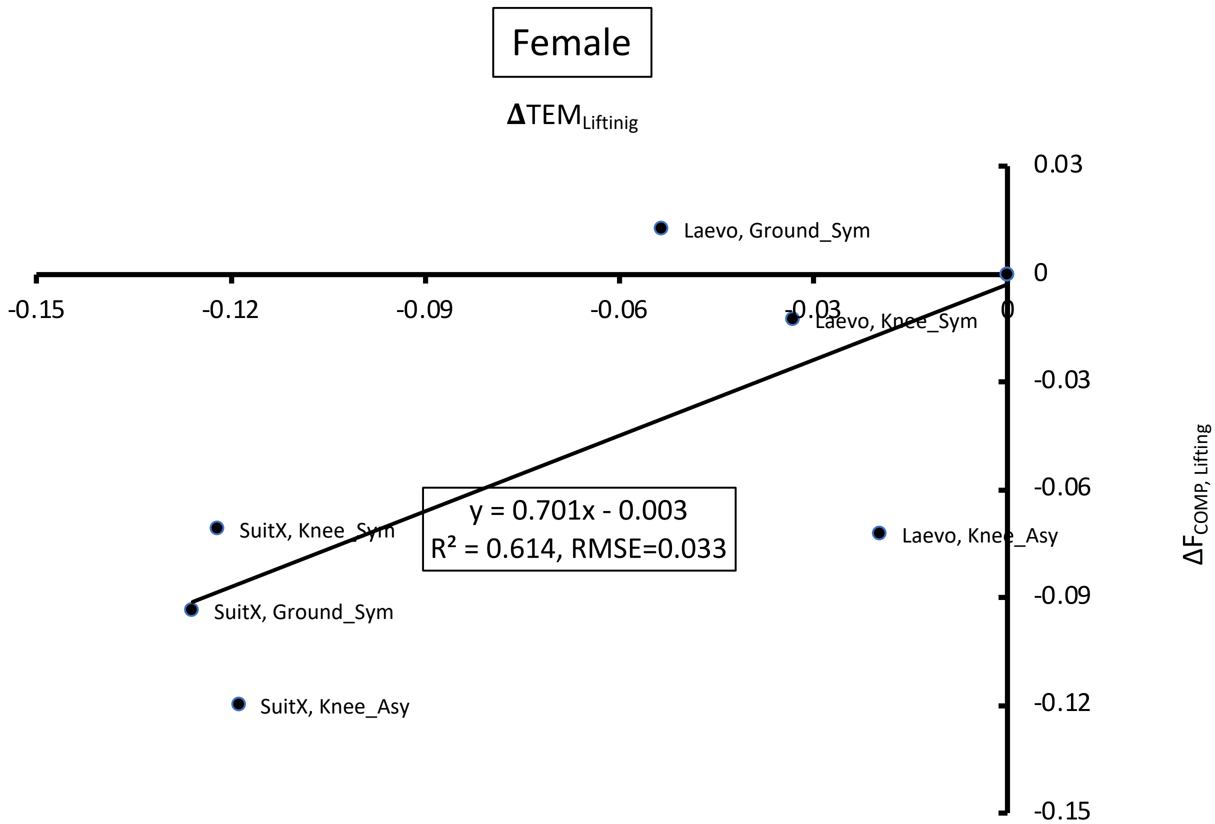
Table 4.3. Correlation coefficients (r values) between relative changes in simulated spinal forces and the corresponding changes in empirical metrics of trunk muscle activity, by Intervention and Task Condition. Bold values indicate statistically significant correlation coefficients.

| | Intervention | Task Condition | Simulated Forces | Empirical measures: TEM Lowering | TEM Lifting | TTM Lowering | TTM Lifting |
|---|---|---|---|---|---|---|---|
| Female | BackX | Sym_Ground | $F_{COMP}$ | -0.260 | -0.320 | -0.308 | -0.408 |
| | | | $F_{AP}$ | -0.211 | -0.101 | -0.231 | -0.221 |
| | | | $F_{ML}$ | 0.095 | 0.442 | 0.024 | 0.225 |
| | | Sym_Knee | $F_{COMP}$ | -0.204 | 0.064 | -0.181 | -0.051 |
| | | | $F_{AP}$ | -0.293 | 0.141 | -0.216 | 0.011 |
| | | | $F_{ML}$ | -0.437 | 0.345 | -0.448 | 0.254 |
| | | Asy_Knee | $F_{COMP}$ | 0.062 | -0.388 | 0.086 | -0.310 |
| | | | $F_{AP}$ | -0.024 | -0.248 | 0.039 | -0.227 |
| | | | $F_{ML}$ | 0.167 | -0.598 | 0.268 | -0.352 |
| | Laevo | Sym_Ground | $F_{COMP}$ | 0.010 | 0.343 | -0.227 | 0.253 |
| | | | $F_{AP}$ | 0.064 | 0.510 | -0.164 | 0.427 |
| | | | $F_{ML}$ | -0.139 | -0.312 | -0.374 | -0.460 |
| | | Sym_Knee | $F_{COMP}$ | -0.194 | -0.489 | -0.183 | -0.468 |
| | | | $F_{AP}$ | -0.199 | -0.364 | -0.145 | -0.341 |
| | | | $F_{ML}$ | -0.107 | -0.097 | -0.162 | -0.174 |
| | | Asy_Knee | $F_{COMP}$ | -0.103 | -0.079 | -0.172 | -0.376 |
| | | | $F_{AP}$ | -0.053 | 0.034 | -0.197 | -0.179 |
| | | | $F_{ML}$ | -0.156 | -0.514 | -0.274 | **-0.729** |
| Male | BackX | Sym_Ground | $F_{COMP}$ | -0.538 | **-0.720** | **-0.681** | **-0.709** |
| | | | $F_{AP}$ | -0.495 | -0.367 | -0.616 | -0.352 |
| | | | $F_{ML}$ | 0.300 | 0.129 | 0.145 | 0.091 |
| | | Sym_Knee | $F_{COMP}$ | -0.056 | -0.433 | 0.039 | -0.428 |
| | | | $F_{AP}$ | -0.114 | -0.522 | -0.026 | -0.534 |
| | | | $F_{ML}$ | -0.025 | **0.807** | -0.045 | **0.731** |
| | | Asy_Knee | $F_{COMP}$ | 0.362 | -0.509 | 0.463 | -0.405 |
| | | | $F_{AP}$ | 0.390 | -0.503 | 0.394 | -0.423 |
| | | | $F_{ML}$ | 0.155 | -0.479 | 0.299 | -0.357 |
| | Laevo | Sym_Ground | $F_{COMP}$ | -0.581 | -0.424 | -0.391 | -0.229 |
| | | | $F_{AP}$ | -0.608 | -0.452 | -0.405 | -0.264 |
| | | | $F_{ML}$ | 0.204 | 0.015 | 0.211 | 0.120 |
| | | Sym_Knee | $F_{COMP}$ | 0.574 | 0.191 | 0.548 | 0.135 |
| | | | $F_{AP}$ | 0.520 | 0.220 | 0.480 | 0.160 |
| | | | $F_{ML}$ | 0.346 | 0.226 | 0.404 | 0.232 |
| | | Asy_Knee | $F_{COMP}$ | 0.415 | 0.333 | 0.633 | 0.284 |
| | | | $F_{AP}$ | 0.456 | 0.331 | 0.670 | 0.277 |
| | | | $F_{ML}$ | 0.054 | 0.249 | 0.378 | 0.162 |

Figure 4.4. Regression equations, coefficients of determination ($r^2$), and root mean squared errors (RMSE) for the BSE-related changes in metrics of muscle activity (ΔTEM and ΔTTM) and predicted spine forces in the axial ($\Delta F_{COMP}$), anteroposterior ($\Delta F_{AP}$) and mediolateral ($\Delta F_{ML}$) directions during lowering and lifting phases.

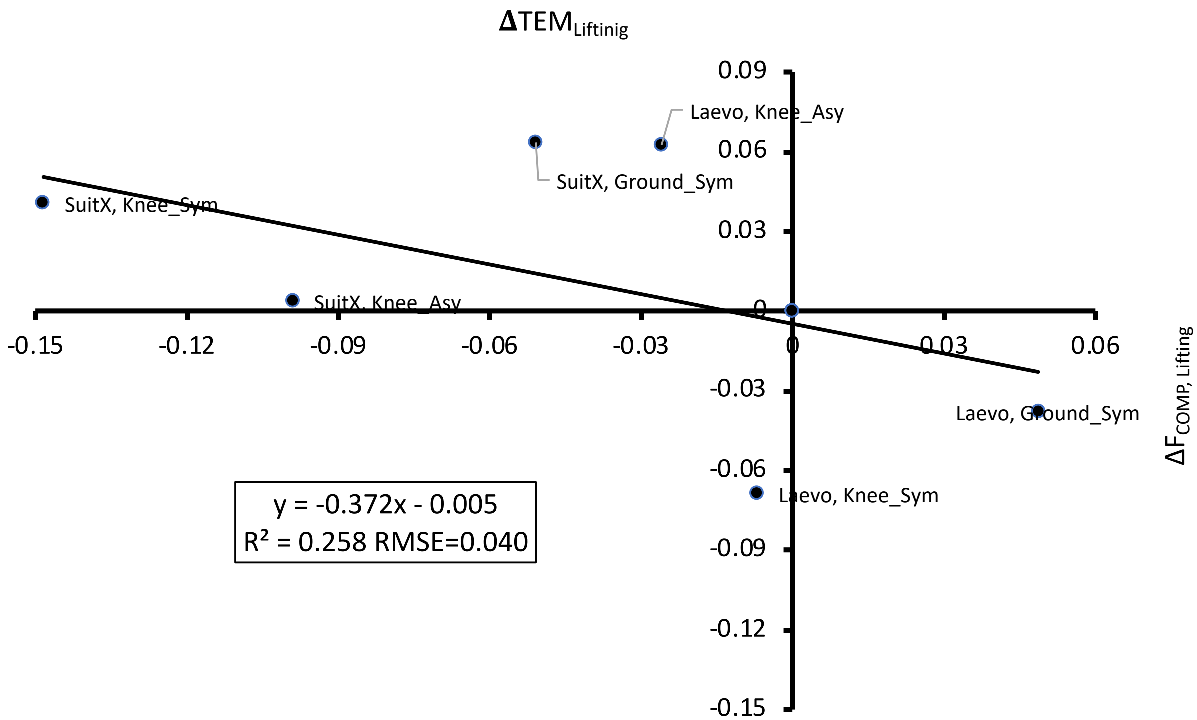


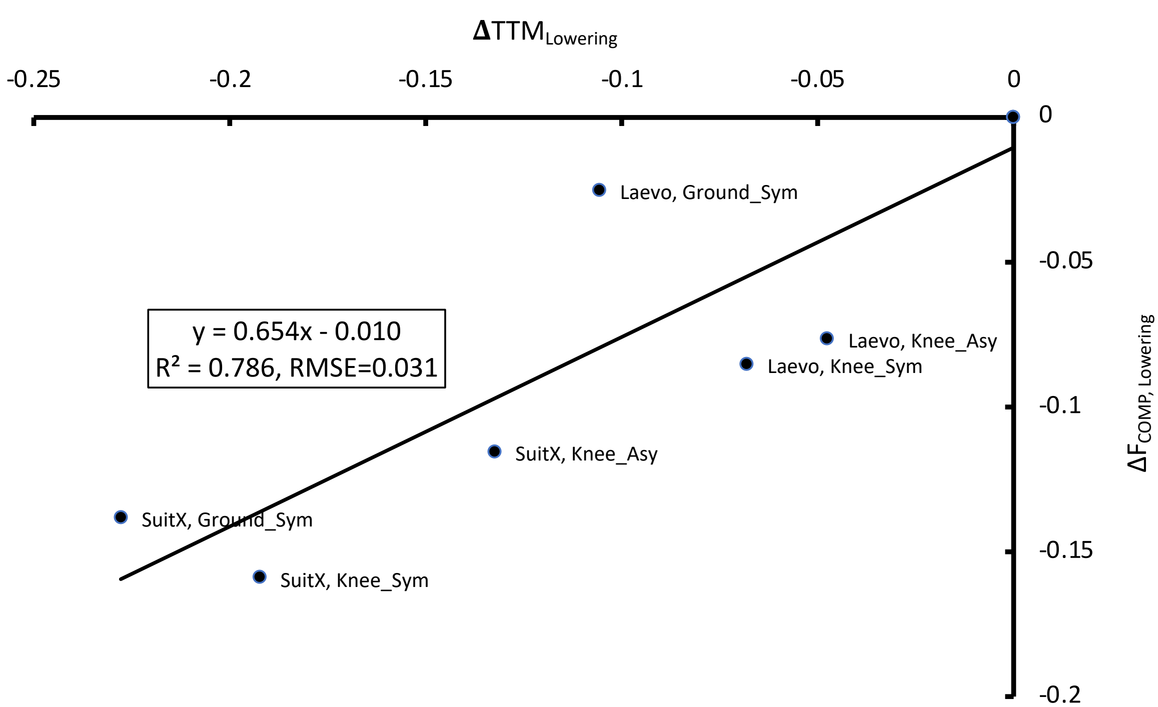


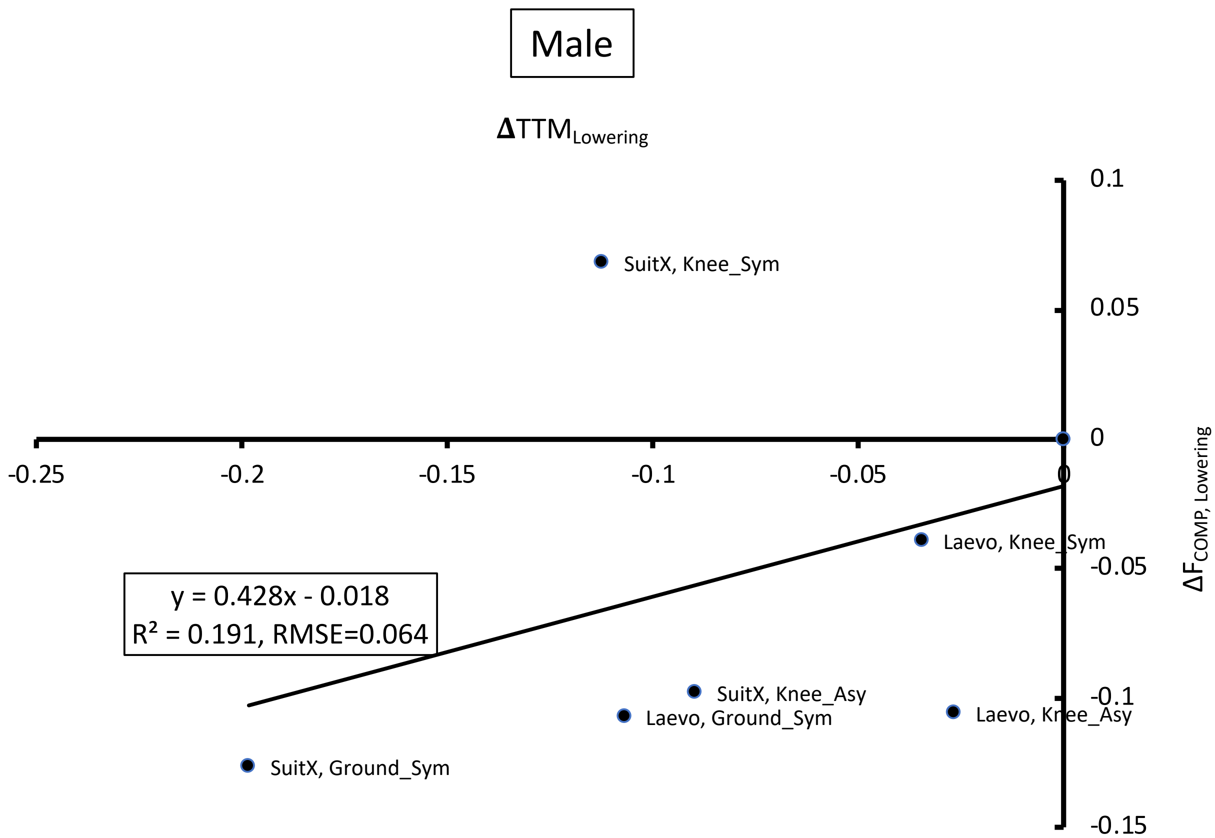
Female
ΔTEM Liftinig
0.03
0
-0.03
-0.06
-0.09
-0.12
-0.15
-0.15
-0.12
-0.09
-0.06
-0.03
0
Laevo, Ground_Sym
Laevo, Knee_Sym
SuitX, Knee_Sym
Laevo, Knee_Asy
SuitX, Ground_Sym
SuitX, Knee_Asy
y = 0.701x - 0.003
R² = 0.614, RMSE=0.033
ΔF COMP, Lifting

Male

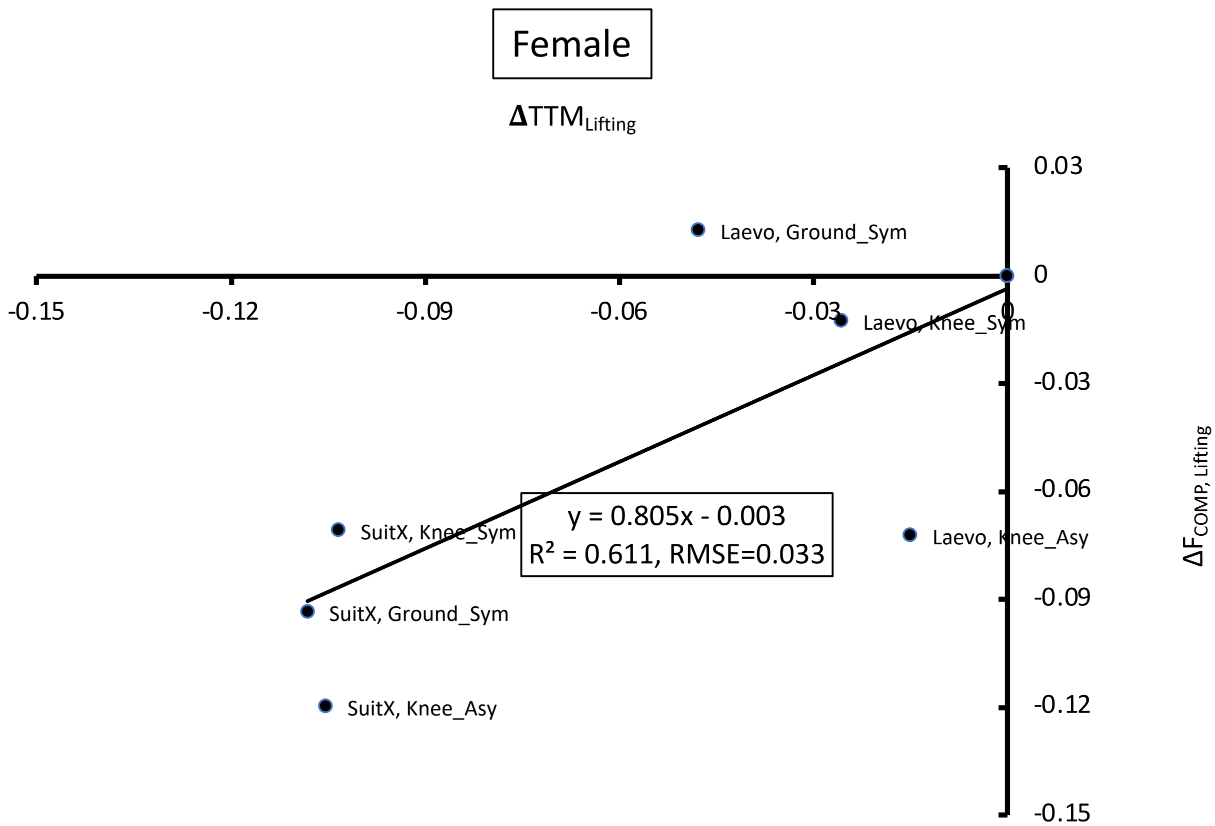

ΔTEMLiftinig
0.09
0.06
0.03
0
-0.03
-0.06
-0.09
-0.12
-0.15
-0.15
-0.12
-0.09
-0.06
-0.03
0.03
0.06
ΔFCOMP, Lifting
Laevo, Knee_Asy
SuitX, Ground_Sym
SuitX, Knee_Sym
SuitX, Knee_Asy
Laevo, Ground_Sym
Laevo, Knee_Sym
y = -0.372x - 0.005
R² = 0.258 RMSE=0.040

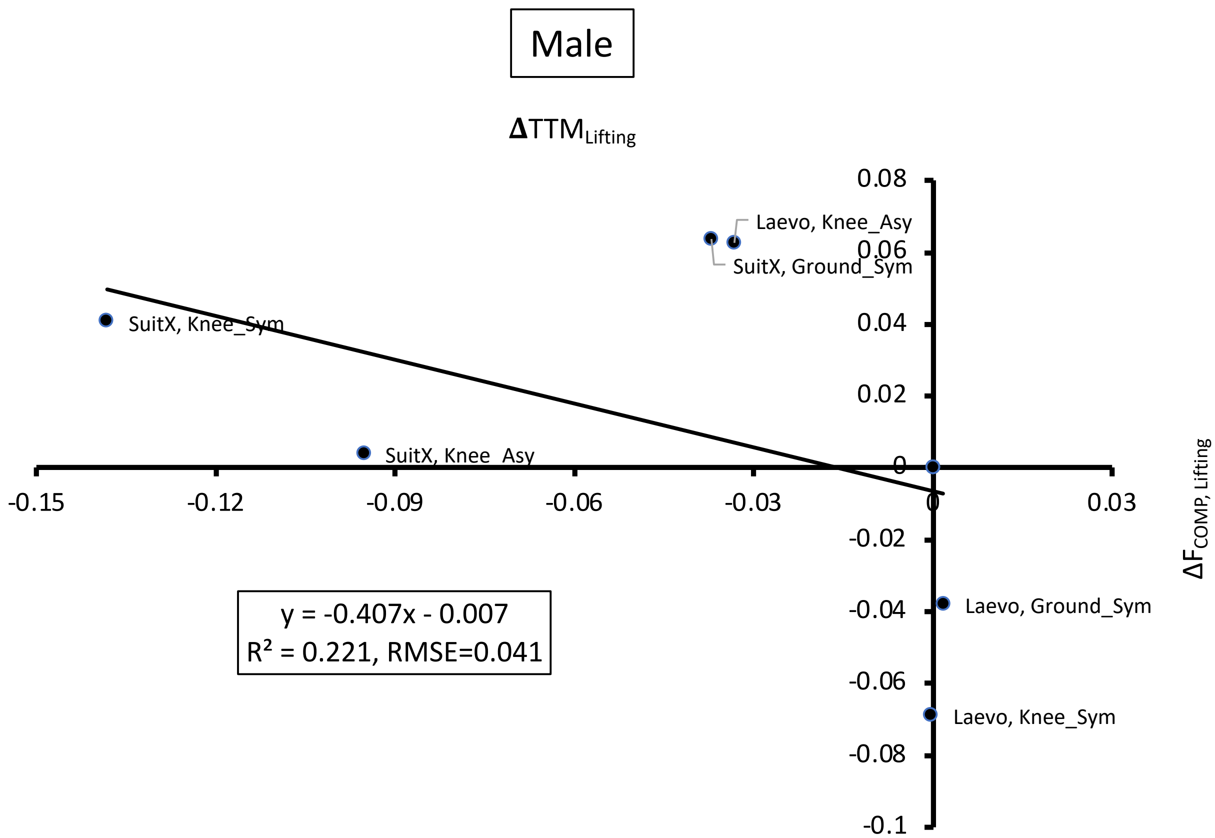
Female

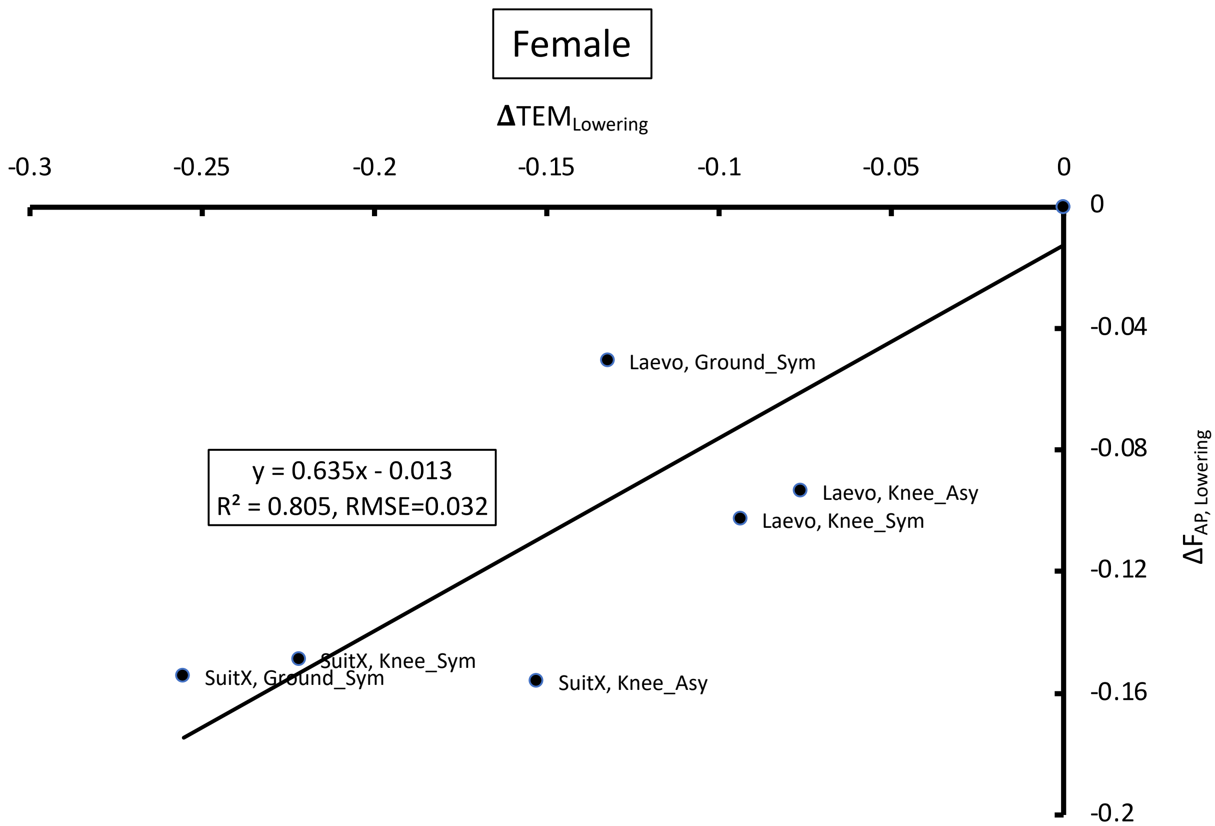

ΔTTMLowering
-0.25
-0.2
-0.15
-0.1
-0.05
0
-0.05
-0.1
-0.15
-0.2
ΔFCOMP, Lowering
Laevo, Ground_Sym
Laevo, Knee_Asy
Laevo, Knee_Sym
SuitX, Knee_Asy
SuitX, Ground_Sym
SuitX, Knee_Sym
y = 0.654x - 0.010
R² = 0.786, RMSE=0.031



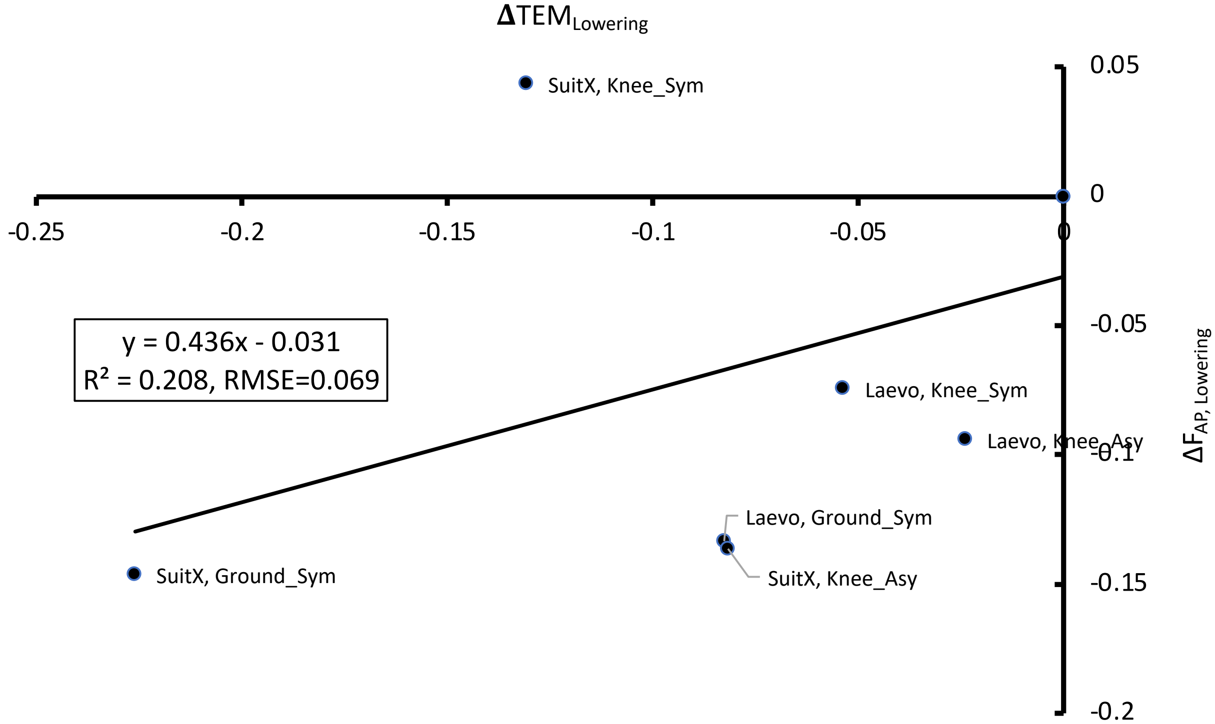
Male
ΔTTM Lowering
ΔF COMP, Lowering
0.1
0.05
0
-0.05
-0.1
-0.15
-0.25
-0.2
-0.15
-0.1
-0.05
0
SuitX, Knee_Sym
Laevo, Knee_Sym
y = 0.428x - 0.018
R² = 0.191, RMSE=0.064
SuitX, Knee_Asy
Laevo, Ground_Sym
Laevo, Knee_Asy
SuitX, Ground_Sym


Female
ΔTTM$_{Lifting}$
0.03
0
-0.03
-0.06
-0.09
-0.12
-0.15
-0.15
-0.12
-0.09
-0.06
-0.03
0
Laevo, Ground_Sym
Laevo, Knee_Sym
SuitX, Knee_Sym
y = 0.805x - 0.003
R² = 0.611, RMSE=0.033
Laevo, Knee_Asy
SuitX, Ground_Sym
SuitX, Knee_Asy
ΔF$_{COMP, Lifting}$

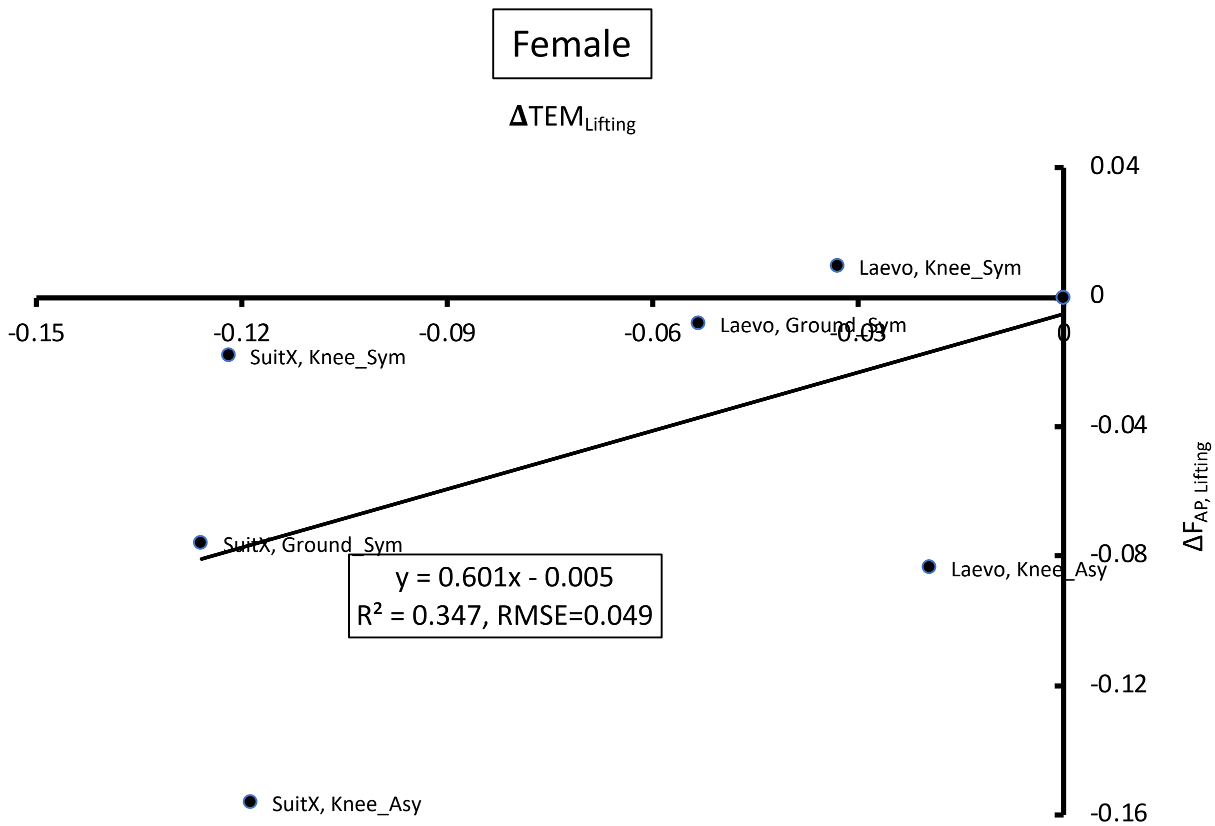

Male
ΔTTM Lifting
0.08
0.06
0.04
0.02
0
-0.02
-0.04
-0.06
-0.08
-0.1
-0.15
-0.12
-0.09
-0.06
-0.03
0
0.03
ΔF COMP, Lifting
Laevo, Knee_Asy
SuitX, Ground_Sym
SuitX, Knee_Sym
SuitX, Knee_Asy
Laevo, Ground_Sym
Laevo, Knee_Sym
y = -0.407x - 0.007
R² = 0.221, RMSE=0.041

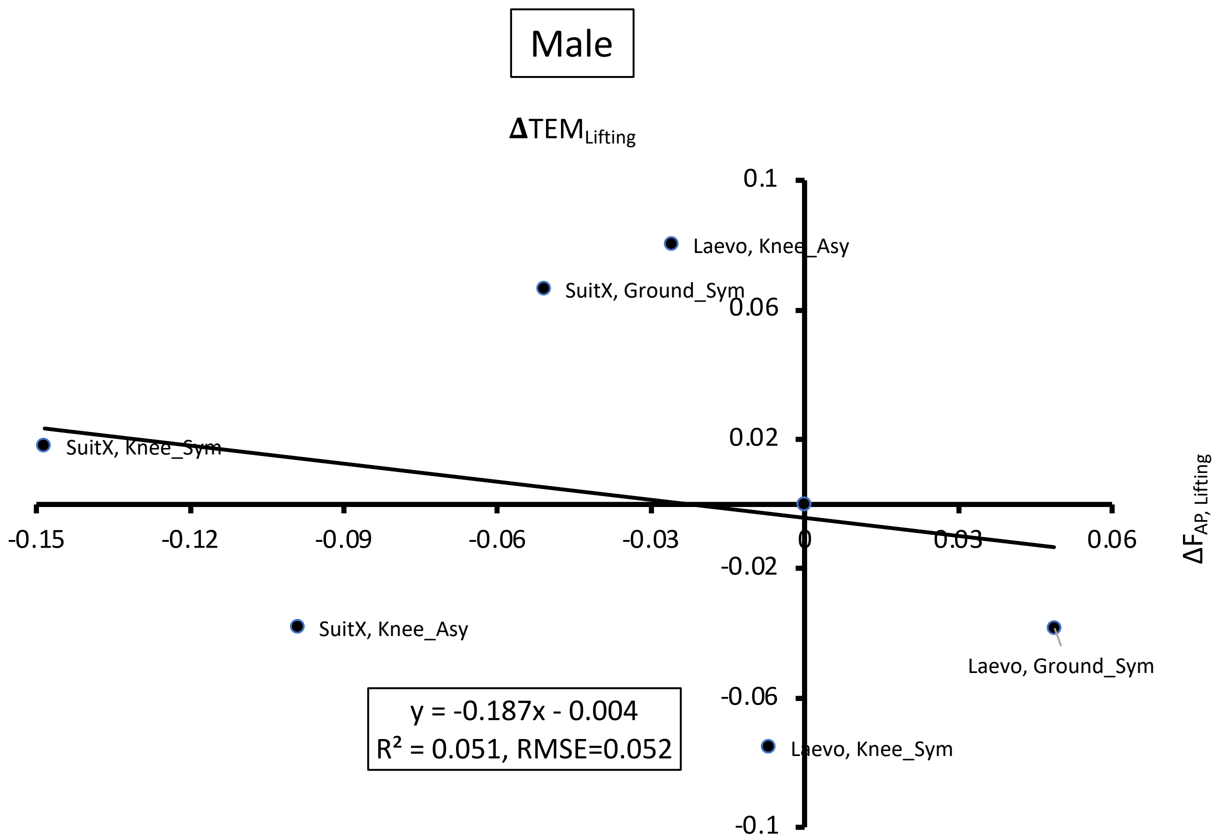

Female
ΔTEM_Lowering
ΔF_AP, Lowering
y = 0.635x - 0.013
R² = 0.805, RMSE=0.032
Laevo, Ground_Sym
Laevo, Knee_Asy
Laevo, Knee_Sym
SuitX, Knee_Sym
SuitX, Ground_Sym
SuitX, Knee_Asy
-0.3
-0.25
-0.2
-0.15
-0.1
-0.05
0
-0.04
-0.08
-0.12
-0.16
-0.2

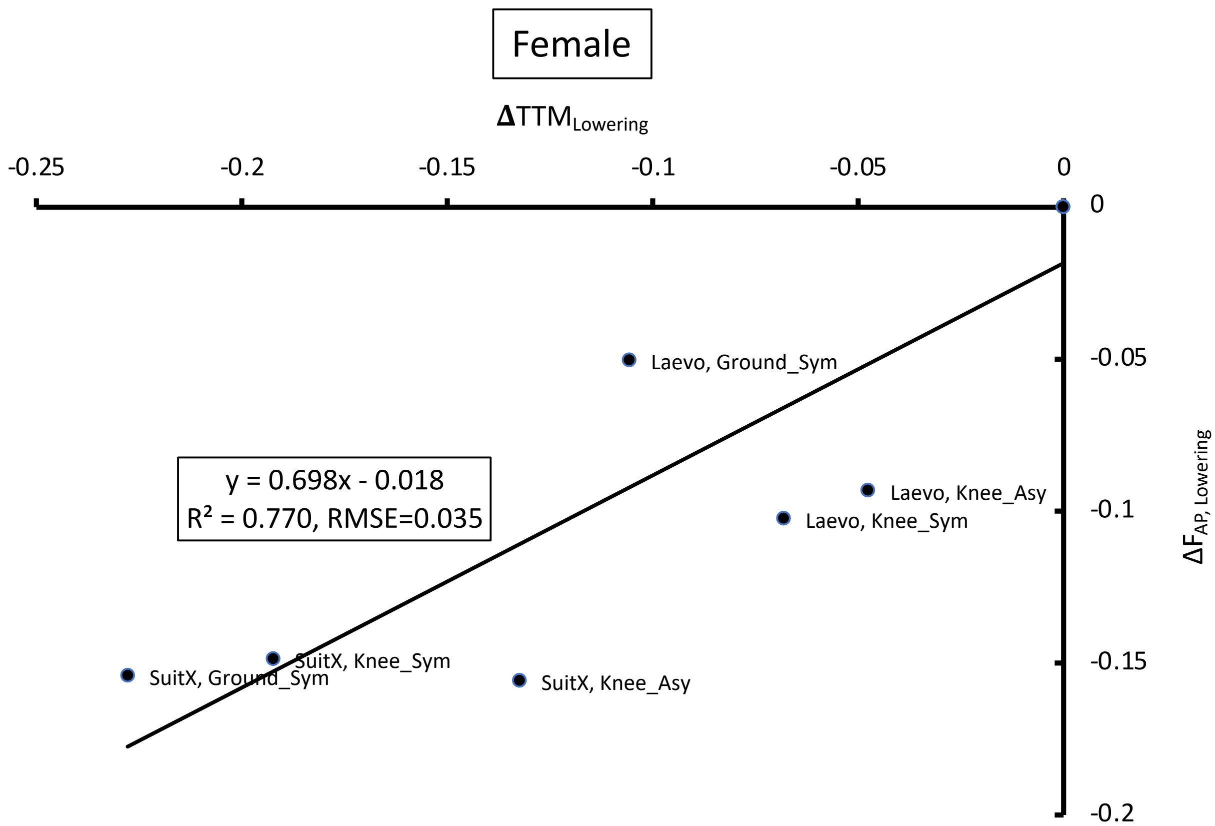
Male

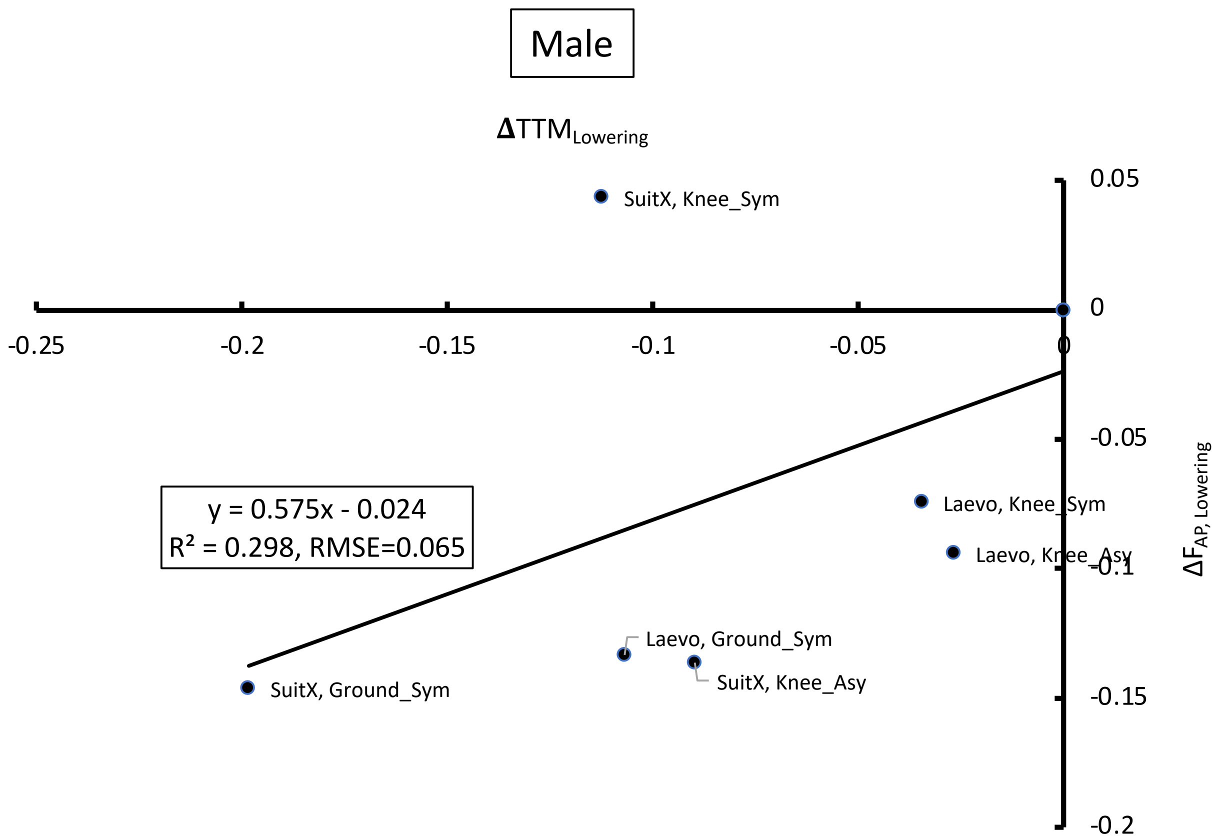


ΔTEM_Lowering
0.05
0
-0.05
-0.1
-0.15
-0.2
-0.25
-0.2
-0.15
-0.1
-0.05
0
ΔF_AP, Lowering
y = 0.436x - 0.031
R² = 0.208, RMSE=0.069
SuitX, Knee_Sym
Laevo, Knee_Sym
Laevo, Knee_Asy
Laevo, Ground_Sym
SuitX, Knee_Asy
SuitX, Ground_Sym

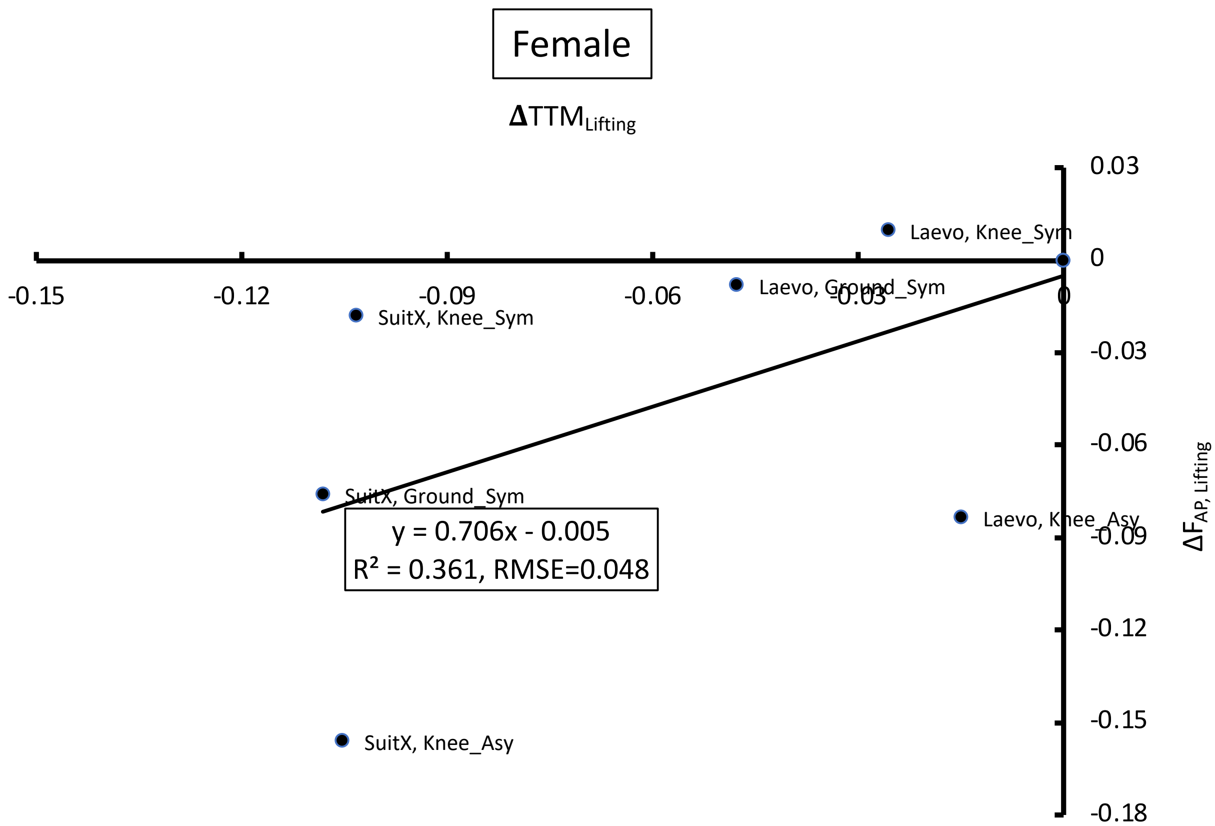

Female
ΔTEM Lifting
ΔF AP, Lifting
0.04
0
-0.04
-0.08
-0.12
-0.16
-0.15
-0.12
-0.09
-0.06
-0.03
0
Laevo, Knee_Sym
Laevo, Ground_Sym
SuitX, Knee_Sym
SuitX, Ground_Sym
Laevo, Knee_Asy
SuitX, Knee_Asy
y = 0.601x - 0.005
R² = 0.347, RMSE=0.049


Male
ΔTEM_Lifting
0.1
Laevo, Knee_Asy
SuitX, Ground_Sym
0.06
0.02
SuitX, Knee_Sym
-0.15
-0.12
-0.09
-0.06
-0.03
0
0.03
0.06
ΔF_AP, Lifting
-0.02
SuitX, Knee_Asy
Laevo, Ground_Sym
-0.06
y = -0.187x - 0.004
R² = 0.051, RMSE=0.052
Laevo, Knee_Sym
-0.1


Female
ΔTTM_Lowering
-0.25
-0.2
-0.15
-0.1
-0.05
0
0
-0.05
-0.1
-0.15
-0.2
ΔF_AP, Lowering
y = 0.698x - 0.018
R² = 0.770, RMSE=0.035
Laevo, Ground_Sym
Laevo, Knee_Asy
Laevo, Knee_Sym
SuitX, Knee_Sym
SuitX, Ground_Sym
SuitX, Knee_Asy

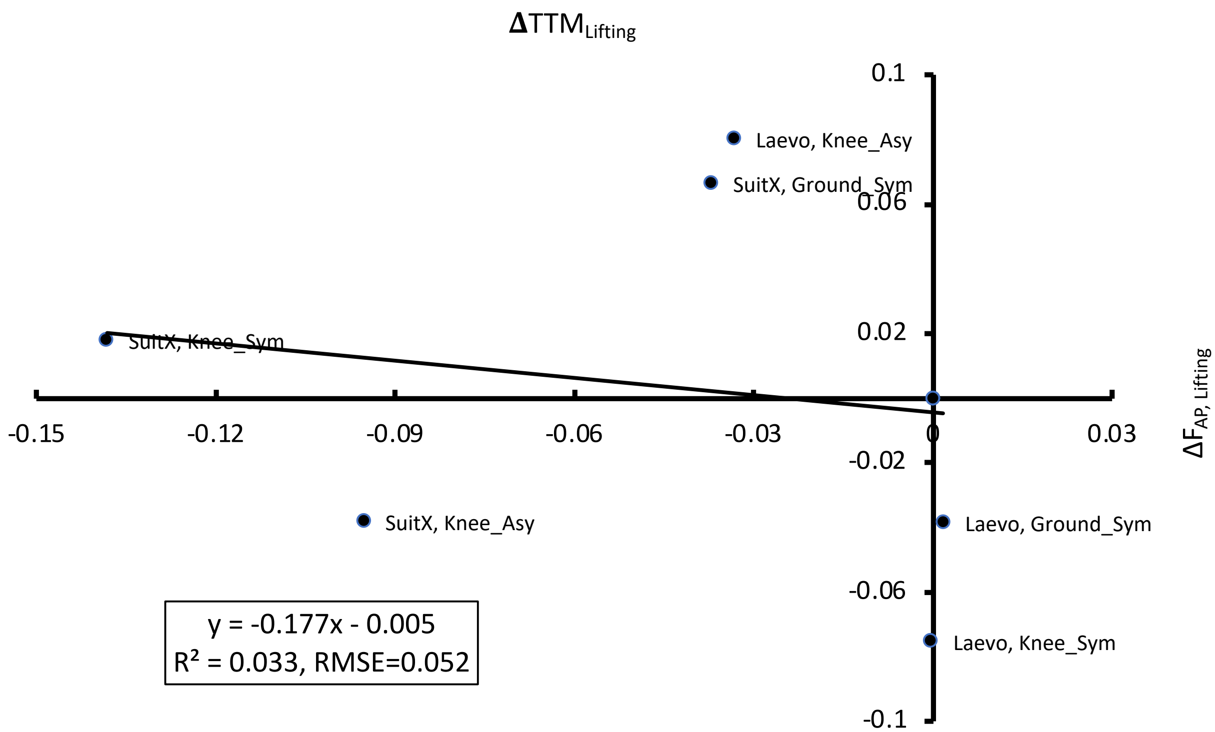


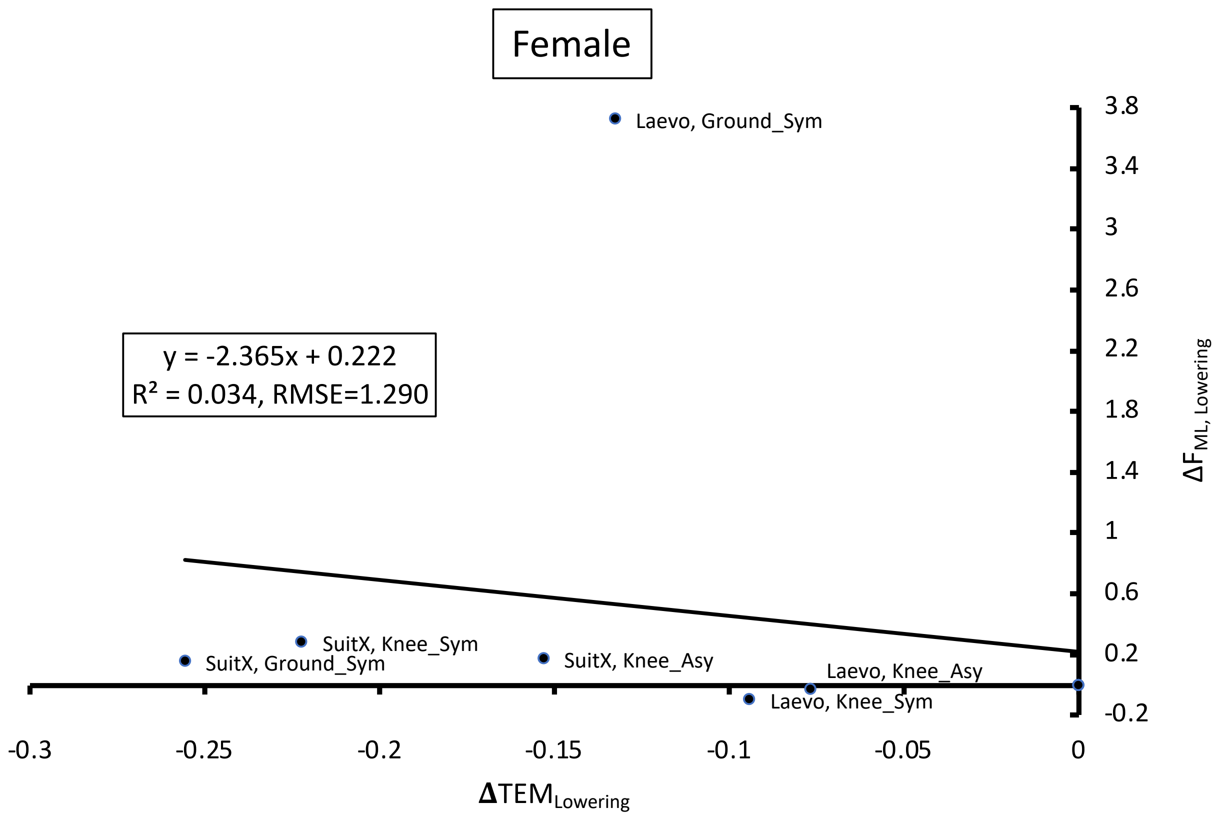
Male
ΔTTM_Lowering
0.05
0
-0.05
-0.1
-0.15
-0.2
-0.25
-0.2
-0.15
-0.1
-0.05
0
ΔF_AP, Lowering
SuitX, Knee_Sym
Laevo, Knee_Sym
Laevo, Knee_Asy
Laevo, Ground_Sym
SuitX, Knee_Asy
SuitX, Ground_Sym
y = 0.575x - 0.024
R² = 0.298, RMSE=0.065



Female
ΔTTM_Lifting
0.03
0
-0.03
-0.06
-0.09
-0.12
-0.15
-0.18
-0.15
-0.12
-0.09
-0.06
-0.03
0
ΔF_AP, Lifting
Laevo, Knee_Sym
Laevo, Ground_Sym
SuitX, Knee_Sym
SuitX, Ground_Sym
Laevo, Knee_Asy
SuitX, Knee_Asy
y = 0.706x - 0.005
R² = 0.361, RMSE=0.048

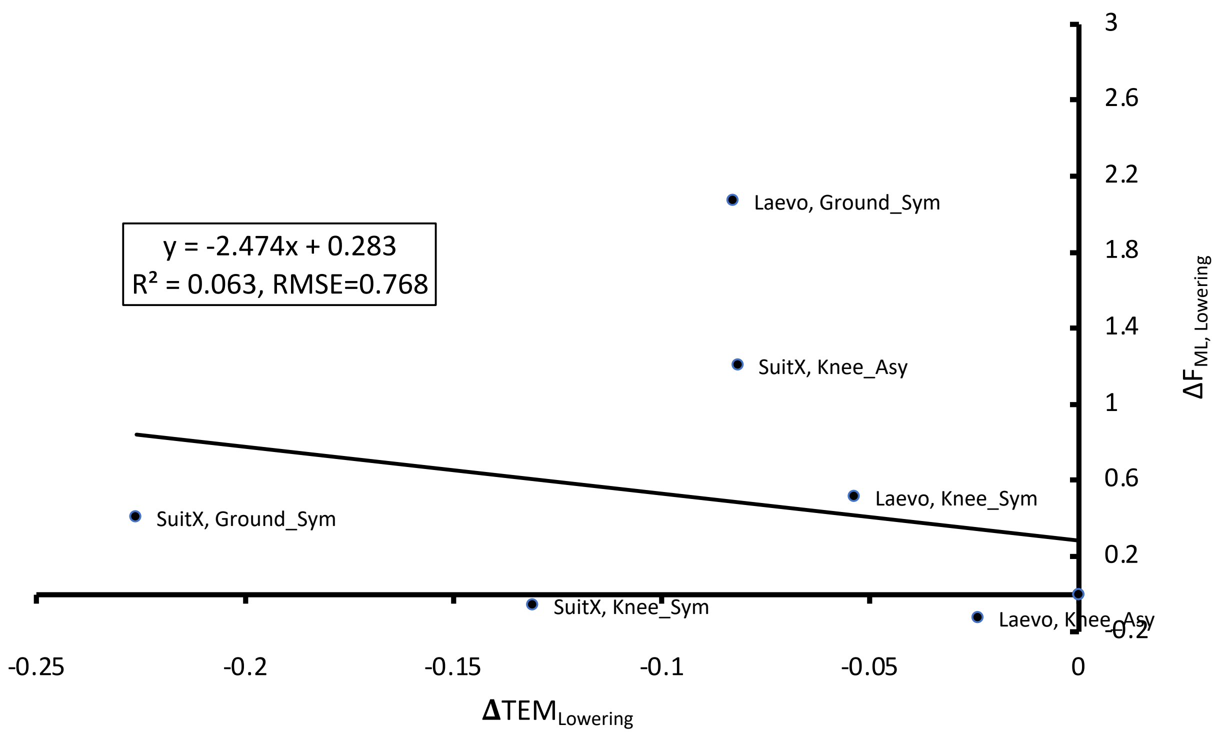
Male


ΔTTM_Lifting
ΔF_AP, Lifting
0.1
0.06
0.02
-0.02
-0.06
-0.1
-0.15
-0.12
-0.09
-0.06
-0.03
0
0.03
Laevo, Knee_Asy
SuitX, Ground_Sym
SuitX, Knee_Sym
SuitX, Knee_Asy
Laevo, Ground_Sym
Laevo, Knee_Sym
y = -0.177x - 0.005
R² = 0.033, RMSE=0.052

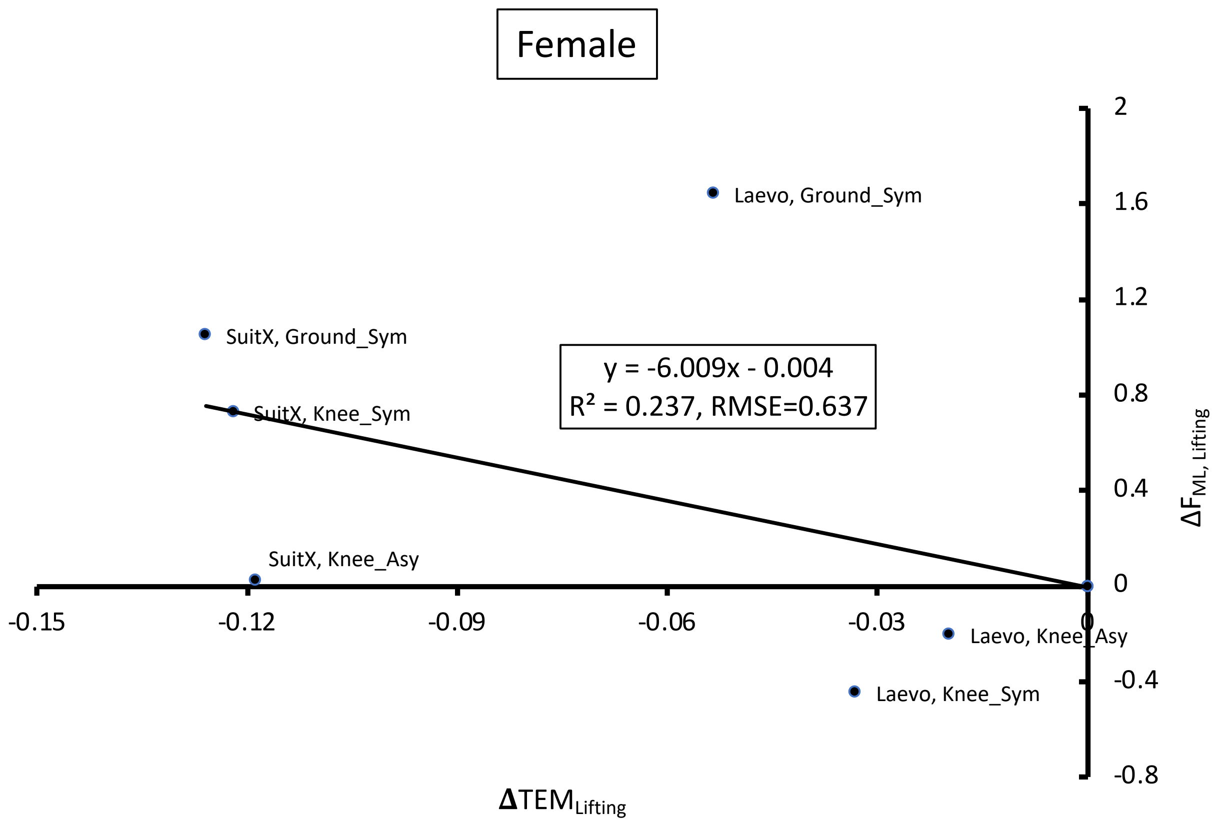


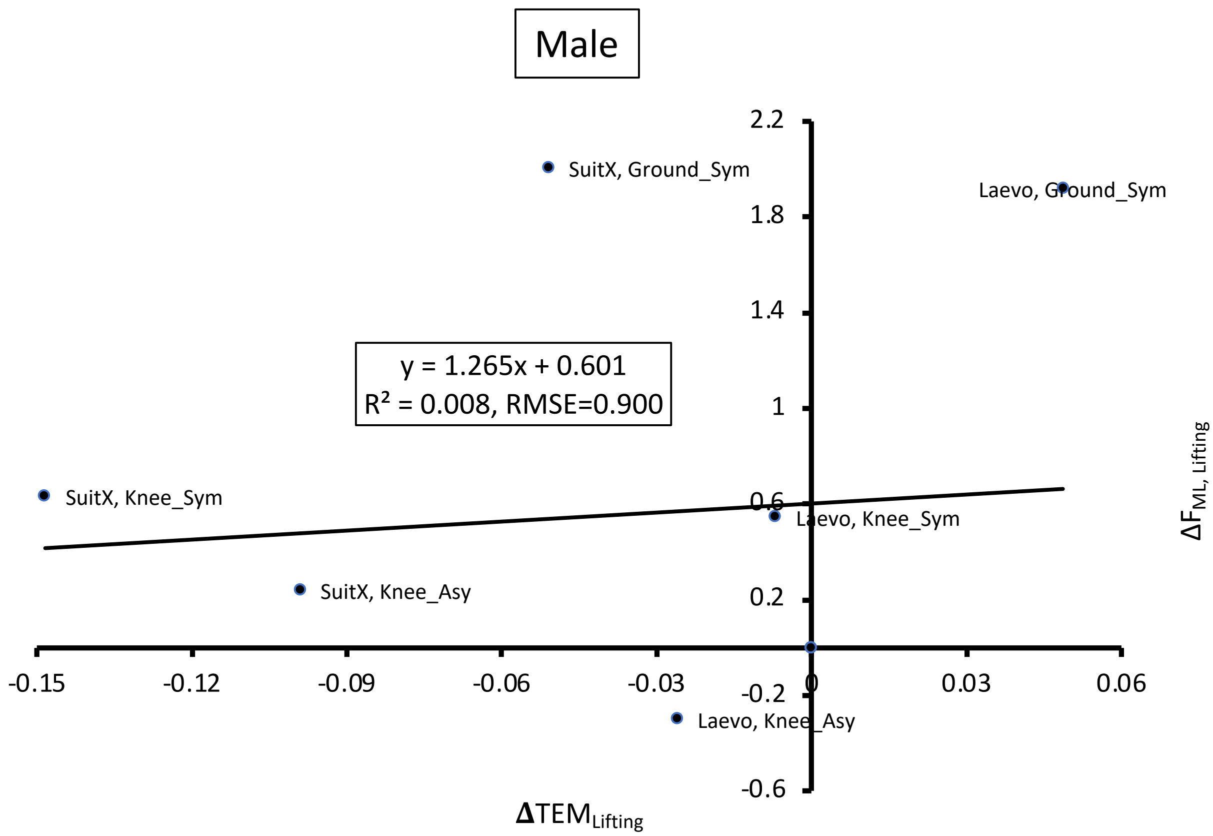
Female
y = -2.365x + 0.222
R² = 0.034, RMSE=1.290
Laevo, Ground_Sym
SuitX, Knee_Sym
SuitX, Ground_Sym
SuitX, Knee_Asy
Laevo, Knee_Asy
Laevo, Knee_Sym
3.8
3.4
3
2.6
2.2
1.8
1.4
1
0.6
0.2
-0.2
ΔF ML, Lowering
-0.3
-0.25
-0.2
-0.15
-0.1
-0.05
0
ΔTEM Lowering

Male

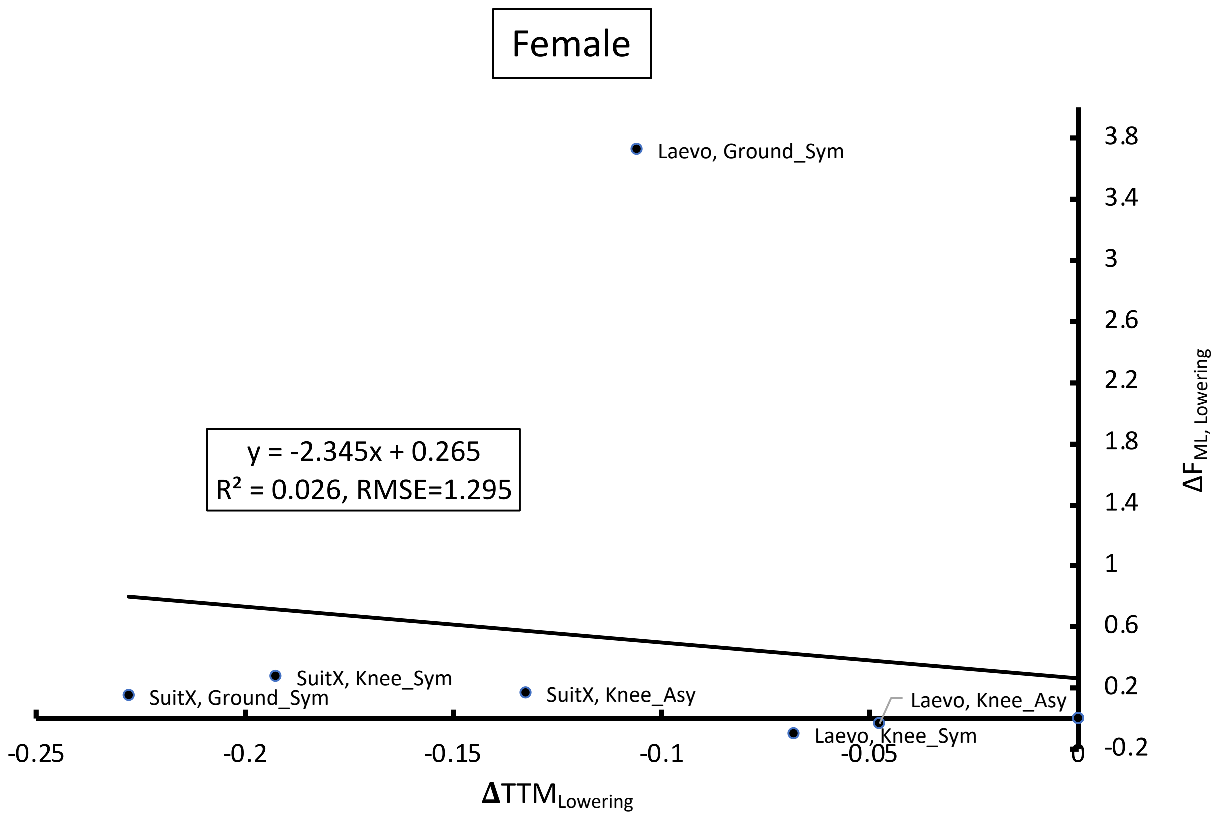


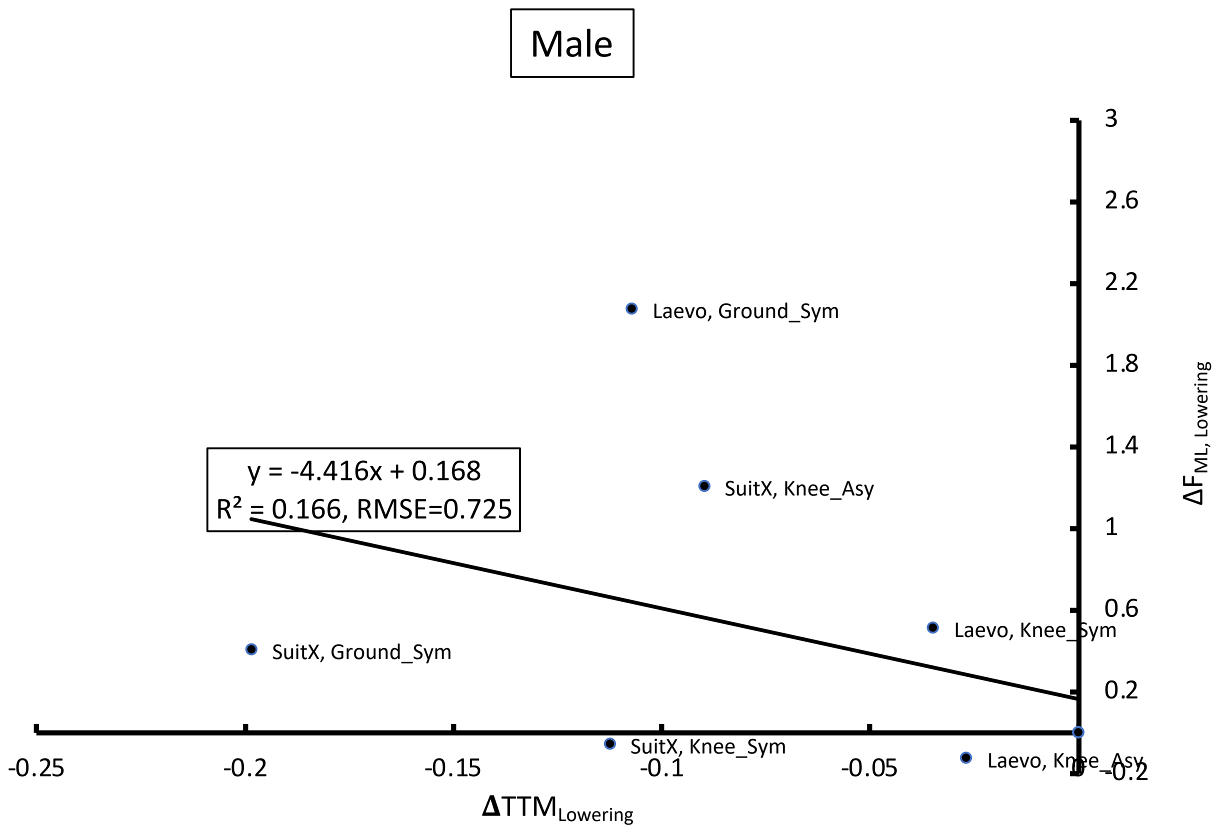
y = -2.474x + 0.283
R² = 0.063, RMSE=0.768
Laevo, Ground_Sym
SuitX, Knee_Asy
Laevo, Knee_Sym
SuitX, Ground_Sym
SuitX, Knee_Sym
Laevo, Knee_Asy
3
2.6
2.2
1.8
1.4
1
0.6
0.2
-0.2
-0.25
-0.2
-0.15
-0.1
-0.05
0
ΔTEM Lowering
ΔF ML, Lowering


Female
2
1.6
1.2
0.8
0.4
0
-0.4
-0.8
Laevo, Ground_Sym
SuitX, Ground_Sym
SuitX, Knee_Sym
SuitX, Knee_Asy
Laevo, Knee_Asy
Laevo, Knee_Sym
y = -6.009x - 0.004
R² = 0.237, RMSE=0.637
-0.15
-0.12
-0.09
-0.06
-0.03
0
ΔF ML, Lifting
ΔTEM Lifting

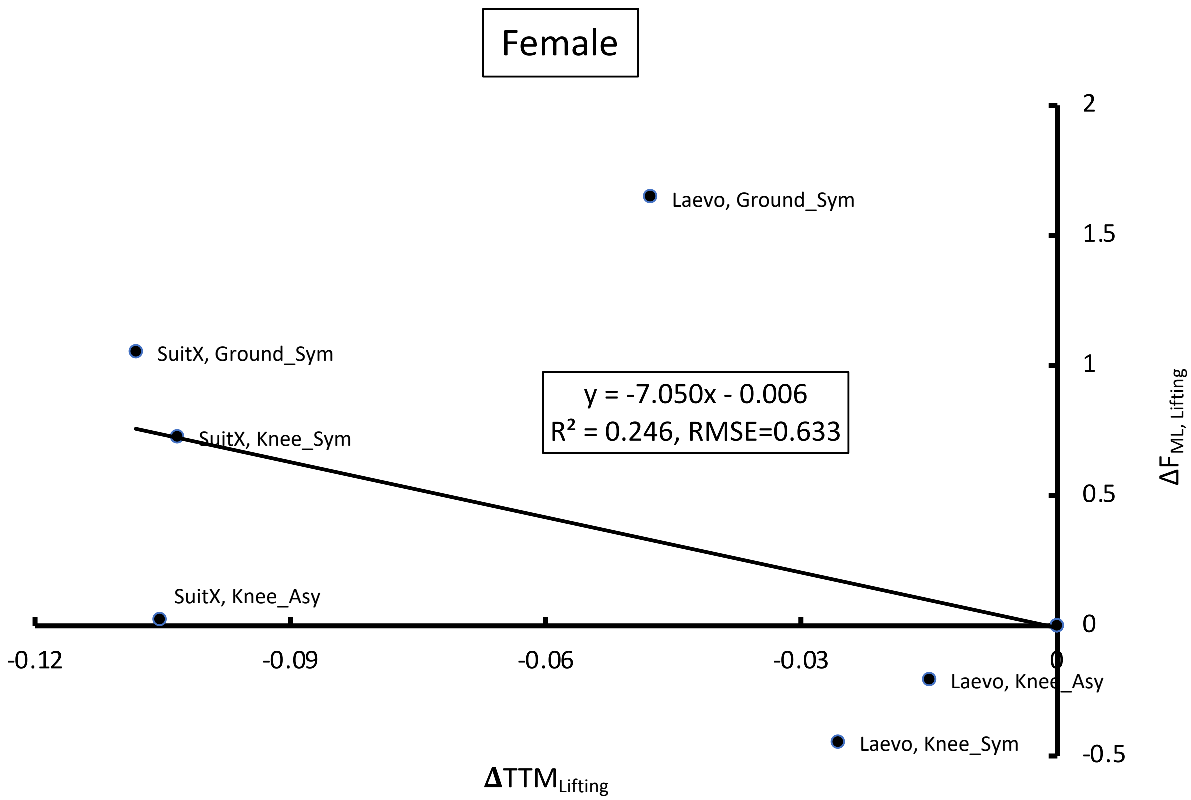


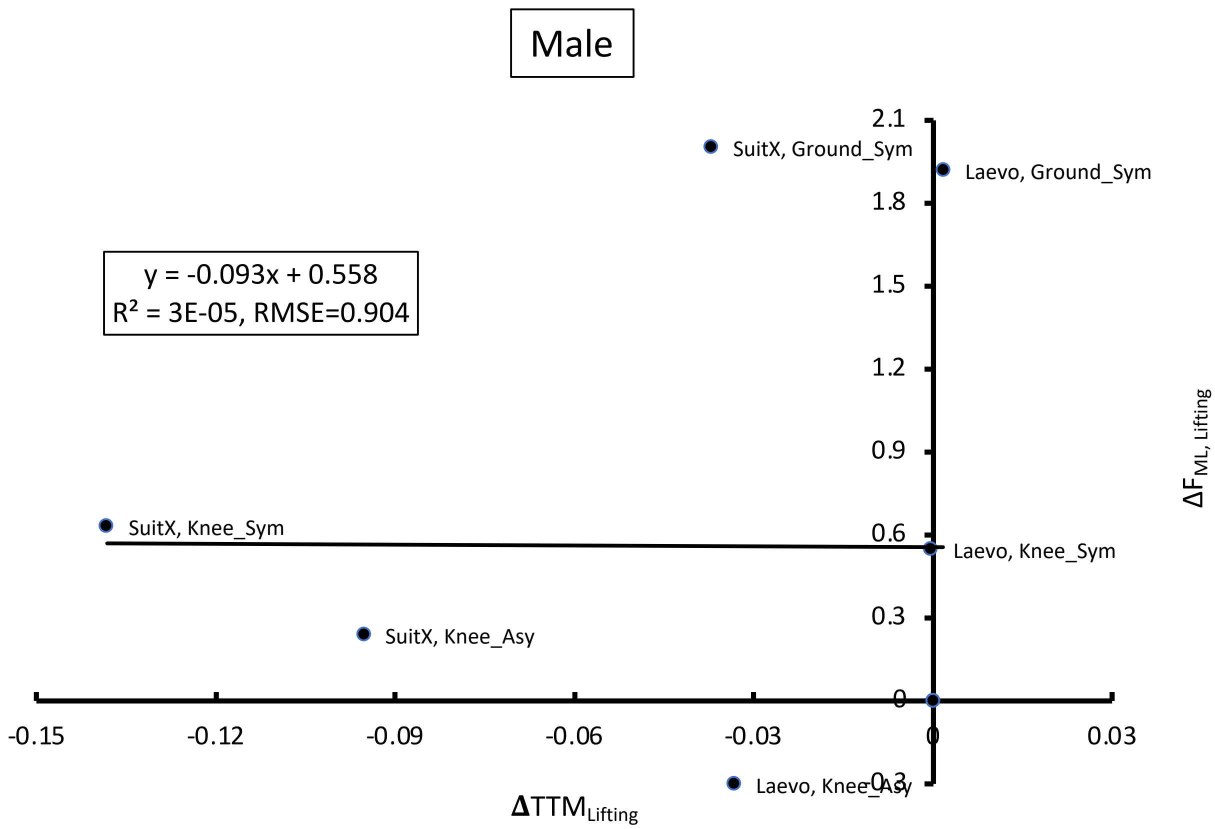
Male
y = 1.265x + 0.601
R² = 0.008, RMSE=0.900
SuitX, Ground_Sym
Laevo, Ground_Sym
SuitX, Knee_Sym
Laevo, Knee_Sym
SuitX, Knee_Asy
Laevo, Knee_Asy
2.2
1.8
1.4
1
0.6
0.2
-0.2
-0.6
-0.15
-0.12
-0.09
-0.06
-0.03
0
0.03
0.06
ΔTEM Lifting
ΔF ML, Lifting


Female
Laevo, Ground_Sym
y = -2.345x + 0.265
R² = 0.026, RMSE=1.295
SuitX, Knee_Sym
SuitX, Ground_Sym
SuitX, Knee_Asy
Laevo, Knee_Asy
Laevo, Knee_Sym
3.8
3.4
3
2.6
2.2
1.8
1.4
1
0.6
0.2
-0.2
-0.25
-0.2
-0.15
-0.1
-0.05
0
ΔTTM Lowering
ΔF ML, Lowering



Male
y = -4.416x + 0.168
R² = 0.166, RMSE=0.725
Laevo, Ground_Sym
SuitX, Knee_Asy
Laevo, Knee_Sym
SuitX, Ground_Sym
SuitX, Knee_Sym
Laevo, Knee_Asy
3
2.6
2.2
1.8
1.4
1
0.6
0.2
-0.2
-0.25
-0.2
-0.15
-0.1
-0.05
0
ΔF ML, Lowering
ΔTTM Lowering


Female
y = -7.050x - 0.006
R² = 0.246, RMSE=0.633
Laevo, Ground_Sym
SuitX, Ground_Sym
SuitX, Knee_Sym
SuitX, Knee_Asy
Laevo, Knee_Asy
Laevo, Knee_Sym
2
1.5
1
0.5
0
-0.5
-0.12
-0.09
-0.06
-0.03
0
ΔTTM_Lifting
ΔF_ML, Lifting


Male
y = -0.093x + 0.558
R² = 3E-05, RMSE=0.904
SuitX, Ground_Sym
Laevo, Ground_Sym
SuitX, Knee_Sym
Laevo, Knee_Sym
SuitX, Knee_Asy
Laevo, Knee_Asy
2.1
1.8
1.5
1.2
0.9
0.6
0.3
0
-0.3
-0.15
-0.12
-0.09
-0.06
-0.03
0
0.03
ΔTTM Lifting
ΔF ML, Lifting

# 5 Chapter 5: Effects of Back-Support Exoskeleton Use on Trunk Neuromuscular Control during Repetitive Lifting: A Dynamical Systems Analysis[2]

## 5.1 Abstract

Back-support exoskeletons (BSEs) are a promising ergonomic intervention to mitigate the risk of occupational low back pain. Although growing evidence points to the beneficial effects of BSEs, specifically in reducing low-back physical demands, there is limited understanding of potential unintended consequences of BSE use on neuromuscular control of the trunk during manual material handling (MMH). We quantified the effects of two passive BSEs (BackX™ AC and Laevo™ V2.5) on trunk dynamic stability and movement coordination during a repetitive lifting task. Eighteen participants (gender-balanced) completed four minutes of repetitive lifting in nine different conditions, involving symmetric and asymmetric postures when using the BSEs (along with no BSE as a control condition). Maximum Lyapunov exponents (short-term: $\lambda_{\mathrm{max-s}}$; long-term: $\lambda_{\mathrm{max-l}}$) and Floquet multipliers ($FM_{\max}$) were respectively calculated to quantify the local dynamic and orbital stability of thorax and pelvis trajectories. Thorax-pelvis segmental coordination was also quantified using the continuous relative phase. Wearing the Laevo™ significantly increased $\lambda_{\mathrm{max-s}}$ for the pelvis (by ~8%) and $FM_{\max}$ for the thorax and pelvis (by ~5-10%). Use of either BSE decreased the in-phase coordination pattern for the thorax-pelvis coupling (by ~15%). These results suggest that BSE use can compromise neuromuscular control of the trunk, and caution should thus be used in selecting a suitable BSE for use in a given MMH task.

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[2] This study has been completed: Madinei, S., Kim, S., Srinivasan, D. and Nussbaum, M.A. (2020). Effects of Back-Support Exoskeleton Use on Trunk Neuromuscular Control during Repetitive Lifting: A Dynamical Systems Analysis. *Journal of Biomechanics, 123, 110501, https://doi.org/10.1016/j.jbiomech.2021.110501*.

Future work is needed, however, to assess the generalizability of different BSE design approaches in terms of unintended short-term and long-term effects on trunk neuromuscular control.

## 5.2 Introduction

Manual material handling tasks expose workers to well-documented risk factors for work-related musculoskeletal disorders (WMSDs), such as forceful exertions and repetitive bending and lifting (Hoogendoorn et al., 2000; da Costa & Vieira, 2010), with the back being the most affected body region (BLS, 2019). Industrial back-support exoskeletons/exosuits (BSEs) have been introduced as a new intervention to reduce physical demands on the spine (De Looze et al., 2016). BSEs are wearable assistive devices designed to support, augment, and/or assist the back and hip muscles, by passively or actively producing restorative torques (Lee et al., 2012). Among different types of BSE, passive devices are currently predominant in the commercial market due to their availability, cost-effectiveness, and ease of implementation (Nussbaum et al., 2019).

A majority of current evidence points to the beneficial effects of passive BSEs, specifically in reducing low back physical demands. Decreases in low back muscular activation, lumbar forces and moments, and localized muscle fatigue have been reported in simulated lifting tasks (e.g., Godwin et al., 2009; Wehner et al., 2009; Lamers et al., 2018; Näf et al., 2018; Baltrusch et al., 2019; Alemi et al., 2020; Madinei et al., 2020a) and while maintaining non-neutral trunk postures for prolonged durations (e.g., Graham et al., 2009; Ulrey & Fathallah, 2013; Bosch et al., 2016; Koopman et al., 2019; Lamers et al., 2020; Madinei et al., 2020b). The magnitude of such reductions, however, depend upon specific BSE designs and task characteristics.

At present, there is relatively limited evidence on potential unintended or adverse consequences of BSE use. Earlier biomechanical studies, for example, have emphasized the importance of maintaining trunk stability to avoid low back disorders (Cholewicki & McGill, 1996; Panjabi, 2003), which requires sufficient muscle activation levels and coordination, along with contributions of various passive tissues (Hoffman & Gabel, 2013). By providing external structural frames over the trunk and supportive torques about the hip/low back, a BSE may interfere with the active and passive systems and their neuromuscular control for effectively maintaining trunk stability. While an early BSE design was found to increase overall trunk stability, yet the stabilizing contributions of trunk flexors and extensors decreased by up to 46% during a sustained trunk bending task (Agnew & Stevenson, 2008). Use of this same BSE enhanced local dynamic stability of trunk movements during a repetitive lifting task (Graham et al., 2011), and led to a more synchronous intersegmental coordination between the trunk and pelvis during a repetitive lifting task, suggesting an altered motor control strategy (Agnew & Stevenson, 2008). We are not aware of any evidence regarding how current, commercially-available BSEs affect trunk stability during occupationally-relevant lifting tasks, or if these effects vary between BSE designs.

We explored the impacts of two different passive BSEs on trunk neuromuscular control behavior during repetitive symmetric and asymmetric lifting tasks. Neuromuscular control behavior in maintaining stability has often been characterized using dynamical systems theory (DST: Dingwell & Cusumano, 2000; Granata & England, 2006; Asgari et al., 2015). DST helps quantify the spatiotemporal evolution of system dynamics over a period of time, which can encompass steady state as well as abrupt qualitative changes in movement behavior (Beek et al., 1995; van Emmerik et al., 2016). Diverse nonlinear measures based on DST have been used in recent literature to quantify changes in trunk control as a function of factors such as pace, symmetry and load during repetitive/bending tasks (Lee & Nussbaum, 2013; Asgari et al., 2015;

Mokhtarinia et al., 2016; Asgari et al., 2020), and several such measures were obtained here. Based on existing evidence, we expected that trunk dynamic stability and coordination would differ when using either BSE, due to changes in active trunk stiffness and postural adaptations when using a wearable assistive device. Such changes, however, were expected to vary depending on the specific BSE design and task conditions in which the BSE is used.

## 5.3 Methods

### 5.3.1 Experimental design

Data used herein were obtained from a prior study, with details of the experimental design, procedures, and simulated tasks previously reported (Madinei et al., 2020a). Briefly, a lab-based simulation of repetitive lowering/lifting tasks was designed to examine the effects of two BSEs under three different *Lifting Conditions* with different levels of task symmetry and lifting height. Two commercially-available passive BSEs were used: SuitX BackX™ (model AC) and Laevo™ (V2.5). Both incorporate passive torque generation mechanisms about the hip that are intended to augment the trunk extensor muscles, yet they have distinct design features. Further details about these BSEs and their specific designs are provided elsewhere (Kim et al., 2020; Madinei et al., 2020a).

Symmetric lowering/lifting was done to/from two target locations set based on individual anthropometry (i.e., mid-shank and knee level), which are subsequently referred to as Sym_Ground and Sym_Knee. Asymmetric lowering/lifting was done to/from a target location set at each individual's knee level and 90° to the right of the mid-sagittal plane (Asy_Knee). This latter task only included one height, since reaching to/from mid-shank height was challenging for many participants. Tasks were performed using a wooden box, the mass of which was set to 10%

of individual body mass. Each experimental condition lasted for 4 minutes, and the pace was controlled at 10 lower/lift cycles per minute using a digital auditory metronome.

A convenience sample of 18 gender-balanced participants completed the study, none of whom had any current or recent (i.e., past 12 months) musculoskeletal disorders or injuries. Anthropometric criteria (e.g., height, waist size) were adopted from BSE user manuals to screen potential participants and ensure a proper fit to the exoskeletons. The research procedures were approved by the Institutional Review Board at Virginia Tech, and participants provided informed consent prior to any data collection.

A within-subject (repeated measure) design was used in which participants completed nine lowering/lifting trials, involving all combinations of three *Interventions* (i.e., BackX™, Laevo™, and Control = no BSE) and the three *Lifting Conditions.* The presentation order of *Intervention* conditions was first counter-balanced using 3×3 Latin Squares, and then for a given *Intervention* condition, the presentation order of Task Conditions was counter-balanced using different 3×3 Latin Squares. A minimum of 5-min resting period was provided between lowering/lifting trials. Each participant completed an initial training session (~1 hour) followed by an experimental session (~4.5 hours) on a subsequent day. Participants were allowed to freely choose their lowering/lifting style and feet location while maintaining a consistent feet location for the entire trial.

### 5.3.2 Instrumentation and data processing

Whole-body segmental kinematics were monitored (60 Hz) using a wearable inertial motion capture system (MVN Awinda, Xsens Technologies B.V., Netherlands). Kinematic data were low-pass filtered (cutoff frequency = 10 Hz) and were analyzed following the standard rotation sequence recommended by the ISB (Wu et al., 2002). We repeated the analyses with unfiltered data and confirmed that the results were consistent in both approaches. Similar results were found in earlier work (e.g., Mehdizadeh & Sanjari, 2017; Mehdizadeh, 2018; Raffalt et al., 2020), specifically that information loss with filtering is unlikely when the filtering frequency is relatively high (e.g., >7-10 Hz). In each trial, triaxial orientations and angular velocities were obtained for the thorax, pelvis, and lumbar spine (thorax vs. pelvis), and were used to derive the measures described below. We initially performed the local dynamic and orbital stability analyses on the thorax vs. pelvis movement as well. Overall results were consistent with the thorax and pelvis measures. In the interest of brevity, we only report here measures obtained from the thorax and pelvis movements. Data analyses were performed using a custom MATLAB code equipped with the predictive maintenance toolbox (Mathworks, Natick, MA). These analyses are briefly summarized here, with additional details provided in Appendix F.

Local dynamic stability: Maximum Lyapunov exponents ($\lambda_{\max}$) were determined for the thorax and pelvis using the algorithm of Rosenstein et al. (1993). A constant sample number (i.e., 14,400 = 40 cycles × 6 sec × 60 Hz) was used to ensure that the estimates were not biased by time series length or the number of lifting cycles (Bruijn et al., 2009). Time-delay embedding was used to reconstruct multidimensional state spaces from each original time series and time-delayed copies, and was set to 10% of mean cycle duration trials (Granata & England, 2006; Graham et al., 2014; Bourdon et al., 2019). Specifically, a 12D state space was reconstructed from the 3D orientations and angular velocities of the thorax and pelvis inertial sensors. The divergence curves were visually examined to be stable prior to computing short- and long-term LyEs. Short-term LyE ($\lambda_{\max-\mathrm{s}}$) and long-term LyE ($\lambda_{\max-\mathrm{l}}$) were computed to characterize local dynamic stability, over 0-0.5 and 4-10 cycles, respectively (Dingwell & Cusumano, 2000; Dingwell & Marin, 2006;

England & Granata, 2007). Negative and positive exponents respectively indicate local stability and local instability, with larger exponents indicating a greater sensitivity to local perturbations (Kantz & Schreiber, 2004).

<u>Orbital stability</u>: Orbital stability was determined by computing maximum Floquet multipliers ($FM_{\max}$) using the Poincare section method (Hurmuzlu & Basdogan, 1994). A Poincare section is an imaginary hypersurface placed across the flow of repetitive trajectories (Argyris et al., 1994), here of trunk movements. Intersections of the trajectories with the Poincare section define a recurrence map that can quantify the behavior of neighboring trajectories. The state space was divided into 101 Poincare sections, representing increments from 0 to 100% of the lowering/lifting cycles. Since cycles started with lowering, 0% corresponded to the upright standing posture. Similar to earlier work (Asgari et al., 2015; Asgari et al., 2020), $FM_{\max}$ is reported at Poincare sections defined at 25, 50, 75, and 100% of the lowering/lifting cycle. A maximum FM <1 is interpreted as indicating that perturbations would have faded away in every direction and that the corresponding orbit remains stable (Hilborn et al., 2000).

<u>Continuous relative phase (CRP)</u>: Thorax-pelvis coordination was calculated using methods described earlier (Stergiou et al., 2004; van Emmerik et al., 2016). Segment angles and velocities in the sagittal plane were first divided into individual lowering/lifting cycles, and interpolated to 101 data points corresponding to 0-100% of the cycle. The resultant time series were then projected into a phase plane, with all values normalized to [-1, 1] at each time frame to account for differences in amplitude and frequency between segments. Normalized data were then transformed into phase angles (in rad) using the *arctangent*. Finally, differences between phase angles of the two segments were obtained to provide thorax-pelvis CRP. Two measures were then extracted from the CRP curves: 1) mean absolute relative phase (MARP); and 2) deviation phase

(DP). MARP was calculated from the mean ensemble curve, by averaging the relative phase values over the ensemble CRP curve points; DP was calculated by averaging the standard deviations of the ensemble CRP curve (deviation phase curve) at each percent of the task cycle (Stergiou et al., 2001; Mokhtarinia et al., 2016).

### 5.3.3 Statistical analyses

Separate three-way, mixed-factor analyses of variance (ANOVAs) were used to assess the effects of *Intervention*, *Lifting Condition*, and *Gender* on each of the outcome measures. Presentation orders of *Task Condition* and *Intervention* were included as blocking factors. Statistical analyses were completed using JMP Pro (v.15, SAS, Cary, NC), parametric model assumptions were verified, and significant effects were concluded when $p < 0.05$. Summary outcomes are reported as means (SD), and effect sizes were quantified using eta-squared ($\eta^2$). *Post hoc* paired comparisons were completed using Tukey's HSD where relevant. Given the study goals, the subsequent presentation of results and the discussion primarily emphasize the main and interactive effect of *Intervention*.

## 5.4 Results

Summary statistics and ANOVA results for each dependent measure are presented in Appendix F (Tables 5.1 and 5.2, respectively).

### 5.4.1 Local dynamic stability.

Compared to the Control condition, using the Laevo™ significantly increased $\lambda_{\text{max-s}}$ for the pelvis, by 8.3% (Table 5.1 – Appendix F); though not significant, there was also a 2.9%

increase using the BackX™. There were no significant effects of *Intervention* on $\lambda_{\text{max-l}}$. *Task Condition* main effects were significant for trunk and pelvis $\lambda_{\text{max-s}}$; both values were larger in the Asy_Knee condition.

### 5.4.2 Orbital stability.

$FM_{\text{max}}$ was <1 for all combinations of *Intervention* and *Task Condition* across all Poincare sections (Table 5.1 – Appendix F). Significant *Intervention* main effects were found for thorax and pelvis $FM_{\text{max}}$ at the 25, 50, and 100% sections, in which using the Laevo™ increased $FM_{\text{max}}$ in both the thorax and pelvis (Figure 5.1). *Task Condition* significantly affected $FM_{\text{max}}$ for the pelvis at the 25, 50, and 75% sections, which was significantly larger for the Sym_Ground condition. $FM_{\text{max}}$ for the thorax was significantly higher among males, by 6.1%.

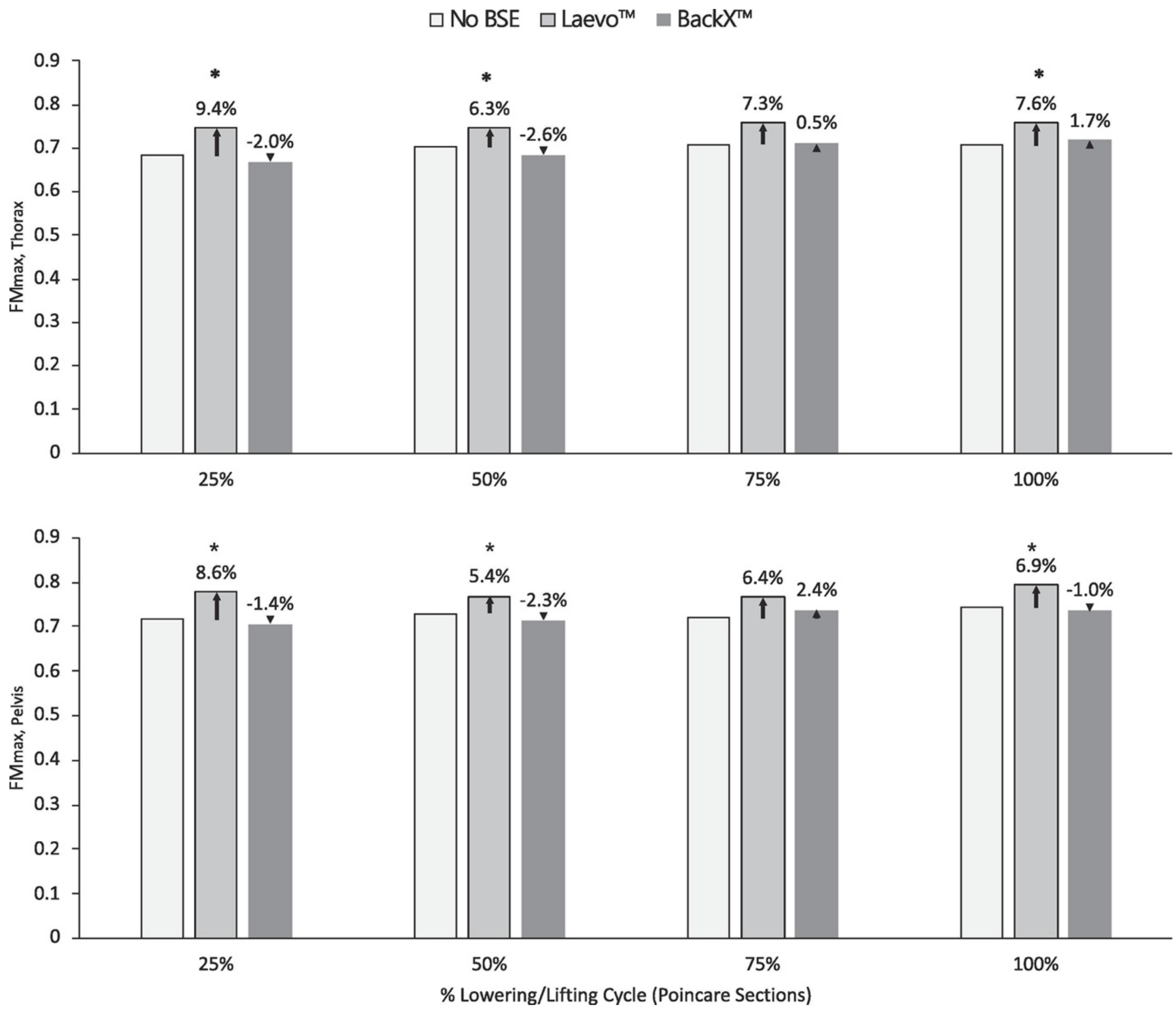


Figure 5.1. *Intervention* effects on $FM_{max}$ at defined Poincare sections for the thorax (top) and pelvis (bottom). Note that the symbol * denotes a significant difference from the Control condition (i.e., no BSE), and arrows and percentage values indicate changes from the Control condition.

### 5.4.3 Continuous relative phase (CRP).

An initial analysis of the CRP measures indicated effects of *Intervention* on MARP and DP that approached significance ($p \simeq 0.06$), with either BSE increasing MARP and DP values by 12.6 and 9.5% respectively using Laevo™, and by 12.6 and 10.0% using BackX™. Residual analysis, however, suggested the presence of a potential outlier, elimination of which resulted in a significant effect of *Intervention* on MARP ($p = 0.02$). MARP and DP were also significantly affected by the *Gender* × *Task Condition* interaction, with females exhibiting larger values only in the Asy_Knee condition.

## 5.5 Discussion

We found that using a BSE can affect trunk and pelvis dynamics during repetitive lifting, though these effects appeared specific to the BSE design and stability measure. Using a BSE did not affect local dynamic stability (short or long-term Lyapunov exponents) of the trunk, yet using the Laevo™ increased the short-term maximum Lyapunov exponent of the pelvis. Using the Laevo™ also reduced orbital stability of both the trunk and pelvis (i.e., increased Floquet multiplier values). Furthermore, the trunk-pelvis coordination pattern became more out-of-phase and more variable using either BSE tested here.

### 5.5.1 Local dynamic stability

BSE use did not affect trunk local dynamic stability over either a short or long-term scale. This finding suggests that the load re-distribution via the chest pad/interface of the BSEs, which is a main source of discomfort in the chest region (Bosch et al., 2016; Madinei et al., 2020a), did not affect trunk local dynamic stability. Our results are in partial conflict with those of Graham et al. (2011), who found that using a personal lift-assistive device (PLAD) had no effect on $\lambda_{\mathrm{max-s}}$ of the trunk, yet led to a negative $\lambda_{\mathrm{max-l}}$. Graham et al. (2011) suggested that trunk dynamics were stable using the PLAD (i.e., attracted to a stable attractor, indicated by a negative exponent value) over 10 repetitive lifting cycles, possibly because of superior corrections or adjustments of the trunk to local perturbations. However, the reported exponent values were quite small [(-6.4E-04, 2.4E-03) vs. (9.6E-06, 5E-04) here], and it might be more reasonable to conclude that long-term trunk dynamics was conservative (i.e., $\lambda_{\max-l} \simeq 0$; steady-state) during repetitive lifting with the use of a BSE, as was found here. Using the Laevo™ further increased $\lambda_{\max-s}$ for the pelvis, though the magnitude of this increase was rather small [e.g., 0.24 (0.04) vs. 0.22 (0.04) with no BSE in the Sym_Knee condition]. The relatively low-profile and minimal waist design approach used in Laevo™ may not provide a sufficiently stable connection to the waist region, potentially causing friction or perturbations to the pelvis dynamics during repetitive lifting over a short-term time scale.

Additionally, $\lambda_{\max-s}$ for the trunk and pelvis depended on *Task Condition*, with higher values found in the Asy_Knee condition. This may not be surprising, though, given the additional burden placed on the neuromuscular system to control movements in an asymmetric task. For example, we earlier found higher abdominal muscle co-activation in this condition (by up to ~160%), likely as a strategy to maintain stability (Madinei et al., 2020a). Our current results, however, differ from earlier findings of a decreased $\lambda_{\max-s}$ for the trunk during asymmetric trunk bending or lifting (Granata & England, 2006; Lee & Nussbaum, 2013; Asgari et al., 2020). However, diverse protocols have been used, making direct comparisons challenging.

### 5.5.2 Orbital stability

Using the Laevo™ also caused higher Floquet multipliers at all Poincare sections examined (Figure 5.1). $FM_{\max}$ values of the pelvis and the trunk both increased, respectively by up to 8.6 and 9.4%, compared to the control condition. These increases suggest that using the Laevo™ could have compromised the ability of the neuromuscular system to achieve steady-steady motions of the trunk and the pelvis during repetitive lifting, and which may indicate poorer stability in response to small (local) perturbations. $FM_{\max}$ values, however, were <1 in all experimental conditions, suggesting that orbital stability was maintained during the repetitive lifting trials (Hurmuzlu & Basdogan, 1994). Thus, the increase in $FM_{max}$ using the Laevo™ might suggest that participants became intrinsically less stable in response to local perturbations. Though it is unclear what aspects of Laevo™ design contributed to this, one possible aspect is the design of the physical interface of Laevo™ with the wearer's body segments. For example, the interface may not provide a stable connection to body segments and interaction forces at the interface may not be distributed effectively over the body.

*Task Condition* had a substantial effect on pelvis $FM_{max}$ across all Poincare sections, with Sym_Ground exhibiting ~10% higher values higher than the other conditions. We found earlier (Madinei et al., 2020a) that participants often had higher trunk flexion/extension velocities in the Sym_Ground condition, which reduces the reaction time required for the neuromuscular system to attain desired objectives for a particular movement direction. This increase in $FM_{max}$ is consistent with earlier work, suggesting speed-control tradeoffs in human movements with temporal and spatial constraints (Granata & England, 2006; Asgari et al., 2015; Asgari et al., 2020).

### 5.5.3 Continuous relative phase

MARP was higher using either BSE, by ~15%, and DP values were also ~11% higher (albeit not significant). Higher MARP values indicate that the coordination pattern became more out-of-phase (asynchronous) with BSE use. Participants may not have fully adapted to external assistance from a BSE, and were thereby unable to consistently control trunk and pelvis movements. Our results, however, contrast with earlier evidence that PLAD use caused a more in-phase (synchronous) lumbar-hip coordination during repetitive lifting (Agnew & Stevenson, 2008). The latter authors suggested that a more synchronous pattern reflects a change in lifting strategies to benefit from PLAD assistance. The PLAD has elastic elements extending from the shoulder and the knee to the lever arm (dorsally projected from the backside of the waist belt), and these elements augment the lumbar extensor and knee flexor musculatures. In contrast, torque generation mechanisms in Laevo™ and BackX™ are located laterally to the waist belt, and these generate external torques based on the included angle between the trunk and thigh. This difference in design could account for the observed change to a more asynchronous coordination when either of the current BSEs was used. Interestingly, participants here also exhibited a less consistent coordination pattern with BSE use, as indicated by higher DP values, which may again be a consequence of participants still adapting to BSE use. Further, earlier studies have reported that pain can affect movement coordination (Lamoth et al., 2006; Seay et al., 2011; Mokhtarinia et al., 2016), and some have reported an increase in trunk coordination variability among those with low back pain during lifting (Mokhtarinia et al., 2016; Pranata et al., 2018). When using a BSE, discomfort at body regions interfacing the BSE was typically reported (Madinei et al., 2020a) and it is likely that such discomfort caused an increase in DP. Further investigation is clearly needed to understand the effects of BSE-induced coordination patterns on worker health in a long-term period.

Finally, females had larger CRP measures in the Asy_Knee condition (by 33.7%). Less synchronous and more variable coordination among females might be related to anthropometry, as we found earlier that females used more hip flexion and less trunk flexion in the Asy_Knee condition (Madinei et al., 2020a). Gender differences in lumbo-pelvic rhythm and stability measures have also been reported (Hall & Lysell, 1995; Granata & Orishimo, 2001). Our findings are generally consistent with those of Mokhtarinia et al. (2016), who found increased DP during asymmetric repetitive bending, though only male participants were tested and no external load was included.

## 5.6 Limitations

A few limitations of the present study need to be noted. First, the paradigms used to evaluate dynamic stability and coordination are only valid for systems that are not externally perturbed, and therefore the derived stability levels should be interpreted cautiously. While the proposed metrics offer indirect evidence regarding the degree of stability, the relationship between local and global stability is yet to be established (van Emmerik et al., 2016). Second, we obtained outcome measures from 40 lowering/lifting cycles, yet there is no consensus on the minimum number of cycles required to accurately estimate trunk dynamic stability. However, 30 movement cycles appear to be sufficient to reach an acceptable level of precision for measures of trunk dynamic stability (Dupeyron et al., 2013). Third, participants were familiarized with each BSE and practiced the lifting tasks only during an initial training session. Whether this training was sufficient for participants to acclimate to and benefit fully from the BSEs remains unknown. Fourth, participants selected preferred support levels for each BSE and lifting condition. The range of selected levels might have led to differences in device stiffness across participants or between conditions, though such self-selection was considered practically relevant. Finally, we focused here on relatively short-term effects of different BSEs (i.e., 4 min of repetitive lowering/lifting), and it is unclear if the BSE effects reported here can be generalized to more prolonged and/or frequent use of a BSE.

## 5.7 Conclusions

Our results provide new evidence of potential "unintended" consequences of BSE use on neuromuscular behaviors of the trunk and pelvis when using passive BSEs during repetitive lifting. Specifically, short-term use of a passive BSE can adversely influence trunk dynamic stability and coordination patterns. This information can be useful when considering BSE adoption, since compromised trunk neuromuscular control could raise safety concerns for future BSE use. More research, however, will be needed to assess the generalizability of different BSE design approaches in terms of unintended short-term and long-term effects on the neuromuscular control of the spine.

## 5.8 Conflict of interest statement

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

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Appendix F

Detailed methods for calculating local dynamic stability, orbital stability, and continuous relative phase

***Local dynamic stability.***

Time-delay embedding was utilized to reconstruct multidimensional state spaces from each original time series and its time delayed copies, using:

$$X(t) = [(\varphi_1, \varphi_2, \varphi_3, \dot{\varphi}_1, \dot{\varphi}_2, \dot{\varphi}_3)_{(t)},\ \ (\varphi_1, \varphi_2, \varphi_3, \dot{\varphi}_1, \dot{\varphi}_2, \dot{\varphi}_3)_{(t+T_d)}] \quad (1)$$

where *X(t)* represents the 12D state space; $\varphi_1$, $\varphi_2$, and $\varphi_3$ represent 3D orientations; and $\dot{\varphi}_1, \dot{\varphi}_2,$ and $\dot{\varphi}_3$ represent 3D angular velocities. $T_d$ is the time delay, and was set to be 10% of the mean number of samples per cycle across all trials (Granata & England, 2006; Graham et al., 2014; Bourdon et al., 2019). $\lambda_{\max}$ was calculated from the distance, $d_j(i)$, between nearest neighbors in the reconstructed state-space, $X(t)$. Specifically, $\lambda_{\max}$ was approximated as the slope of a least-squares fit created using:

$$y(i) = \frac{1}{\Delta t}\langle ln d_j(i)\rangle \quad (2)$$

where $\langle ln d_j(i)\rangle$ represents the mean of the natural logarithm of the distance, $d_j(i)$, for initially close neighbors, *j*, throughout short-term and long-term time steps (Rosenstein et al., 1993). Short-term LyE ($\lambda_{\max-\mathrm{s}}$) and long-term LyE ($\lambda_{\max-\mathrm{l}}$) were computed to characterize local dynamic stability over short- and longer-time periods, respectively. Values of $\lambda_{\max-s}$ and $\lambda_{\max-l}$ were determined as the respective slopes of lines fitted over 0–0.5 and 4–10 cycles (Dingwell & Cusumano, 2000; Dingwell & Marin, 2006; England & Granata, 2007).

***Orbital stability.***

Floquet theory assumes that the $(k+1)^{th}$ state of a system is a function of the $k^{th}$ state:

$$X_{k+1} = F(X_k) \qquad (3)$$

where $k$ is the lifting cycle (Hurmuzlu & Basdogan, 1994). When the system attains dynamic equilibrium, the limit cycle trajectory – considered as the mean across all cycles within the state space (i.e., the equilibrium state) – will cross the Poincare section at a single point $X^*$ that satisfies $X^* = F(X^*)$. A linearized approximation of the Poincare map function was used to calculate the distance between the states $X_k$ and $X_{k+1}$ from the single point $X^*$ as the system evolves (Hilborn, 2000):

$$[X_{k+1} - X^*] \cong J(X^*)[X_k - X^*] \qquad (4)$$

where $J(X^*)$ is an $n \times n$ matrix ($n$ is the state vector dimension) whose eigenvalues define Floquet multipliers (Hilborn, 2000). The maximum eigenvalue of $J(X^*)$ is then defined as the

$FM_{max}$, and if it is <1 the limit cycle is stable. Here, $FM_{max}$ is reported at Poincare sections defined at 25, 50, 75, and 100% of the lowering/lifting cycle, similar to earlier work (Asgari et al., 2015; Asgari et al., 2020).

***Continuous relative phase (CRP).***

Segment angles and velocities were first divided into individual lowering/lifting cycles, and were interpolated to 101 data points corresponding to 0-100% of the cycle. Subsequently, all angular orientations ($\theta$) and velocities ($\omega$) were normalized to [-1, +1] at each time frame to account for differences in amplitude and frequency between segments.

$$\theta_{i,norm} = 2 \times \frac{\theta_i - \min(\theta)}{\max(\theta) - \min(\theta)} - 1 \qquad (5)$$

$$\omega_{i,norm} = 2 \times \frac{\omega_i}{\max[max(\omega), \max(-\omega)]} \qquad (6)$$

Normalized data were then transformed into phase angles, using the four-quadrant inverse tangent function (*atan2*) in MATLAB™:

$$\varphi = \tan^{-1} \frac{\omega_{i,norm}}{\theta_{i,norm}} \qquad (7)$$

A CRP curve was derived for thorax-pelvis coupling from the difference between the phase angles of these two segments. Two measures were then extracted from the CRP curves: 1) mean absolute relative phase (MARP); and 2) deviation phase (DP). MARP was calculated from the mean ensemble curve by averaging the relative phase values over the ensemble CRP curve points using:

$$MARP = \sum_{i=1}^{101} \frac{|\varphi_{relative\ phase}|_i}{101} \qquad (8)$$

MARP values close to 0 represent a more in-phase coordination between the thorax and pelvis, whereas MARP values close to $\pi$ signify an out-of-phase coordination. To quantify coordination variability, DP was calculated by averaging the standard deviations of the ensemble CRP curve (deviation phase curve) at each percent of the task cycle.

$$DP = \sum_{i=1}^{101} \frac{SD_i}{101} \qquad (9)$$

where, $SD_i$ is the between-curve standard deviation at each percent of the cycle. DP values closer to 0 indicate less coordination variability or more coordination stability (Stergiou et al., 2001).

Table 5.1. Summary of outcome measures for each combination of *Intervention* and *Task Condition*. Cell entries are means (SD).

| Body Segment | Outcome Measures | Sym_Ground | | | Sym_Knee | | | Asy_Knee | | |
|---|---|---|---|---|---|---|---|---|---|---|
| | | Control | Laevo™ | BackX™ | Control | Laevo™ | BackX™ | Control | Laevo™ | BackX™ |
| Thorax | $\lambda_{max\_s}$ | 0.21 (0.04) | 0.20 (0.04) | 0.21 (0.04) | 0.23 (0.04) | 0.24 (0.04) | 0.23 (0.04) | 0.20 (0.03) | 0.19 (0.04) | 0.20 (0.04) |
| | $\lambda_{max\_l}$ | 3.17E-04 (1.26E-03) | -1.18E-04 (1.07E-03) | 2.21E-04 (2.01E-03) | -1.06E-04 (1.16E-03) | 1.72E-04 (9.63E-04) | 1.80E-04 (9.54E-04) | 5.08E-04 (1.44E-03) | 3.65E-04 (1.70E-03) | 4.00E-04 (1.28E-03) |
| | $FM_{max, 25\%}$ | 0.71 (0.11) | 0.78 (0.16) | 0.67 (0.12) | 0.68 (0.11) | 0.75 (0.16) | 0.66 (0.11) | 0.66 (0.11) | 0.71 (0.1) | 0.67 (0.10) |
| | $FM_{max, 50\%}$ | 0.71 (0.14) | 0.75 (0.16) | 0.69 (0.13) | 0.71 (0.12) | 0.73 (0.13) | 0.69 (0.09) | 0.69 (0.10) | 0.76 (0.12) | 0.67 (0.11) |
| | $FM_{max, 75\%}$ | 0.75 (0.12) | 0.76 (0.13) | 0.72 (0.1) | 0.70 (0.10) | 0.74 (0.11) | 0.74 (0.14) | 0.67 (0.10) | 0.77 (0.13) | 0.67 (0.09) |
| | $FM_{max\ 100\%}$ | 0.75 (0.12) | 0.80 (0.13) | 0.72 (0.13) | 0.70 (0.10) | 0.71 (0.12) | 0.76 (0.16) | 0.67 (0.11) | 0.76 (0.15) | 0.67 (0.11) |
| Pelvis | $\lambda_{max\_s}$ | 0.21 (0.04) | 0.22 (0.05) | 0.21 (0.04) | 0.22 (0.04) | 0.24 (0.04) | 0.23 (0.04) | 0.19 (0.04) | 0.21 (0.04) | 0.20 (0.04) |
| | $\lambda_{max\_l}$ | 2.25E-04 (1.18E-03) | 4.41E-05 (9.31E-04) | 3.55E-04 (1.34E-03) | -2.69E-05 (7.62E-04) | 9.57E-06 (1.05E-03) | -1.40E-05 (6.57E-04) | 3.53E-04 (1.16E-03) | 2.98E-04 (1.94E-03) | 2.30E-04 (1.16E-03) |
| | $FM_{max, 25\%}$ | 0.77 (0.14) | 0.83 (0.15) | 0.74 (0.12) | 0.67 (0.12) | 0.72 (0.1) | 0.69 (0.11) | 0.71 (0.11) | 0.78 (0.13) | 0.68 (0.12) |
| | $FM_{max, 50\%}$ | 0.77 (0.14) | 0.83 (0.14) | 0.75 (0.13) | 0.69 (0.12) | 0.73 (0.09) | 0.69 (0.11) | 0.73 (0.1) | 0.75 (0.12) | 0.70 (0.11) |
| | $FM_{max, 75\%}$ | 0.76 (0.14) | 0.80 (0.14) | 0.78 (0.12) | 0.70 (0.10) | 0.74 (0.15) | 0.75 (0.13) | 0.71 (0.1) | 0.75 (0.13) | 0.68 (0.12) |
| | $FM_{max\ 100\%}$ | 0.77 (0.15) | 0.83 (0.15) | 0.77 (0.12) | 0.76 (0.12) | 0.75 (0.12) | 0.73 (0.14) | 0.71 (0.09) | 0.8 (0.12) | 0.71 (0.11) |
| Lumbar (Thorax vs. Pelvis) | MARP (rad) | 0.89 (0.20) | 0.97 (0.22) | 1.04 (0.21) | 0.87 (0.22) | 0.94 (0.14) | 0.97 (0.22) | 0.81 (0.29) | 1.04 (0.37) | 0.94 (0.41) |
| | DP (rad) | 1.21 (0.16) | 1.29 (0.21) | 1.36 (0.22) | 1.18 (0.24) | 1.24 (0.15) | 1.29 (0.23) | 1.12 (0.39) | 1.3 (0.42) | 1.20 (0.41) |

Table 5.2. Summary of ANOVA results regarding the main and interaction effects of *Gender*, *Intervention*, and *Task Condition* on $\lambda_{max\text{-}s}$, $\lambda_{max\text{-}l}$, FMmax, MARP, and DP. Cell entries are *F* values (*p* value, $\eta^2$).

| Body Part | Dependent Variables | Gender (G) | Intervention (I) | Task Condition (TC) | G × I | I × TC | G × TC | G × I × TC |
|---|---|---|---|---|---|---|---|---|
| Thorax | $\lambda_{max\_s}$ | 2.71 (0.119, 0.095) | 1.21 (0.301, 0.003) | **60.44 (<.0001, 0.154)** | 1.15 (0.321, 0.003) | 1.32 (0.265, 0.007) | 1.27 (0.285, 0.003) | 0.89 (0.475, 0.005) |
| | $\lambda_{max\_l}$ | 1.35 (0.262, 0.005) | 0.32 (0.727, 0.004) | 0.91 (0.405, 0.012) | 0.17 (0.848, 0.002) | 0.18 (0.946, 0.005) | 0.52 (0.597, 0.007) | 1.78 (0.138, 0.045) |
| | $FM_{max,\ 25\%}$ | 0.08 (0.775, 0.001) | **7.87 (0.001, 0.084)** | 1.44 (0.241, 0.015) | 0.001 (0.999, <.001) | 0.63 (0.641, 0.014) | 0.96 (0.384, 0.01) | 1.88 (0.118, 0.04) |
| | $FM_{max,\ 50\%}$ | 0.29 (0.595, 0.002) | **3.82 (0.025, 0.046)** | 0.14 (0.868, 0.002) | 0.14 (0.869, 0.002) | 0.29 (0.882, 0.007) | 2.65 (0.075, 0.032) | 1.74 (0.146, 0.042) |
| | $FM_{max,\ 75\%}$ | **7.21 (0.016, 0.033)** | 2.88 (0.06, 0.034) | 1.25 (0.291, 0.015) | 0.02 (0.981, <.001) | 1.82 (0.129, 0.044) | 0.49 (0.611, 0.006) | 1.68 (0.16, 0.04) |
| | $FM_{max\ 100\%}$ | 1.04 (0.324, 0.009) | **3.13 (0.047, 0.036)** | 2.35 (0.1, 0.027) | 0.9 (0.41, 0.01) | 1.37 (0.247, 0.032) | 2.23 (0.112, 0.026) | 0.47 (0.758, 0.011) |
| Pelvis | $\lambda_{max\_s}$ | 3.42 (0.083, 0.117) | **8.36 (0.0004, 0.03)** | **16.94 (<.0001, 0.06)** | 0.84 (0.435, 0.003) | 0.6 (0.661, 0.004) | 0.04 (0.957, <.001) | 1.75 (0.143, 0.012) |
| | $\lambda_{max\_l}$ | 0.22 (0.648, 0.001) | 0.21 (0.813, 0.003) | 0.87 (0.42, 0.011) | 0.02 (0.985, <.001) | 0.11 (0.979, 0.003) | 0.43 (0.649, 0.006) | 1.02 (0.401, 0.026) |
| | $FM_{max,\ 25\%}$ | 0.1 (0.76, 0.001) | **5.81 (0.004, 0.061)** | **7.31 (0.001, 0.077)** | 0.15 (0.858, 0.002) | 0.39 (0.812, 0.008) | 1.53 (0.222, 0.016) | 1.93 (0.109, 0.041) |
| | $FM_{max,\ 50\%}$ | 0.38 (0.545, 0.002) | **3.37 (0.038, 0.04)** | **5.98 (0.003, 0.071)** | 1.16 (0.318, 0.014) | 0.12 (0.974, 0.003) | 0.13 (0.88, 0.002) | 1.67 (0.162, 0.04) |
| | $FM_{max,\ 75\%}$ | 0.29 (0.599, 0.002) | 3.05 (0.051, 0.036) | **3.39 (0.037, 0.04)** | 1.08 (0.342, 0.013) | 0.75 (0.562, 0.018) | 0.61 (0.547, 0.007) | 0.66 (0.618, 0.016) |
| | $FM_{max\ 100\%}$ | 2.19 (0.159, 0.011) | **4 (0.021, 0.047)** | 2.12 (0.124, 0.025) | 1.47 (0.234, 0.017) | 1.04 (0.391, 0.024) | 1.47 (0.235, 0.017) | 1.65 (0.165, 0.039) |
| Lumbar (Thorax vs. Pelvis) | MARP (rad) | 0.57 (0.462, 0.007) | **4.04 (0.020, 0.043)** | 0.6 (0.5505, 0.006) | 0.23 (0.797, 0.002) | 0.72 (0.581, 0.015) | **6.21 (0.003, 0.066)** | 0.31 (0.872, 0.007) |
| | DP (rad) | 0.23 (0.64, 0.003) | 2.71 (0.071, 0.03) | 1.28 (0.2821, 0.014) | 0.3 (0.743, 0.003) | 0.45 (0.771, 0.01) | **6.15 (0.003, 0.068)** | 0.38 (0.823, 0.008) |

# 6 Chapter 6. Conclusions

Occupational tasks involving repetitive lifting/lowering can be challenging to eliminate or modify in practice, and alternative interventions such as assistive devices are promising. Back-support exoskeletons (BSEs) are a rapidly-emerging technology that has clear potential for mitigating the risk of LBP, by reducing muscular exertion and spine loading. Passive BSEs, requiring no actuators or power supply, are of specific interest, due to their cost-efficiency, ease of implementation, and predominance in the commercial market. Understanding both the "intended" and "unintended" consequences of BSE use on physical risk factors for LBP, however, is an essential prerequisite for the safe and effective implementation of this technology in actual workplaces. This dissertation aimed to determine diverse impacts of different BSE designs on spine biomechanics using four experiments, each addressing different aspects of intended and unintended consequences of BSE use.

Specifically, a systematic, comparative evaluation of two BSE designs was conducted during symmetric and asymmetric repetitive lifting using a broad set of biomechanical, physiological, and usability measures (Chapter 2). In a subsequent analysis, the assistive torque profiles of two BSE designs were empirically measured, using a computerized dynamometer (Chapter 3). In the following study, the impacts of BSE use on compressive and shear forces at the lower lumbar level were determined using an optimization-based model (Chapter 4). Finally, trunk dynamic stability and movement coordination were analyzed empirically during exoskeleton use, via an approach based on dynamical systems theory (Chapter 5). Overall, the findings of these studies demonstrated that the efficacy of BSEs can vary substantially between BSE designs and the specific tasks for which a BSE is used. Important safety features of the exoskeletons were also identified, providing insights on the BSE performance boundaries. For example, the BSEs tested were found overall to be more effective and safer in tasks closer to the mid-sagittal plane and with moderate degrees of trunk flexion.

Findings from the study on the effects of two BSE designs on physical demands (metabolic and physiologic) indicated significant reductions in these outcome measures (vs. unassisted control), though the magnitude of such beneficial effects varied between exoskeleton designs and task conditions. While we observed reductions in the activity of agonist muscles, there were slight (and non-significant) increases in antagonist muscles. For a given BSE and simulated task, there were inconsistency and task-dependency in the "directions" of different effects (e.g., reduced muscle activity levels of specific muscle groups with increased or unchanged metabolic demands). Such diverging evidence would likely be of particular interest for those in industry considering/evaluating exoskeleton adoption. While the overall subjective feedback and usability ratings were positive towards both BSEs, interference of the devices with body movement was also evident, especially in extreme postures that adversely affected perceived discomfort and pressure for the users. Further, important gender-specific differences were apparent regarding the BSE fit and effectiveness due to the anthropometric differences and user preferences.

Understanding the mechanical behavior of the torque-generating mechanisms embedded in passive exoskeletons is essential to determine the efficacy of these devices in reducing the physical loads in manual material handling tasks. In the second study, we introduced a novel approach to quantify the assistive torque profiles of two passive back-support exoskeletons (BSEs) using a computerized dynamometer. The feasibility of this approach was examined for both human subjects and a mannequin. Clear differences in assistive torque magnitudes were evident between the BSEs, and both devices generated more assistive torques during flexion than extension. The assistive torques obtained from the human subjects were often within similar ranges to those from the mannequin, though values were more comparable over a narrow range of flexion/extension angles. Characterizing exoskeleton assistive torque profiles can help in better understanding how to select a torque profile for given task requirements and user anthropometry, and can assist in predicting the potential impacts of exoskeleton use by incorporating measured torque profiles in a musculoskeletal modeling system.

Estimating the loads on the lumbar spine is a critical factor for determining the efficacy of a BSE during the performance of manual material handling tasks. The third study thus examined whether and to what extent use of a passive BSE can reduce the mechanical loads imposed on the lumbar spine using an optimization-based modeling approach. We found significant reductions in spine compression and shear loads when using either BSE, but that the magnitudes of those reductions were task-specific and dependent on the BSE design. Our findings further suggested that empirical measures of muscle activity may not have consistent utility in predicting changes in spine forces resulting from BSE use, likely because the former has only a partial role in contributing to the resultant spine loadings. Practically, this implies that invalid conclusions could be found if investigations rely on surface EMG measures to assess the impacts of BSE on spine loads. These findings further our understanding of the efficacy of BSEs in reducing physical demands on the spine, and thus will help guide the proper selection and application of BSEs during repetitive lifting tasks.

The final study evaluated the impacts of BSE use on trunk dynamic stability and coordination patterns during repetitive lifting. By supplying additional mechanical support, wearing a BSE was expected to reduce the contribution of active muscle stiffness to spine stability. This expectation was based on the results in Chapter 2 showing decreased muscle activation when wearing a BSE (via EMG), and evidence of a direct relationship between muscle stiffness and activation level. Our results indicated that BSE use can compromise neuromuscular control of the trunk when wearing a BSE, yet the magnitude of this effect depended on the BSE design. These results suggest that caution should be used in selecting a suitable BSE for use in a given MMH task.

## 6.1 Overall limitations

A few common limitations of this work are worth noting. First, the study sample only included young healthy adults (20–35 yrs), so caution should be taken in generalizing the findings for an older population. Second, participants were familiarized with each BSE and practiced the tasks only during an initial training session. Whether this training was sufficient for participants to benefit fully from the BSEs remains unknown. Third, we focused here on relatively short-term effects of different BSEs (i.e., 4 min of repetitive lowering/lifting), and it is unclear if the BSE effects reported here can be generalized for more prolonged and/or frequent use of a BSE. Fourth, the current lifting tasks were simulations, performed in a controlled laboratory environment, and thus the relevance of our results to actual work settings, especially with suboptimal working conditions (uneven ground surfaces, restricted working space, etc.), warrants further investigation. Fifth, participants selected preferred support levels for each BSE and lifting condition. The range of selected levels might have led to differences in device stiffness across participants or between conditions, though such self-selection was considered practically relevant. Finally, only repetitive lowering and lifting was investigated in this research. Future work is needed to assess the efficacy of BSE use in other manual material handling tasks, such as pushing, pulling, carrying, and holding.

## 6.2 Future Work

This work here provided important new evidence to guide future efforts in assessing the efficacy of back-support exoskeletons in several ways. First, the results highlighted the importance of considering diverse technological approaches, as well as incorporating a wide range of occupational task demands to assess the efficacy of BSE use in manual material handling tasks. Second, we showed that a broad range of evaluative approaches are also useful, and perhaps needed, to assess the range of potential intended and unintended impacts of an exoskeleton on a user. Third, the results suggested the need for future laboratory-based and field studies with larger samples, more diverse tasks, and longer-term evaluations to assess the potential intended (e.g., physical demands, performance) and unintended effects (e.g., safety, fatigue, stability) of BSE use. Fourth, findings from the current work support the development of effective simulation approaches to predict the impacts of exoskeletons in a broad range of occupational tasks. Given the resource-intensive nature of lab- or field-based testing, the use of optimization-based biomechanical model seemed to be promising, especially for deployment in field testing. However, future work is recommended to explore the feasibility of other biomechanical models to reliably quantify changes in mechanical loads on the spine caused by using a BSE, and if such models would be effective for implementation in both laboratory and field testing. Fifth, we identified BSE fit as a potential source of discomfort among users and led to high variability in the outcome measures. As such, we believe there is a clear need to characterize the critical design features of exoskeletons to improve adjustability and ensure inclusion for a broad set of users (e.g., of different anthropometry, genders, ages) as well as allow for use in a wide range of tasks. Sixth, our results characterized the assistive torque profiles of different BSE designs (e.g., torque-angle-velocity relationship), which we believe should be a specific consideration in future investigations. We specifically noted that there are likely benefits to adapting or customizing this profile on a task-specific basis, such that the support is commensurate with the task demands that occur over a range of postures.

Overall, the work here provided important new evidence to guide the effective selection and application of passive BSEs, help avoid unintended/preventable side effects resulting from this technology, and aid in maximizing the benefits of BSE use. Results of the current work further contribute to proactive development of safety and ergonomics guidelines and best practices for the safe and effective implementation of BSE technology in the occupational environments. In addition, evidence gained from this work supports future enhancements of passive BSE design characteristics, for example by tuning support profiles (torque vs. angle).