

Assessing the Effects of Exoskeleton Use on Balance and Postural Stability

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Dissertation submitted to the faculty of the Virginia Polytechnic Institute and State University in
partial fulfillment of the requirements for the degree of

Doctor of Philosophy

In

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August 18, 2021

Blacksburg, Virginia

Keywords: occupational exoskeleton, back-support exoskeleton, workplace fall,
postural control, bipedal stance, unipedal stance, gait performance, gait stability,
walking kinematics, walking kinetics, balance recovery, reactive stepping

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ABSTRACT

There is emerging evidence for the potential of occupational back-support exoskeletons (BSEs) to reduce physical demands, and thereby help control/prevent the risk of overexertion injuries associated with manual material handling. However, it is important to understand whether BSEs also introduce any unintended safety challenges. One potential risk associated with BSE use is increased risk of falls, since their extra weight, rigid structure, and external hip extension torque may increase demands on the postural control system. However, there is currently limited evidence on whether, and to what extent, BSE use alters postural stability and/or fall risk. The primary goal of this work was to understand the effects of exoskeleton use, and quantify the effects of exoskeleton design parameters, on balance and postural stability, with a focus on passive BSEs used for repetitive lifting work. A comprehensive evaluation of BSE use was performed under controlled laboratory conditions, focusing on three classes of human activity that form the basis of maintaining postural balance in diverse real-life scenarios: maintenance of a specified posture, voluntary movement, and reaction to an external perturbation.

The first study demonstrated that during quiet bipedal stance, BSE use increased median frequency and velocity of the center of pressure in the anterior-posterior direction. In the second study on level walking, BSE use caused an increase in gait step width and gait variability, and decrease in the margin of stability. BSE use with high supportive torque led to adapted gait patterns in early-stance phase. Hip range of motion and peak hip flexion velocity also decreased, and participants exhibited different strategies to increase mechanical energy for propelling the leg in late-stance phase: these effects increased with increasing torque applied by the exoskeleton. In the final study, BSE use did not alter the maximal lean angle from which individuals could successfully execute single step balance recovery, following a forward loss of balance. However, several recovery responses were negatively affected by BSE use, including increased reaction time, impeded hip flexion, and reduced margin of stability in the high-torque condition.

This is the first systematical investigation to quantify the effects of passive BSEs with multiple supportive torque levels on balance and postural stability. While exoskeleton effects on static balance were minimal, more substantial changes in gait spatiotemporal parameters, hip joint kinematics, and dynamic margins of stability were observed in the later studies. Our results indicate that postural stability deteriorated with exoskeleton use in dynamic conditions, and provide mechanistic insight into how stability is altered by different exoskeleton design factors such as added mass, restricted range of motion, and external hip extension torque. While our results are suggestive of increased fall risk, especially in the high-torque condition, fall risk in real life is moderated by a complex combination of individual and environmental conditions. Future work should consider more complex, realistic tasks and also include a more diverse sample that is studied under longer exposure durations, to further elucidate these findings. Our characterizations of a wide variety of postural responses as a function of exoskeleton torque settings are expected to contribute to improving both design and practice guidelines to facilitate the safe adoption of BSEs in the workplace.

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GENERAL AUDIENCE ABSTRACT

Occupational back-support exoskeletons (BSEs) – wearable mechanical systems designed to support, augment, and/or assist back extension – are expected to serve as an alternative workplace intervention to control and prevent overexertion injuries related to manual material handling tasks. While recent studies have shown the beneficial effects of BSE use in terms of physical load reduction on the low back, some concerns have also been raised on unexpected or unintended effects of exoskeletons. One potential risk associated with exoskeleton use is increased risk of falls, since a BSE's extra weight, rigid structure, and external hip extension torque are expected to place increased demands on the postural control system. Increase in fall risk is a critical safety concern, as occupational falls are a serious problem in terms of injuries, medical/industrial cost, and lost work time.

However, there exists limited evidence on whether the use of a BSE alters postural stability and/or increases fall risk. Hence, the goal of our study was to quantify the effects of BSE use on postural stability in various conditions related to real-life scenarios, such as standing balance, walking stability and how one would respond to a loss of balance following an external perturbation.

Our results showed that during quiet standing, BSE use slightly increased postural sway. In level walking tasks, BSE use had adverse effects on step length, step width, and dynamic stability. Furthermore, wearing a BSE with high supportive torque led to adapted gait patterns in early-stance phase, whereas participants showed different strategies to increase mechanical energy for propelling the leg in late-stance phase. In the final study investigating single step balance recovery following a forward loss of balance, we found that BSE use negatively affects balance recovery, mainly by impeding hip flexion.

Thus, our work suggests that exoskeleton use can deteriorate balance and/or postural stability in situations of static standing, voluntary walking, and reacting to an external perturbation, thereby potentially leading to an increase in fall risk. These effects may be more pronounced among specific population sub-groups such as older workers, and may also affect individuals more severely under conditions of stress or fatigue. Hence, future studies must include more rigorous testing of BSE use using a variety of challenging and realistic scenarios, and also include more diverse population samples. The findings from this work are expected to contribute to improving design and practice guidelines to facilitate the safe adoption of BSEs in the workplace.

Acknowledgement

I would like to deeply thank my advisor, Dr. Divya Srinivasan, for her kind guidance, patience, and encouragement throughout this journey. The completion of this dissertation would not have been possible without her unwavering support. I would also like to thank my committee members, Dr. Sunwook Kim, Dr. Michael Madigan, and Dr. Maury Nussbaum for their continued mentorship and invaluable feedback.

I would like to especially recognize Youngjae Lee and Willow Lawton for their help in data collection and processing. I am also grateful to Randy Waldron for his time and technical support in experimental setup, and to Dr. Jay Kim for his constructive advice.

I am thankful to the current and former members of the Occupational Ergonomics and Biomechanics Lab. In particular, I would like to thank Saman Madinei, Jessica Aviles, Leigh Allin, and Saad Alabdulkarim for their kind and helpful discussions. Additionally, I would like to express a special gratitude to Sophia Ulman for her friendship from the very first day of graduate school.

My appreciation also extends to Hwanik Ju, Hyeen Kong, Jeeseop Kim, Gibong Son, Moonhyun Lee, and Mincheol Sung for many warm memories in Blacksburg. I would also like to thank Jungwook Park for his encouragement and positive energy.

Lastly, I would like to express my sincere gratitude to my wife Somi Lee and my son Doyoon Park for their love, and to my parents and in-laws for their commitment and faith in my plan. I will always remember my beautiful memories in Blacksburg where I spent 5 years for my PhD studies, where our marriage started, and where our son was born.

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Chapter 1. Introduction

1. 1. Occupational back-support exoskeletons (BSEs)

Manual material handling (MMH) tasks (e.g., heavy or repetitive lifting and carrying) frequently involve forceful exertions, repetitive motions, and non-neutral/extreme postures. Hence, MMH workers are exposed to a substantially high risk of musculoskeletal injuries and disorders in many industry sectors (Dennerlein et al., 2012; Deros et al., 2010; Kuiper et al., 1999). For example, exposures to overexertion and bodily reaction (e.g., bending, reaching) explained ~31% of all lost workday cases in the United States (Bureau of Labor Statistics, 2019). In response to musculoskeletal complaints related to MMH, exoskeletons – wearable devices that augment, enable, assists, or enhance motion, posture, or physical activity (ASTM F3323-20, 2020) – are showing promise as a workplace intervention, as evidence is emerging that this new technology can assist workers in performing their job more efficiently and with less physical effort (de Looze et al., 2016; Nussbaum et al., 2019; Voilqué et al., 2019; Zhu et al., 2021).

Occupational exoskeletons are categorized based on the supported body part(s), of which the most common types are shoulder/arm-support and back-support exoskeletons, assisting overhead work and repetitive lifting/prolonged bending, respectively (de Looze et al., 2016; Zhu et al., 2021). The aim of back-support exoskeletons (BSE) is to contribute to trunk extension by providing assistive torques at the lower back (Toxiri et al., 2019). BSEs (Figure 1-1) can be classified into passive and active systems, based on actuators, controls, and power sources utilized (de Looze et al., 2016; Toxiri et al., 2019). Passive BSEs use spring, damper, and/or elastic materials to store energy during the bending phase of human movement and release it during the lifting phase (e.g., Koopman et al., 2019a; Madinei et al., 2020a). Active BSEs use electric motors and/or hydraulic actuators to generate adaptive force or torque profiles to assist the user's trunk extension (e.g., Toxiri et al., 2018; Zhang and Huang, 2018). At present, passive BSEs are relatively more mature for occupational use due to their lower cost, design maturity, and ease of implementation in the workplace (de Looze et al., 2016; Nussbaum et al., 2019).

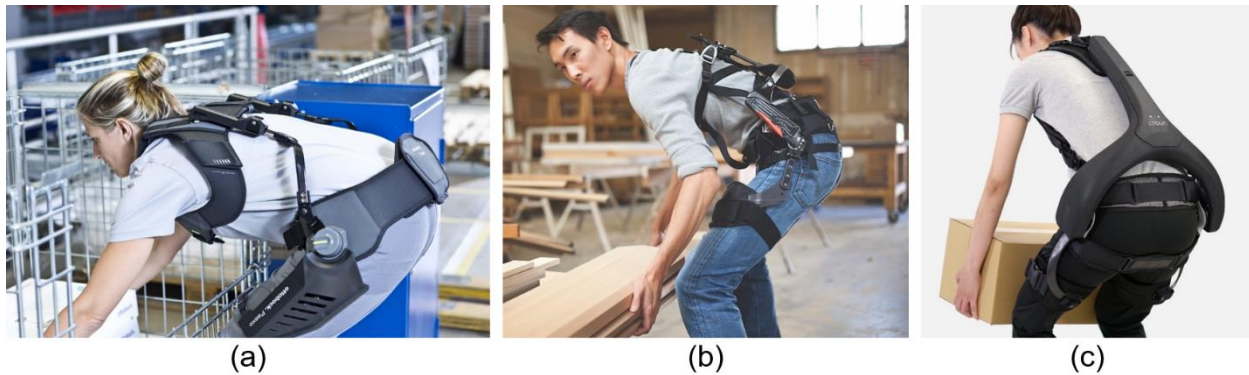


Figure 1-1. Example of commercially-available BSEs: (a) Paexo Back™ (passive, www.paexo.com), (b) SuitX backX™ (passive, www.suitx.com), and (c) AWN™ (active, www.atoun.co.jp).

Previous investigations have shown the beneficial effects of BSE use in terms of physical load reduction on the wearer: use of a BSE was associated with reduction in lumbar muscle activity during repetitive lifting tasks (Abdoli-e et al., 2006; Abdoli-e and Stevenson, 2008; Ko et al., 2018; Lanotte et al., 2018; Madinei et al., 2020a; von Glinski et al., 2019), prolonged periods of lifting (Godwin et al., 2009), forward bending and static holding (Graham et al., 2009; Wei et al., 2020), lowering (Whitfield et al., 2014), stooped posture (Ulrey and Fathallah, 2013), and farm tasks (Thamsuwan et al., 2020). Wearing a BSE also significantly decreased external bending moments on lumbar spine during static holding tasks (Heydari et al., 2013; Koopman et al., 2019a), and metabolic costs of repetitive lifting (Baltrusch et al., 2018a; Baltrusch et al., 2019). Reductions in perceived discomfort/exertion have also been reported with BSE use, during automotive assembly tasks (Graham et al., 2009), manual waste collection (Ziaei et al., 2021), forward bending and static holding tasks (Bosch et al., 2016), and in manual assembly tasks involving static forward and lateral bending postures (Madinei et al., 2020b). Based on this growing body of evidence, it is conceivable that BSEs could serve as an alternative workplace intervention to control and prevent overexertion related to low back disorders.

However, some evidence has also emerged to indicate that current BSEs may have some unintended, negative effects on the wearer. For example, BSE use was associated with significant increase in discomfort and/or perceived pressure on the chest and shoulders due to load redistribution among other body parts (Giustetto et al., 2021; Hensel and Keil, 2019; Qu et al., 2021). In a recent study (Madinei et al., 2020b), it

was reported that wearing a BSE increased trunk muscle activities during manual assembly tasks at below floor level. Furthermore, BSE use increased quadriceps muscle activity during a lifting task, while it reduced back muscle activity (von Glinski et al., 2019). Hwang et al. (2021) found that wearing BSEs reduced back muscle activities but also increased trunk/shoulder flexion angle during patient transfer tasks. BSE use was also associated with decreased trunk velocity and acceleration during lifting tasks (Koopman et al., 2019b) - although decreased trunk velocity and acceleration can lead to decrease in spinal loading (Marras et al., 2000), Koopman and colleagues (Koopman et al., 2019b) noted that such kinematic changes may be reflective of increase in the rigidity of movements, and may hence be undesirable. It is also conceivable that if the exoskeleton limits one's trunk velocity and acceleration, the ability of the wearer to respond to sudden perturbations such as loss of balance may be impacted, creating safety challenges. Moreover, the misalignment between BSE hinges and anatomical rotation axes has also been recognized as a potential problem because the misalignment may impede the correct transfer of assistive torques on the wearer, and can thus produce unintended joint loads and movement patterns (Näf et al., 2018a). A recent study (Huysamen et al., 2018) also pointed out that BSE design should take into account that the skin surface of the lumbar spine involves greater elongation than the skin surface of the thoracic spine during deep lifting, and that such discrepancy may cause potential discomfort and safety challenges by limiting user's voluntary ranges of motion. Thus, current BSEs appear to have some issues in regard to effective reduction of physical strain in situations that the BSE is not designed to support, alteration in range of motion, rigid movement, and misalignment with human anatomy and kinematics.

1. 2. BSE use and fall risk

Among unintended consequences, increase in fall risk due to wearing a BSE may be a critical safety concern. Occupational falls are a serious problem in terms of injuries, lost work time, and medical/industrial costs (Courtney et al., 2001; Webster, 2000). Specifically, data from the Bureau of Labor Statistics (BLS) show that, in 2019, job-related falls resulted in 800 fatalities and 244,000 nonfatal injuries in the United States (Bureau of Labor Statistics, 2020). These falls were associated with an average financial burden of over

\$46,000 per worker in initial medical and worker's compensation costs (National Safety Council, 2019). Two distinct types of falls are recognized in the workplace: falls to the same level that include slip and trip; falls from elevation that include falls on stairs, ladders, or scaffolding (Leamon and Murphy, 1995). BLS has reported that falls on the same level accounted for more injuries with days away from work, but fewer fatal injuries, compared to falls from elevation (Bureau of Labor Statistics, 2020).

The use of BSEs may increase the risk of falls due to BSE characteristics such as added mass, their rigid structure, and the supportive trunk-extension torque, as these characteristics are expected to increase the demands on the postural control system for maintaining postural stability. Among marketed BSEs, the mass of passive BSEs range from 2.8 to 4.5 kg, and the mass of active BSEs ranges from 3.1 to 11 kg (Exoskeleton Report LLC, 2021; Voilqué et al., 2019). Considering mean body mass (BM) for adult males (88.2 kg) and females (75.4 kg) (Fryar et al., 2018), passive BSEs account for roughly 5% and 6% BM for men and women respectively, and active BSEs account for roughly 12.5% and 15% BM for men and women respectively. While there is extensive evidence documenting the effects of external masses (such as backpacks or heavy military armor) on postural stability and fall risk, extrapolating this evidence to conclude what the effects of a BSE will be, are rather complicated because the distribution of the external mass would affect how the center of mass (CoM) of the body-exoskeleton system shifts. Furthermore, exoskeletons also have other characteristics like restricting specific joint ranges of motion and providing supportive torques to assist certain movements, which may interact with the additional mass effect, to produce changes in postural stability.

Wearing a BSE may lead to kinematic constraints and rigid movements due to the BSE's mechanical design (e.g., rigid structure, straps, additional torque and mass, etc.) and this may in turn adversely affect postural control. In a recent review (Kermavnar et al., 2021), the authors reported that BSE use decreases performance during tasks that required increased agility. Similarly, Näf et al. (2018c) suggested that there is a trade-off between the supportive torque provided by the BSE and trunk range of motion. In a recent experiment where individuals were tested when falling freely vs. falling when their body joints were

constrained by a lower-extremity exoskeleton, falling rigidly (i.e., with joints constrained by exoskeleton) also caused more physical impact, which could lead to more severe injuries (e.g., traumatic brain injury), compared to falling with an appropriate buffer action (Qiu et al., 2019). Thus, in addition to changes to the risk of falling, exoskeletons may also increase the risk of sustaining fall-related injuries.

Lastly, most BSEs generate trunk extension torque, to augment physical performance and capacity for lifting tasks. Examples of typical supportive torque profile from passive BSEs (SPEXOR and Laevo™ V2.4) are shown in Figure 1-2. This supportive torque provided by BSEs may, however, negatively affect postural stability during the maintenance of static postures (as is common in industrial work) and during common dynamic tasks such as walking and load carriage. Baltrusch et al. (2018b) has shown that, among twelve different occupational tasks, tasks involving hip flexion were associated with higher perceived difficulty when wearing a BSE. Furthermore, such supportive torque may also hinder critical corrective/reactive movements required to prevent falls if one loses balance, such as taking a recovery step. However, there is currently little evidence on whether, when, and to what extent, the use of a BSE will alter the risk of falls in industrial work.

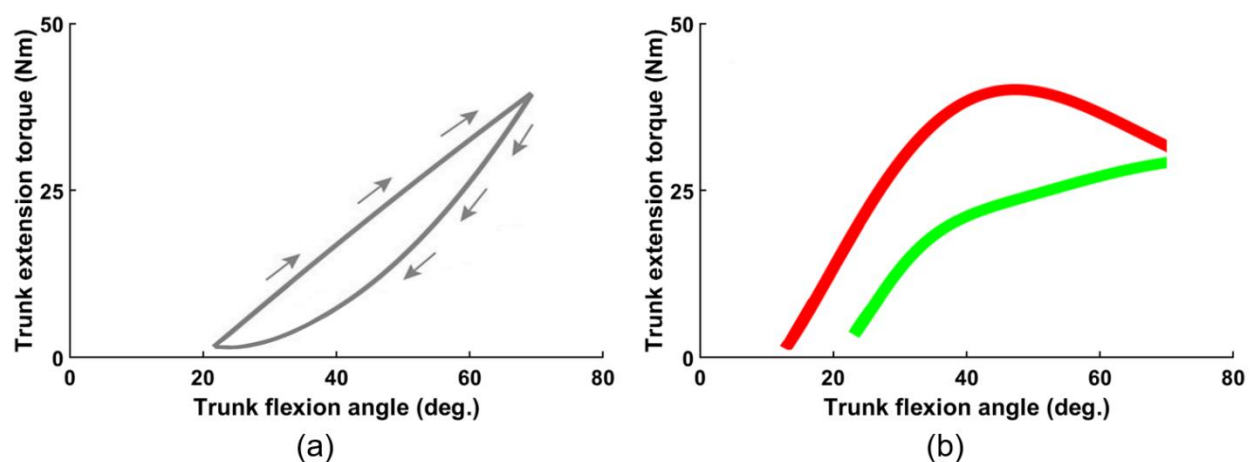


Figure 1-2. Examples of angle-torque relationship of passive BSEs: (a) SPEXOR torque profile during trunk flexion and extension. This figure is reproduced from Näf et al. (2018b); (b) Laevo™ torque profile for supportive torque-high (red) and -low (green) during trunk flexion. This figure is reproduced from Koopman et al. (2019a).

1. 3. Fall risk assessment

Fall risk factors can be classified as being intrinsic or extrinsic (Karlsson et al., 2013; Tinetti et al., 1988). Intrinsic risk factors for falls include personal characteristics such as age, gender, musculoskeletal disease, sensory impairments, gait and balance disorders, and history of prior falls. In contrast, extrinsic factors are largely environmental factors that cause slipping, tripping, or loss of balance (e.g., slippery or uneven floors, poor lighting and unstable footwear). In industrial settings, alleviation of environmental risk factors are major concerns for operations managers to prevent employees from sustaining fall injuries (Brown, 1996; Drebit et al., 2010). In the same context, NIOSH guidelines also have emphasized the importance of reducing extrinsic risk factors in the workplace (Afanuh et al., 2012; Hsiao and Stout, 2010). It is currently unclear whether wearing a BSE increases intrinsic or extrinsic risk factors for fall, as exoskeletons may cause changes in postural control mechanisms, which may in turn affect all individuals or only put certain groups of susceptible individuals at greater risk by interacting with specific personal characteristics. However, irrespective of whether the source of fall risk is extrinsic or intrinsic, there is extensive literature showing that fall risk can be predicted through postural stability evaluation in laboratory-based settings (e.g., Maki et al., 1990; Terry et al., 2014; Wojcik et al., 1999). Other studies also reported that postural instability is one of the most common risk factors leading to fall accidents, especially among older adults (Ambrose et al., 2013; Hsiao and Simeonov, 2001).

Postural stability can be defined as the inherent ability of an individual to maintain, achieve, or restore a state of balance and not fall (Pollock et al., 2000). Postural control, to achieve postural stability, requires corrective movements (i.e., anticipatory and reactive postural control) to maintain the projected Center of Mass (CoM) of the human body within the base-of-support (BoS), the area under and between the feet (Winter, 2009). Specific postural control mechanisms have been identified to govern each of the following three broad classes of human activity (Berg et al., 1989; King et al., 1994): 1) the maintenance of a specified posture, such as sitting or standing; 2) initiation and maintenance of voluntary movements; and 3) the reaction to an external perturbation, such as a slip, a trip, or a push. These three activities place different

demands on the postural control system (Pollock et al., 2000): 1) static balance* while sitting/standing requires the maintenance of a moving CoM within a fixed BoS to achieve balance; 2) voluntary movements such as walking require control of moving CoM, while the BoS is also moved (voluntarily, in an expected manner); and 3) reacting to an external perturbation requires the maintenance of a moving CoM to be within the BoS, when there is an unexpected movement to the BoS, in order to restore balance. These three scenarios form the basis of maintaining postural balance in diverse real-life scenarios, including those encountered in the workplace, when wearing a BSE. Hence, our work will quantify postural stability and control in each situation and using appropriate measures, in order to assess the impacts of BSE use on fall risk.

1. 4. Study aims

The primary goal of this work is to understand the effects of exoskeleton use and exoskeleton design parameters on postural stability and control, with a focus on passive BSEs used for repetitive lifting work. A comprehensive and systematic evaluation of passive BSEs was performed under controlled laboratory conditions, focusing on three classes of human activity: 1) the maintenance of a specified posture, 2) voluntary movement, and 3) the reaction to an external perturbation. Specifically, four aims were completed in support of the overarching goal:

Specific Aim 1: Quantify the effects of BSE use on postural balance during quiet stance, and on functional limits of stability.

Hypothesis: BSE use will cause a decrease in postural balance, and changes in postural control performance would be more pronounced during more challenging postural tasks.

Specific Aim 2: Assess the effects of BSE use on gait performance and stability during level walking.

* While mechanical balance (or equilibrium) indicates the state of an object when the resultant force acting on it is zero, human postural balance refers to the ability of a person to not fall Pollock, A.S., Durward, B.R., Rowe, P.J., Paul, J.P., 2000. What is balance? Clinical rehabilitation 14, 402-406..

Hypothesis: Compared to the control condition, wearing a BSE will decrease dynamic postural stability, and modify gait performance by increasing gait kinematic variability.

Specific Aim 3: Investigate the effects of BSE use on lower limb joint kinematics and kinetics during level walking.

Hypothesis: Wearing a BSE will hinder hip joint motion (i.e., decrease in hip range of motion), and lead to an increase in hip flexion torque, due to the external torque generated by the device. Furthermore, the external torque constraints associated with the hip joint will cause synchronized or compensatory changes in knee/ankle kinematics and kinetics.

Specific Aim 4: Determine the extent to which BSE use impairs balance recovery following external perturbations.

Hypothesis: Wearing a BSE will negatively affect reactive movements to restore balance following an external perturbation. In addition, the greater the external hip extension torque generated by the BSE, the greater the deterioration expected in step recovery movements.

This work is unique in being the first to systematically study the effects of BSE use on multiple aspects of postural control including static posture maintenance, initiation and maintenance of voluntary movements, and reaction to external perturbations, thereby contributing to the basic scientific literature on elucidating the association between exoskeleton use and fall risk. Further, by utilizing multiple levels of supportive torque, these studies will pave the way for developing a generalizable relationship between supportive torque profiles and changes in postural stability and balance recovery ability, thereby contributing to future exoskeleton design. Since most commercially-available passive BSEs have similar structures and working principles, the results of this work are expected to be largely applicable to BSEs beyond the models investigated in our studies. Thus, findings from this work will, in the long-term, contribute towards practical guidelines to facilitate the safe, successful, and productive adoption and implementation of occupational exoskeletons.

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Chapter 2. Effects of two passive back-support exoskeletons on postural balance during quiet stance and functional limits of stability[†]

Abstract

While occupational back-support exoskeletons (BSEs) are considered as potential workplace interventions, BSE use may compromise postural control. Thus, we investigated the effects of passive BSEs on postural balance during quiet upright stance and functional limits of stability. Twenty healthy adults completed trials of quiet upright stance with differing levels of difficulty (bipedal and unipedal stance; each with eyes open and closed), and executed maximal voluntary leans. Trials were done while wearing two different BSEs (SuitX™, Laevo™) and in a control (no-BSE) condition. BSE use significantly increased center-of-pressure (COP) median frequency and mean velocity during bipedal stance. In unipedal stance, using the Laevo™ was associated with a significant improvement in postural balance, especially among males, as indicated by smaller COP displacement and sway area, and a longer time to contact the stability boundary. BSE use may affect postural balance, through translation of the human+BSE center-of-mass, restricted motion, and added supportive torques. Furthermore, larger effects of BSEs on postural balance were evident among males. Future work should further investigate the gender-specificity of BSE effects on postural balance and consider the effects of BSEs on dynamic stability.

Keywords: occupational exoskeleton, bipedal stance, unipedal stance, center of pressure, postural control

2. 1. Introduction

Occupational back-support exoskeletons (BSEs) are wearable mechanical systems designed to provide external forces/torques assisting back extension (de Looze et al., 2016; Toxiri et al., 2019). While passive BSEs involve spring, dampers, or elastic materials to generate the back extension forces/torques, active BSEs involve active actuators and associated power supplies to do the same. Those forces/torques are

[†]This study has been published as a journal article: Park, Jang-Ho, et al. "Effects of two passive back-support exoskeletons on postural balance during quiet stance and functional limits of stability." *Journal of Electromyography and Kinesiology* 57 (2021): 102516.

transferred to the wearer via rigid structures and physical attachments that surround or support chest, waist, and/or thigh. Among marketed BSEs, the mass of passive BSEs ranges from 2.2 to 4.5 kg, and the mass of active BSEs ranges from 3.1 to 11 kg (Voilqué et al., 2019). BSEs are expected to serve as an alternative workplace intervention to control and prevent overexertion injuries related to low back disorders. Recent investigations have highlighted the beneficial effects of BSEs use, in terms of reducing physical loads on the low back (e.g., Frost et al., 2009; Koopman et al., 2019a). However, BSEs use may also introduce some unexpected or unintended challenges. These concerns have been related specifically to increased physical strain in situations that the BSE is not designed to support (von Glinski et al., 2019), more rigid movements (Koopman et al., 2019b), decreased range-of-motion (Abdoli-Eramaki et al., 2007), and misalignment with human anatomy and kinematics (Huysamen et al., 2018).

One potential risk associated with using a BSE is decreased postural balance, which can be defined as the ability to maintain or control the projection of the whole body center-of-mass (COM) within the base-of-support (BOS) (Winter, 1995), since the extra weight of the device could apply more strain on the postural control system, and the rigid structure and external hip extension torque of the BSE may hinder corrective postural movements. Furthermore, working scenarios that involve wearing BSEs could include lifting, walking, load carriage, and stair ascent/descent, all of which are more likely to result in a loss of postural balance for the wearer than when using exoskeletons while stationary. As BSE technologies are seeing increasing occupational use (Nussbaum et al., 2019; Toxiri et al., 2019), and since slips, trips and falls continue to be a major problem in workplaces around the world (Bureau of Labor Statistics, 2019), it is critical to understand if and to what extent BSEs adversely affect postural balance.

Consistent effects of on-body external loads and uneven weight distribution on postural control performance have been demonstrated during quiet standing. For example, wearing an external load has been shown to adversely affect postural balance by increasing postural sway (Heller et al., 2009). An asymmetrical weight distribution of such an additional load leads to greater muscle strain and tension (Attwells et al., 2006). Rugelj and Sevšek (2011) and Park et al. (2014) showed that carrying moderate-

heavy loads (i.e., >10 kg) in a backpack, along with body armor with different weight distributions, decreased postural balance. Park et al. (2014) found that the uneven weight distribution (in the anterior-posterior or medio-lateral directions) in military armor increased postural sway for all loads considered (ranging from 9-27 kg). While BSEs, especially passive devices requiring no actuators or power supply, are in the range of 2.2 - 4.5 kg, and hence not as heavy as the loads considered in these prior studies, it is currently unknown how BSE weight and weight-distribution will affect postural control performance.

Increased joint stiffness and reduced joint motion from artificially locking joints have been shown to increase movement rigidity and correlate with increased postural imbalance (Nijhuis et al., 2008; Pavol et al., 2001). In a recent experiment (Qiu et al., 2019), individuals were tested when falling freely vs. falling when their body joints were constrained by a lower-extremity exoskeleton; falling rigidly (i.e., with joints constrained by exoskeleton) caused more physical impact, which could lead to more severe injuries (e.g., traumatic brain injury), compared to falling with an appropriate moderating action. Although the mobility constraints and assistive torque of a BSE occur primarily at the hip joint, these constraints could cause synchronized or compensatory movements in the knee and ankle joints, and in turn influence even basic functional activities such as upright standing (Günther et al., 2009; Hsu et al., 2007). However, there is currently limited empirical evidence on the effects of using BSEs on postural balance. While one report indicated that using an arm-support exoskeleton was associated with an increase in sway velocity (Kim et al., 2018), and another of a lower-body exoskeleton indicated significant reductions in the magnitude of postural sway during quiet standing (Schiffman et al., 2008), it is unclear if previous results obtained using arm and leg devices will transfer to a BSE.

The objective of this study was thus to assess the effects of wearing occupational BSEs on postural balance during quiet upright stance and on the functional limit-of-stability (FLoS). Based on the rationale that the external load and/or rigid structure of a BSE can increase demands on the postural control system, we hypothesized that wearing BSEs will decrease postural balance during quiet standing. We also expected this decrease in postural balance would be more pronounced during more challenging postural tasks.

Traditional linear measures based on center-of-pressure (COP) data (e.g., sway frequency, velocity, displacement, and area) have been widely utilized in assessing postural balance. Metrics quantifying nonlinear COP dynamics, though, reveal additional mechanisms underlying postural control (Collins et al., 1995; Ramdani et al., 2009). Hence, both traditional and nonlinear descriptors of COP time series were used to quantify postural balance. In addition, wearing BSEs may affect a user's perceived balance and thereby affect their voluntary reach envelopes in occupational environments. This aspect was assessed using functional limits of stability (FLoS), since FLoS is a measure of the extent to which individuals are willing to maximally lean in different directions (Holbein and Chaffin, 1997), and this measure is influenced by internal postural control abilities as well as subjective perceptions and environmental factors (Holbein-Jenny et al., 2007; Holbein and Redfern, 1997).

2. 2. Methods

2. 2. 1. Participants

A convenience sample of 20 young and healthy participants completed the study. Participants included 10 males with mean (SD) age of 25.2 (3.8) years, stature of 176.4 (7.4) cm, and body mass of 76.7 (8.8) kg; corresponding values for the 10 females were 27.5 (2.7) years, 166.5 (5.4) cm, and 61.2 (8.6) kg, respectively. This research 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 data collection. All participants had no self-reported injuries or disorders.

2. 2. 2. Exoskeletons

Among commercially-available passive BSEs, two different designs were included (Figure 2-1), which differed in weight, form, and anchor points: Laevo™ V2.5 (en.laevo.nl) and SuitX™ BackX AC (www.suitx.com). Passive devices were of specific interest because, at present, they (vs. active or powered) are relatively more relevant for occupational use, due to their lower cost, design maturity, and ease of implementation (de Looze et al., 2016; Lee et al., 2012). The respective masses of the Laevo™ and SuitX™

were 2.8 and 4.5 kg. Both devices offset external flexion torques at the hip by applying forces in three body regions – thighs, pelvis, and chest. In both devices, resistance is applied at the chest and the legs from a spring joint when the wearer bends forward, generating a torque in parallel to that created by the torso extensor muscles. The SuitX™ has two support modes: *Instant* vs. *Standard*. In the instant mode, external torque is generated immediately with the onset of torso flexion, while in the standard mode external torque is engaged with torso flexion past 30° - 45°. Since supportive torque in the Laevo™ is also engaged immediately when the wearer bends forward, similar to the SuitX™ instant mode, the SuitX™ instant mode was not tested here. In total, four *EXO* conditions were included in this study: NoExo (no exoskeleton, control condition), Laevo_{ON} (Laevo™ – supportive torque on), Laevo_{OFF} (Laevo™ – supportive torque off), and SuitX (SuitX™ standard mode – supportive torque on).

2. 2. 3. Experimental design and procedures

A repeated-measures design was used to investigate the effects of wearing BSEs on postural balance during quiet standing and on FLoS. Each participant completed a single experimental session, during which they completed quiet standing trials and a set of maximum voluntary leans in each of the four different *EXO* conditions. The presentation order of *EXO* conditions was counter-balanced across participants using 4×4 Latin Squares. In each *EXO* condition, the device was first fitted to the participant and they were provided with a familiarization period. Participants completed quiet standing trials in four different conditions that differed in the level of difficulty, specifically standing on both feet (bipedal stance) and on one foot (unipedal stance), each with eyes open and closed. For a given *EXO* condition, bipedal and unipedal conditions were tested sequentially, and the presentation order of eyes open vs. closed conditions was alternated between participants. Each condition was replicated twice, with one minute of rest between trials.

Quiet standing. For bipedal stance, participants were asked to maintain an upright stance, as still as possible, while barefoot, for 60 s, with their feet close together, arms crossed at their chest, and head facing straight ahead (Figure 2-1 (a)). For unipedal stance, participants were asked to maintain postural balance while barefoot, for 20 s, standing on their dominant foot (i.e., the foot used to kick a ball), with the other leg

slightly flexed, the arms $\sim 60^\circ$ abducted, and head facing straight ahead (Figure 2-1 (b)). All quiet standing trials were completed while participants stood on a force platform (AMTI OR6-7-1000, Watertown, MA, USA).

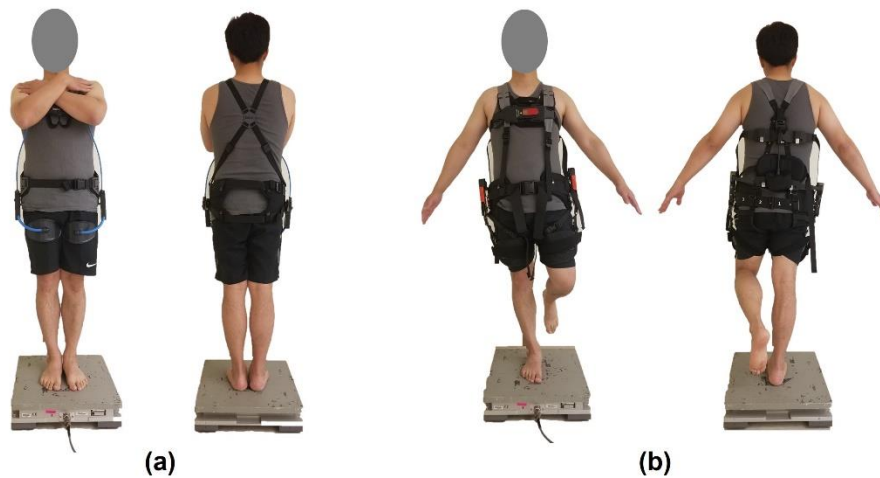


Figure 2-1. Illustrations of the experimental setup for quiet standing trials: (a) bipedal stance wearing the Laevo™ and (b) unipedal stance wearing the SuitX™.

Functional and geometric limits of stability. For bipedal stance, FLoS was measured by having each participant lean in eight different directions (i.e., forward, forward-right, right, backward-right, backward, backward-left, left, forward-left). They were instructed to lean as far as possible and then return to an upright posture by only pivoting about their ankles and specifically without bending their knees or hips (Slobounov et al., 2006). FLoS boundaries for bipedal stance were obtained for each *EXO* and vision condition (i.e., eyes open/closed). For unipedal stance, the geometric limit of stability (GLoS), an outer rectangle that encompassed the participant's standing foot, was measured manually using a ruler (Hertel and Olmsted-Kramer, 2007). Feet (or dominant foot) position remained the same in both the quiet standing and limits of stability trials.

2. 2. 4. Data collection and processing

Triaxial ground-reaction forces and moments in all trials were obtained at 100 Hz from the force platform and using Nexus 2.8 software (Vicon Motion Systems, Oxford, UK). These signals were subsequently low-pass filtered (6th order, zero-phase-lag, Butterworth, 10 Hz cut-off frequency), then used to obtain COP

time series (Winter, 2009). Several traditional COP-based measures were calculated, according to Prieto et al. (1996): 1) median power frequency (MF); 2) mean velocity (MV); 3) root-mean-square distance (RMSD); and 4) elliptical sway area ($Sway_{AREA}$). MF, MV, and RMSD were determined in both the anterior-posterior (AP) and medial-lateral (ML) directions. In general, increases in traditional COP-based measures are interpreted as a deterioration of postural balance (Lin et al., 2008; Prieto et al., 1996). In addition, *Sample entropy (SaEn)* was calculated from the COP time series in both the AP and ML directions. SaEn is a measure of the regularity or complexity in a time series; lower SaEn values indicate a more regular and predictable signal, while higher values indicate more randomness in the signal (Ramdani et al., 2009). Increased SaEn indicates an increase in efficiency or automaticity of postural control, whereas decreased SaEn is usually correlated with greater attention or cognitive involvement in postural control (Donker et al., 2007; Stins et al., 2009). Based on recommendations by Richman and Moorman (2000), $m = 2$ and $r = 0.2 \times SD$ of COP data were used here for determining SaEn. Finally, the *detrended fluctuation analysis (DFA)* exponent (α), which can quantify scaling properties (i.e., presence of long-term correlations) in time series (Delignières et al., 2006; Duarte and Zatsiorsky, 2001), was computed from COP time series in the AP and ML directions. Persistence (i.e., positive correlation) of COP time series is associated with increased muscle activity and decreased postural balance (Collins et al., 1995), while anti-persistence (i.e., negative correlation) has been suggested to reflect a more tightly controlled postural system (Amoud et al., 2007).

Limits of stability and virtual time-to-contact (TimeCont). FLoS was determined by identifying the COP location furthest from the center in the eight directions in which maximum voluntary leans were executed by participants (Figure 2-2 (a)). FLoS was then expressed as a percentage of the BOS area in bipedal standing (Holbein-Jenny et al., 2007). A larger area indicates greater functional limits of stability (Holbein-Jenny et al., 2007; Holbein and Redfern, 1997). GLoS was defined as the BOS for unipedal standing (Figure 2-2 (b)).

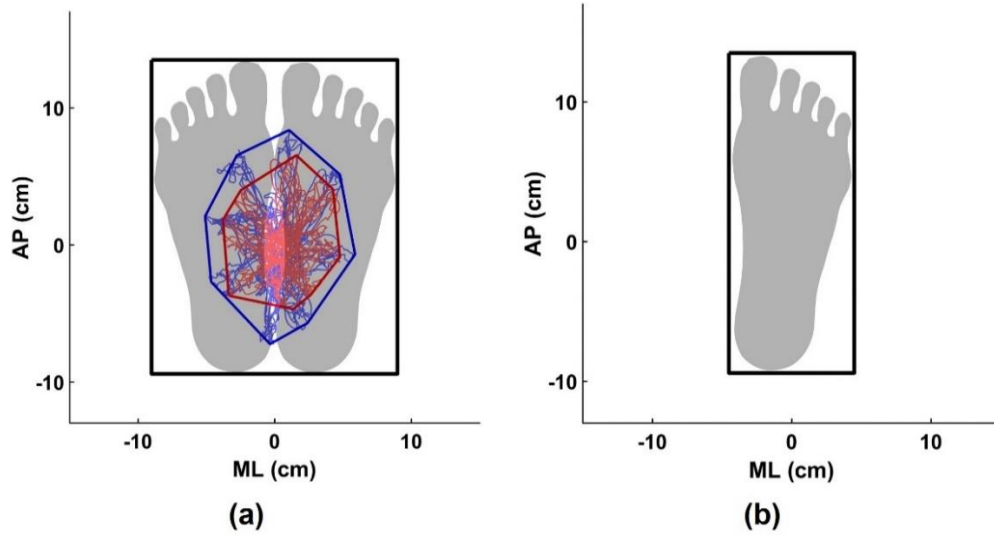


Figure 2-2. Examples of the limits of stability boundary for (a) bipedal standing and (b) unipedal standing; (a) Sample FLoS boundary (octagons) detected based on COP trace (curves) measured during multi-directional maximum leans when wearing an exoskeleton (red) and without wearing an exoskeleton (blue). The outer rectangle indicates the BOS area during bipedal standing; (b) the GLoS boundary is defined as the rectangular area encompassing the participant's dominant foot (i.e., BOS).

Time-to-contact (TimeCont) measures estimate the time it would take for the COP to reach to a limit of stability boundary (Figure 2-3 (a)). To calculate TimeCont, FLoS and GLoS boundaries were used for bipedal and unipedal standing, respectively. For each trial, a projected COP trajectory was obtained using the position, velocity, and acceleration vector; TimeCont was calculated by solving the resulting quadratic equation, to estimate the time at which the projected trajectory would cross the boundary of the FLoS or GLoS (Hertel and Olmsted-Kramer, 2007; Kilby et al., 2014). TimeCont measures here included the overall mean of the TimeCont series ($\text{TimeCont}_{\text{MEAN}}$) and the mean of TimeCont minima ($\text{TimeCont}_{\text{MIN}}$) (Figure 2-3 (b)). A smaller TimeCont indicates greater postural imbalance, as the COP is closer in time to reaching the limit of the stability boundary (Kilby et al., 2014; Slobounov et al., 2008).

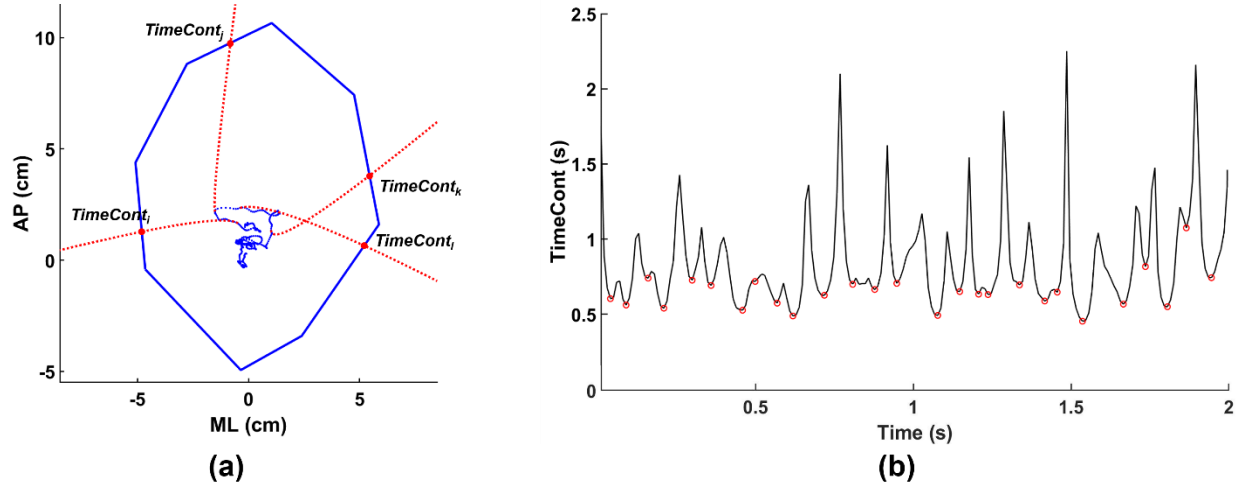


Figure 2-3. (a) Visualization of Time-to-Contact (TimeCont) determination using the COP trace during quiet standing and the FLoS boundary. The blue curve in the center shows a COP trace from 7 s of a bipedal standing trial; the red dashed curves show sample virtual projected COP trajectories at a few time instants; and red circles indicate crossing points with the FLoS boundary; (b) Representative TimeCont series from 2 s of a bipedal standing trial (red circles represent local minima).

2. 2. 5. Statistical analyses

Some outcome measures had significant linear correlations with participant height: MV_{ML} ($r=0.13$, $p=0.001$), MV_{AP} ($r=0.17$, $p<0.001$), $RMSD_{ML}$ ($r=0.24$, $p<0.001$), $RMSD_{AP}$ ($r=0.21$, $p<0.001$), and Sway area ($r=0.26$, $p<0.001$). Hence, these measures were normalized according to the procedure recommended by O'Malley (1996). Briefly, linear regression models relating participant height and each outcome measure were fit, and model-fitted values were used for normalization.

To test our hypothesis, separate repeated-measures analyses of variance (ANOVAs, REML method) were performed on each outcome measure, with *EXO* condition and gender (*GEN*) as within-subject and between-subjects factors, respectively. Significant effects were followed by *post hoc* pairwise comparisons using Tukey's HSD tests, and significant interaction effects were further examined using simple-effects analyses. Partial eta-squared (η_p^2) and Cohen's *d* were used to quantify effect sizes for significant main/interaction effects and pairwise differences, respectively. All statistical analyses were performed using JPM[®] Pro (Version 14, SAS Institute Inc., Cary, NC), and statistical significance was determined when $p < 0.05$.

2. 3. Results

Complete summaries of outcomes in all *EXO* conditions are included in the Appendix (Tables A-1 and A-2). Tables 2-1 and 2-2 respectively summarize ANOVA results, regarding *EXO* and *GEN* effects in the bipedal and unipedal standing sway trials.

Table 2-1. Summary of ANOVA results for the main and interaction effects of *EXO* and *GEN* on outcome measures during bipedal standing. In each cell, *p* values are reported along with effect sizes (η_p^2) in parentheses for each effect, with bold font highlighting significant effects ($p < 0.05$).

	Eyes Open			Eyes Closed		
	<i>EXO</i>	<i>GEN</i>	<i>EXO</i> × <i>GEN</i>	<i>EXO</i>	<i>GEN</i>	<i>EXO</i> × <i>GEN</i>
MF _{ML}	0.041 (0.059)	0.516 (0.020)	0.223 (0.032)	0.606 (0.014)	0.153 (0.106)	0.149 (0.039)
MF _{AP}	0.027 (0.066)	0.316 (0.017)	0.801 (0.007)	0.646 (0.012)	0.965 (0.000)	0.613 (0.013)
MV _{ML}	0.911 (0.004)	0.566 (0.045)	0.454 (0.019)	0.569 (0.015)	0.748 (0.025)	0.256 (0.030)
MV _{AP}	0.102 (0.045)	0.508 (0.089)	0.740 (0.009)	0.038 (0.061)	0.579 (0.149)	0.299 (0.027)
RMSD _{ML}	0.917 (0.004)	0.393 (0.048)	0.700 (0.011)	0.423 (0.021)	0.574 (0.029)	0.159 (0.038)
RMSD _{AP}	0.156 (0.038)	0.911 (0.001)	0.725 (0.010)	0.774 (0.008)	0.566 (0.023)	0.099 (0.046)
Sway _{AREA}	0.822 (0.007)	0.705 (0.014)	0.745 (0.009)	0.853 (0.006)	0.589 (0.026)	0.360 (0.024)
SaEn _{ML}	0.441 (0.020)	0.204 (0.070)	0.995 (0.001)	0.601 (0.014)	0.540 (0.026)	0.813 (0.007)
SaEn _{AP}	0.093 (0.047)	0.958 (0.000)	0.583 (0.014)	0.632 (0.013)	0.785 (0.007)	0.108 (0.044)
DFA _{ML}	0.412 (0.021)	0.392 (0.034)	0.681 (0.011)	0.624 (0.013)	0.857 (0.002)	0.802 (0.007)
DFA _{AP}	0.459 (0.019)	0.942 (0.001)	0.597 (0.014)	0.545 (0.016)	0.813 (0.004)	0.456 (0.019)
FLoS _{AREA}	0.024 (0.068)	0.595 (0.036)	0.030 (0.064)	0.523 (0.017)	0.583 (0.041)	0.026 (0.067)
TimeCont _{MEAN}	0.281 (0.028)	0.427 (0.119)	0.527 (0.016)	0.006 (0.088)	0.205 (0.409)	0.006 (0.089)
TimeCont _{MIN}	0.285 (0.028)	0.444 (0.121)	0.461 (0.019)	0.023 (0.068)	0.205 (0.409)	0.014 (0.076)

Table 2-2. Summary of ANOVA results for the main and interaction effects of *EXO* and *GEN* on outcome measures during **unipedal** standing. In each cell, *p* values are reported along with effect sizes (η_p^2) in parentheses for each effect, with bold font highlighting significant effects ($p < 0.05$).

	Eyes Open			Eyes Closed		
	<i>EXO</i>	<i>GEN</i>	<i>EXO</i> × <i>GEN</i>	<i>EXO</i>	<i>GEN</i>	<i>EXO</i> × <i>GEN</i>
MF _{ML}	0.334 (0.025)	0.480 (0.008)	0.253 (0.030)	0.565 (0.016)	0.341 (0.018)	0.236 (0.033)
MF _{AP}	0.594 (0.014)	0.913 (0.000)	0.772 (0.008)	0.982 (0.001)	0.555 (0.007)	0.168 (0.039)
MV _{ML}	0.673 (0.011)	0.258 (0.088)	0.079 (0.049)	0.108 (0.047)	0.384 (0.058)	0.436 (0.021)
MV _{AP}	0.078 (0.049)	0.231 (0.148)	0.568 (0.015)	0.090 (0.050)	0.564 (0.024)	0.147 (0.041)
RMSD _{ML}	0.547 (0.016)	0.945 (0.000)	0.617 (0.013)	0.364 (0.025)	0.870 (0.001)	0.571 (0.016)
RMSD _{AP}	0.956 (0.002)	0.456 (0.017)	0.964 (0.002)	0.002 (0.108)	0.678 (0.008)	0.001 (0.121)
Sway _{AREA}	0.921 (0.004)	0.608 (0.010)	0.790 (0.008)	0.013 (0.081)	0.703 (0.008)	0.019 (0.075)
SaEn _{ML}	0.126 (0.042)	0.359 (0.057)	0.120 (0.042)	0.190 (0.037)	0.739 (0.006)	0.460 (0.020)
SaEn _{AP}	0.994 (0.001)	0.390 (0.026)	0.833 (0.006)	0.028 (0.069)	0.492 (0.019)	0.050 (0.060)
DFA _{ML}	0.510 (0.017)	0.236 (0.093)	0.825 (0.007)	0.255 (0.031)	0.244 (0.040)	0.935 (0.003)
DFA _{AP}	0.743 (0.009)	0.462 (0.030)	0.230 (0.032)	0.229 (0.033)	0.386 (0.031)	0.128 (0.044)
TimeCont _{MEAN}	0.084 (0.048)	0.398 (0.147)	0.103 (0.045)	0.014 (0.080)	0.250 (0.221)	0.773 (0.009)
TimeCont _{MIN}	0.036 (0.062)	0.376 (0.155)	0.152 (0.039)	0.012 (0.083)	0.249 (0.227)	0.556 (0.016)

2. 3. 1. Bipedal stance with eyes open

EXO had a significant main effect on MF in both the ML and AP directions (Figure 2-4). Specifically, MF_{ML} was significantly higher in the *Laevo*_{ON} condition compared to *Laevo*_{OFF} ($p = 0.031$, $d = 0.49$), and MF_{AP} was significantly higher in the *Laevo*_{ON} compared to the NoExo condition ($p=0.015$, $d=0.58$). Both the main effect of *EXO* and the *EXO*×*GEN* interaction effect were significant for $FLoS_{AREA}$ (Figure 2-4). Although the SuitX condition led to a lower $FLoS_{AREA}$ than the *Laevo*_{OFF} ($p=0.020$, $d=0.40$), simple-effects analyses showed that this difference was significant only among males ($p=0.004$, $d=0.86$).

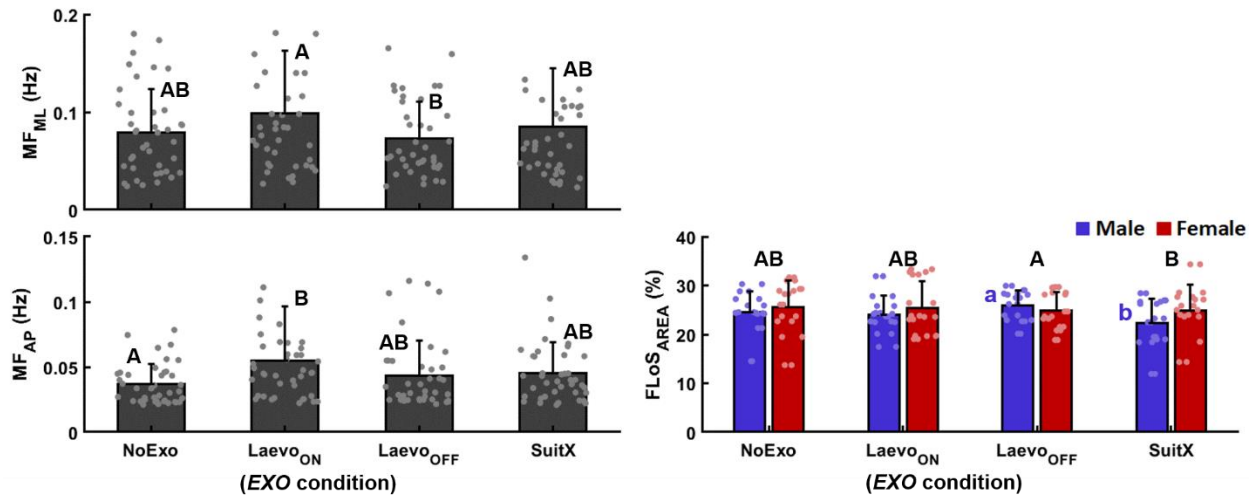


Figure 2-4. Outcome measures in four *EXO* conditions during **bipedal stance with eyes open**. Only measures that showed significant main effects of *EXO* and/or interaction effect of *EXO*×*GEN* from Table 2-1 are shown. Data are represented as group means, with error bars signifying standard deviations. Dots represent individual data for each condition. Upper case letters specify groupings obtained from pairwise comparisons between *EXO* conditions (when the *EXO* main effect was significant). Lower case letters show results from pairwise comparisons between *EXO* conditions within *GEN* (when the *EXO*×*GEN* interaction effect was significant).

2. 3. 2. Bipedal stance with eyes closed

In contrast to the eyes open condition, no effects were significant on COP frequency measures with eyes closed. However, *EXO* had a significant main effect on MV_{AP} , such that MV_{AP} was significantly higher with SuitX use compared to NoExo ($p=0.029$, $d=0.20$, Figure 2-5). $TimeCont_{MEAN}$ and $TimeCont_{MIN}$ were both affected by the *EXO* main and the *EXO*×*GEN* interaction effects. Specifically, both $TimeCont_{MEAN}$ and $TimeCont_{MIN}$ were higher in the *Laevo*_{OFF} conditions, vs. either the *Laevo*_{ON} ($p=0.025$ and $d=0.25$,

$p=0.038$ and $d=0.23$, respectively) or the SuitX ($p=0.008$ and $d=0.25$, $p=0.039$ and $d=0.21$, respectively) conditions. However, simple effects analysis showed that while males in Laevo_{OFF} had significantly higher values than NoExo ($p=0.012$ and $d=0.50$, $p=0.029$ and $d=0.45$, respectively) and SuitX ($p=0.001$ and $d=0.52$, $p=0.032$ and $d=0.46$, respectively) conditions, females did not show any significant differences in either TimeCont_{MEAN} or TimeCont_{MIN} between EXO conditions. Though there was a significant EXO×GEN interaction effect on FLoS_{AREA}, neither males nor females had significant differences between EXO conditions in *post hoc* analyses.

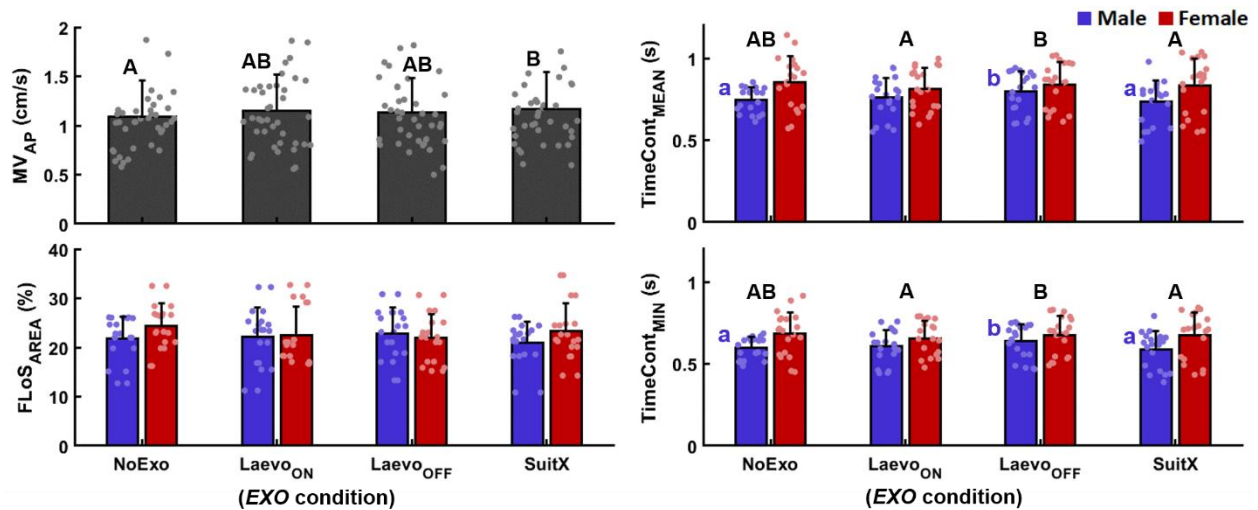


Figure 2-5. Outcome measures in four EXO conditions during **bipedal stance with eyes closed**. Only measures that showed significant main effects of EXO and/or interaction effect of EXO×GEN from Table 2-1 are shown. Data are represented as group means, with error bars signifying standard deviations. Dots represent individual data for each condition. Upper case letters specify groupings obtained from pairwise comparisons between EXO conditions (when the EXO main effect was significant). Lower case letters show results from pairwise comparisons between EXO conditions within GEN (when the EXO×GEN interaction effect was significant).

2. 3. 3. Unipedal stance with eyes open

Only TimeCont_{MIN} differed significantly between EXO conditions in the eyes open condition. TimeCont_{MIN} was significantly higher in the SuitX condition than in Laevo_{ON} ($p=0.037$, $d=0.27$, Figure 2-6).

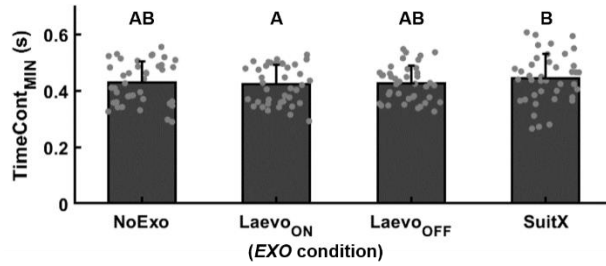


Figure 2-6. Time to contact the geometric base-of-support boundary in four *EXO* conditions during **unipedal stance with eyes open**. This measure showed significant differences with *EXO* (Table 2-2). Data are represented as group means, with error bars signifying standard deviations. Dots represent individual data for each condition. Upper case letters specify groupings obtained from pairwise comparisons between *EXO* conditions.

2. 3. 4. Unipedal stance with eyes closed

In this most challenging condition, more outcome measures showed significant effects than in the other conditions. $SaEn_{AP}$, $TimeCont_{MEAN}$, and $TimeCont_{MIN}$ were each significantly different between *EXO* conditions. $SaEn_{AP}$ was significantly lower in $Laevo_{OFF}$ compared to NoExo ($p=0.048$, $d=0.48$, Figure 2-7). While $TimeCont_{MEAN}$ was significantly higher in $Laevo_{ON}$ compared to NoExo ($p=0.012$, $d=0.37$), $TimeCont_{MIN}$ was significantly higher in both $Laevo_{ON}$ ($p=0.012$, $d=0.37$) and $Laevo_{OFF}$ ($p=0.045$, $d=0.33$, Figure 2-7) compared to NoExo. $RMSD_{AP}$ and $Sway_{AREA}$ differed significantly across *EXO* conditions, and there were significant interaction effects of $EXO \times GEN$. The effects of *EXO* indicated that both $RMSD_{AP}$ and $Sway_{AREA}$ were significantly higher in the $Laevo_{OFF}$ compared to the $Laevo_{ON}$ condition ($p=0.001$ and $d=0.71$, $p=0.008$ and $d=0.60$, respectively). However, simple effects analysis showed that both $RMSD_{AP}$ and $Sway_{AREA}$ were lower among males in the $Laevo_{ON}$ compared to the NoExo condition ($p=0.003$ and $d=0.97$, $p=0.012$ and $d=0.79$, respectively) and compared to the $Laevo_{OFF}$ condition ($p<0.001$ and $d=0.98$, $p=0.007$ and $d=0.77$, respectively). Females showed a significant increase in $RMSD_{AP}$ in the SuitX compared to the NoExo condition ($p=0.001$ and $d=0.76$), but no differences were found in $Sway_{AREA}$ between *EXO* conditions.

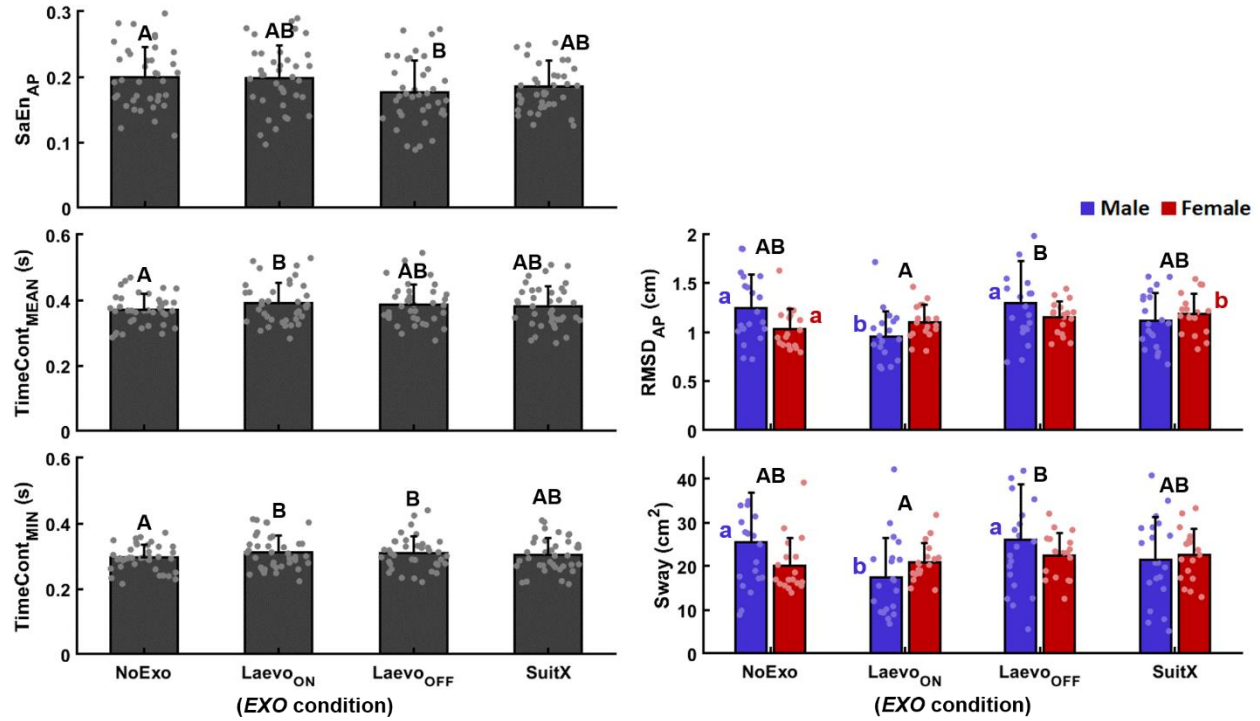


Figure 2-7. Outcome measures in four *EXO* conditions during **unipedal stance with eyes closed**. Only measures that showed significant main effects of *EXO* and/or interaction effect of *EXO*×*GEN* from Table 2-2 are shown. Data are represented as group means, with error bars signifying standard deviations. Dots represent individual data for each condition. Upper case letters specify groupings obtained from pairwise comparisons between *EXO* conditions (when the *EXO* main effect was significant). Lower case letters show results from pairwise comparisons between *EXO* conditions within *GEN* (when the *EXO*×*GEN* interaction effect was significant).

2. 4. Discussion

Overall, our results indicate that wearing BSEs led to a decrease in postural balance during *bipedal standing*. Specifically, wearing the Laevo™ when the supportive torque mechanism was engaged (i.e., in the on-mode) increased the median frequency of the COP series in the AP direction by 50% (in eyes-open condition, Figure 2-4), and wearing the SuitX™ increased mean velocity of the COP series in the AP direction by 7% (in the eyes-closed condition, Figure 2-5). Results from *unipedal standing* imply that wearing the Laevo™ in the on-mode improved postural balance, especially among males, as indicated by smaller COP displacement and sway area, and longer time to contact stability boundary, as compared to the control condition (Figure 2-7). Wearing the SuitX™ led to lower postural balance among females during unipedal standing, as indicated by an increase in COP displacement in the AP direction (Figure 2-7).

2. 4. 1. Effects of alterations in COM location

According to the simplified inverted pendulum model of the upright human posture (Winter, 2009), postural sway increases as the COM is elevated vertically or deviated anterior-posteriorly. Prior work has shown that a vest-type arm support exoskeleton that elevated the body+exoskeleton COM led to an increase in MV_{AP} during quiet standing (Kim et al., 2018), and that a lower limb exoskeleton that lowered the COM reduced postural sway excursions (Schiffman et al., 2008). Based on this evidence, our result – showing an increased MV_{AP} when wearing SuitX™ during bipedal stance – may be explained by COM elevation. This effect is also reasonable to expect, as the SuitX™ mainly sits on the waist/hip region and includes components spanning from the waist over the shoulder, trunk, hip, and thigh regions. The whole body COM position of a human has been estimated to be at ~56% of a person's stature (Virmavirta and Isolehto, 2014). We roughly approximated the COM of the SuitX™ by hanging it in different orientations. Comparing this estimate to the whole body COM, we observed that for a participant whose COM was just above their greater trochanter, the SuitX™ COM were roughly at the participant's waist level. Wearing the SuitX™ can thus be predicted to elevate the combined COM, to somewhere between the greater trochanter and the waist. Furthermore, mass of the SuitX™ are not symmetrically distributed in the sagittal plane. This may explain increase in MV in only AP direction when wearing the SuitX™.

Several previous investigations have found substantial effects of wearing an additional load on postural balance during upright standing (e.g., Heller et al., 2009; Li et al., 2019). Since the mass of the SuitX™ (4.5 kg [6.5% BW on average]) is considerably less than the backpack or double packs used in the previous studies, we believe that the additional weight of the SuitX™ may not have substantially affected postural balance during quiet bipedal standing. This was also supported by the finding that wearing the Laevo™ when the supportive torque mechanism was engaged (i.e., in the on-mode) yielded most of the significant differences in postural balance, and that these differences were not observed when the same Laevo™ device was worn without the supportive torque mechanism engaged (i.e., when off). BSE use was also associated with a beneficial effect on postural balance during unipedal standing. Hence, while COM elevation and

weight distribution may have had some effects on postural control during quiet standing, other factors like restricted motion and supportive torque may be responsible for explaining the observed changes in postural balance.

2. 4. 2. Effects of restricted motion

It was reasonable to expect that body movements are restricted when wearing the Laevo™ or SuitX™, due to straps and rigid structures. Schiffman et al. (2008) reported that wearing a lower extremity exoskeleton decreased FLoS in the ML direction, and that this effect may have resulted from constrained ML movement. Since we found no significant difference in FLoS_{AREA} between the control and any BSE condition, any restricted motion at the trunk and hips from wearing the Laevo™ or SuitX™ may not have been of a magnitude substantial enough to alter voluntary leaning behaviors. Nevertheless, previous work has shown that restricted motion adversely influences dynamic postural control (Grüneberg et al., 2004). Also, reduced joint motion, due to either stiffness or movement rigidity, was positively correlated with postural imbalance (Nijhuis et al., 2008). Hence, future work is needed to determine whether our findings will also be true for dynamic tasks and during situations in which unexpected large or corrective movements are required to achieve balance recovery; as well as how these behaviors are affected by different exoskeleton design features.

2. 4. 3. Effects of supportive torques

Maintaining upright posture requires the effective control of torques at the ankle, knee, and hip with appropriate levels of stiffness (Edwards, 2007). During bipedal stance with eyes open, wearing the Laevo™ in the on mode significantly increased MF_{AP}. Although some interpret an increase in sway frequency as indicative of a deterioration of postural balance (Lin et al., 2008; Prieto et al., 1996), others argue the opposite, using the rationale that an increase in sway frequency reflects an increase in ankle stiffness, which in turn is associated with improved postural control (Warnica et al., 2014; Winter et al., 1998). We speculate that external hip extension torque generated by the Laevo™ increased hip joint stiffness, and that this in turn may have reciprocally modulated ankle stiffness to preserve postural balance (Hsu et al., 2007; Suzuki

et al., 2012), and thereby increased sway frequency (Vette et al., 2017; Warnica et al., 2014). Future studies, though, need to verify whether hip and ankle joint stiffnesses are actually modulated when wearing a BSE.

Wearing the Laevo™ in the on mode showed beneficial effects on postural balance during unipedal stance with eyes closed. Specifically, wearing the Laevo™ in the on mode significantly increased both $\text{TimeCont}_{\text{MEAN}}$ and $\text{TimeCont}_{\text{MIN}}$. These increases may also have resulted from the external torque about the hip/back generated by the Laevo™, since we observed that participants exhibited more trunk movements in this condition to maintain balance, compared to the unipedal stance with eyes open. The Laevo™ provided external torques about the hip, which could help participants to effectively increase and/or control hip stiffness of the dominant (stance) leg and the trunk during unipedal stance. These effects of supportive torque were further supported by our finding that BSE use did not affect any measures in the ML direction. Additional support for the beneficial effect of supportive torque during unipedal stance also stems from the observed decrease in SaEn_{AP} in the Laevo_{OFF} mode as compared to the no-exoskeleton condition. This decreased SaEn_{AP} may suggest less automatic postural control – specifically more postural control effort required to maintain postural stability in unipedal stance with eyes closed, when wearing the Laevo™ in the off mode (Stins et al., 2009) – whereas wearing the SuitX™ or Laevo™ in the on-mode, which provided higher supportive torque compared to Laevo™ off-mode, did not cause a significant difference in SaEn_{AP} . However, since SaEn typically is strongly dependent on the frequency content of the signal, it may also reflect mechanical changes such as increase in stiffness or inertia. Furthermore, even if the additional torque that results when wearing a BSE could be beneficial for more challenging balance conditions (e.g., unipedal stance with eyes closed), whether there are advantageous effects on postural control in dynamic tasks is questionable. For instance, stiffened lower-extremities during dynamic tasks could adversely affect balance recovery motions.

2. 4. 4. Gender differences in exoskeleton effects on postural balance

Our results indicate that BSE effects on postural balance are gender-specific: while males showed differences in outcome measures between exoskeleton conditions during bipedal stance, females did not

exhibit any significant differences (Figure 2-4 and 2-5). Similarly, males had significant improvements in static balance when wearing the Laevo™ in unipedal stance with eyes closed, yet females did not exhibit such a difference. Further, while males did not show any significant differences wearing the SuitX™, females exhibited a significant increase in $RMSD_{AP}$ (Figure 2-7). Although there are several plausible reasons why gender differences were observed in postural balance with BSE use, such as anthropometric differences leading to differences in fit and body alignment of the exoskeleton, the exact reasons for gender differences could not be determined in the current study.

2. 4. 5. Design differences in exoskeleton effects on postural balance

The two BSEs used in this study have distinct design features. The Laevo™ (2.8 kg) is lighter than the SuitX™ (4.5 kg). The Laevo™ transfers the load through pushing against the chest only, while the SuitX™ includes a structural frame that pulls the torso backwards by distributing the pressure to the shoulder straps and chest pad. There are also differences in the major body anchor points for the torque generation mechanism (chest, waist, and thigh for the Laevo™ vs. upper back, waist, and thigh for the SuitX™). Thus, the SuitX™ may have higher effects on restricted body movement than the Laevo™. Furthermore, supportive torque is engaged immediately with the onset of torso flexion in the case of the Laevo™, but engaged only with torso flexion past 30° - 45° in the case of the SuitX™. This may have led to the device-based differences in postural balance observed in unipedal standing. Future work on BSE design parameters may clarify the causal mechanisms underlying changes in postural balance associated with BSE use.

2. 4. 6. Relevance of quiet standing tasks for industry context

Standing is an essential postural activity in various work contexts, regardless of whether or not a worker wears a BSE. It is also probable that those who present impaired postural balance during upright standing are more likely to lose postural balance during voluntary dynamic movements, such as walking and load carriage, or when reacting to an external perturbation, such as a slip and trip. Thus, it is considered that investigating a change in postural control performance during upright quiet standing is an important first step in exploring the effects of wearing BSEs on fall risk, though it would be important to evaluate postural

balance during more complex tasks (e.g., manual material handling in various environments, such as slopes, stairs, and other uneven terrains) in the future.

2. 4. 7. Limitations

This study had some limitations that should be noted. First, participants were all relatively young, and our results may not apply to older age groups. Second, BSE fit/adjustments were not objectively controlled. A BSE was fitted and adjusted following manufacturers' instructions. Thus, there may have been individual differences in BSE fit/adjustments, leading to differential effects on COM or motion restriction. Third, unlike commonly-used postures for assessing static postural balance, participants in this current study were asked to maintain upright standing with a somewhat distinct posture (i.e., both arms crossed for bipedal standing and the arms $\sim 60^\circ$ abducted for unipedal stance), which was done so as to not interfere with the exoskeletons. This posture should not have caused major differences in the current results, since the arm postures were standardized across all *EXO* conditions and thus was likely a systematic effect. Fourth, barefoot standing that is used in this study protocol is somewhat different to real scenarios in the workplace. However, postural responses across different individuals would likely have been confounded by differences in footwear, as wearing shoes is known to improve postural balance by facilitating somatosensory feedback to the foot (Hijmans et al., 2007; Perry et al., 2007; Wilson et al., 2008). Hence, the standard protocol (i.e., barefoot standing) was adopted in this study to control for such potential differences. Fifth, a wide range of COP-based measures were used to assess postural balance during quiet standing. This may have led to an inflation in overall type I error rate. Lastly, we did not investigate the long-term effects of BSE use on postural balance. As such, our results may not be generalizable to long-term users whose postural control could evolve with repeated use of a BSE.

2. 4. 8. Conclusions

Overall, our findings suggest that BSEs can affect postural balance, through alterations in overall COM (body + BSE) location, restricted motion, and added supportive torques. We speculate that both alterations in COM location and supportive torque affected bipedal standing, that restricted motion effects were

relatively minimal, and that the effects of supportive torque were predominant in unipedal standing. Furthermore, the effects of wearing BSEs on postural balance were larger among males. Future studies are needed to understand further why there may be gender-specific effects of exoskeletons on postural balance, how users may react to long-term use and adaptations to exoskeletons, and to investigate the effects of wearing exoskeletons on postural control in dynamic tasks such as walking, obstacle crossing, or recovering from perturbations.

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Appendix A

Table A-1. Mean (SD) of COP measures during **bipedal** standing.

		Eyes Open				Eyes Closed			
		NoExo	LaevoOn	LaevoOff	SuitX	NoExo	LaevoOn	LaevoOff	SuitX
MF _{ML} (Hz)	Male	0.06 (0.03)	0.10 (0.07)	0.07 (0.04)	0.08 (0.06)	0.12 (0.06)	0.14 (0.06)	0.14 (0.08)	0.12 (0.08)
	Female	0.10 (0.05)	0.10 (0.06)	0.07 (0.04)	0.09 (0.06)	0.17 (0.08)	0.15 (0.07)	0.17 (0.08)	0.18 (0.08)
MF _{AP} (Hz)	Male	0.04 (0.02)	0.06 (0.06)	0.05 (0.03)	0.05 (0.03)	0.10 (0.09)	0.10 (0.09)	0.10 (0.08)	0.09 (0.06)
	Female	0.04 (0.01)	0.05 (0.02)	0.04 (0.02)	0.04 (0.02)	0.08 (0.05)	0.11 (0.07)	0.11 (0.09)	0.10 (0.05)
MV _{ML} (cm/s)	Male	0.93 (0.19)	0.94 (0.26)	0.95 (0.25)	0.92 (0.23)	1.40 (0.32)	1.32 (0.40)	1.33 (0.45)	1.34 (0.33)
	Female	0.87 (0.18)	0.86 (0.18)	0.82 (0.15)	0.86 (0.17)	1.17 (0.29)	1.20 (0.30)	1.17 (0.26)	1.24 (0.26)
MV _{AP} (cm/s)	Male	0.81 (0.21)	0.84 (0.24)	0.85 (0.26)	0.89 (0.28)	1.23 (0.49)	1.26 (0.45)	1.26 (0.44)	1.26 (0.47)
	Female	0.73 (0.14)	0.75 (0.18)	0.74 (0.15)	0.76 (0.19)	0.95 (0.20)	1.04 (0.28)	1.01 (0.23)	1.07 (0.27)
RMSD _{ML} (cm)	Male	0.61 (0.15)	0.63 (0.22)	0.65 (0.15)	0.61 (0.17)	0.75 (0.20)	0.69 (0.17)	0.69 (0.19)	0.71 (0.17)
	Female	0.54 (0.18)	0.54 (0.18)	0.52 (0.12)	0.54 (0.15)	0.60 (0.17)	0.64 (0.17)	0.59 (0.11)	0.65 (0.12)
RMSD _{AP} (cm)	Male	0.77 (0.35)	0.70 (0.36)	0.79 (0.46)	0.76 (0.32)	0.68 (0.22)	0.75 (0.33)	0.77 (0.28)	0.72 (0.19)
	Female	0.80 (0.45)	0.64 (0.22)	0.72 (0.29)	0.68 (0.32)	0.64 (0.15)	0.60 (0.19)	0.60 (0.14)	0.67 (0.14)
Sway (cm ²)	Male	8.57 (4.94)	8.73 (7.40)	9.54 (7.05)	8.95 (5.91)	9.85 (5.53)	10.29 (6.89)	10.14 (6.21)	9.41 (3.75)
	Female	7.66 (3.80)	6.57 (3.66)	7.11 (3.95)	6.76 (3.34)	6.90 (1.98)	7.31 (3.54)	6.54 (2.19)	8.03 (2.26)
SEn _{ML}	Male	0.06 (0.01)	0.06 (0.02)	0.05 (0.01)	0.06 (0.02)	0.07 (0.02)	0.07 (0.02)	0.07 (0.01)	0.07 (0.02)
	Female	0.07 (0.02)	0.06 (0.02)	0.06 (0.01)	0.06 (0.02)	0.08 (0.02)	0.07 (0.02)	0.08 (0.02)	0.08 (0.02)
SEn _{AP}	Male	0.04 (0.02)	0.05 (0.02)	0.04 (0.02)	0.05 (0.02)	0.07 (0.02)	0.07 (0.03)	0.07 (0.03)	0.07 (0.02)
	Female	0.04 (0.02)	0.05 (0.03)	0.05 (0.03)	0.05 (0.03)	0.06 (0.02)	0.07 (0.03)	0.07 (0.03)	0.06 (0.02)
DFA _{ML}	Male	1.45 (0.08)	1.47 (0.07)	1.47 (0.07)	1.47 (0.06)	1.40 (0.07)	1.42 (0.07)	1.41 (0.06)	1.40 (0.08)
	Female	1.44 (0.05)	1.45 (0.07)	1.46 (0.06)	1.44 (0.06)	1.41 (0.06)	1.42 (0.05)	1.40 (0.05)	1.41 (0.08)
DFA _{AP}	Male	1.50 (0.10)	1.47 (0.08)	1.51 (0.08)	1.50 (0.10)	1.43 (0.10)	1.42 (0.10)	1.44 (0.08)	1.45 (0.08)
	Female	1.49 (0.08)	1.49 (0.11)	1.50 (0.10)	1.48 (0.11)	1.45 (0.06)	1.44 (0.08)	1.43 (0.09)	1.45 (0.07)
FLoS _{AREA} (%)	Male	49.09 (8.49)	48.09 (7.75)	51.74 (6.36)	44.56 (9.97)	43.45 (9.04)	44.20 (11.99)	45.73 (10.36)	42.03 (8.41)
	Female	50.94 (11.03)	50.71 (10.9)	49.73 (7.61)	49.79 (10.68)	48.79 (9.15)	45.12 (11.52)	43.91 (9.72)	46.50 (11.46)
TimeCont _{MEAN} (s)	Male	0.92 (0.12)	0.93 (0.15)	0.94 (0.14)	0.89 (0.18)	0.75 (0.08)	0.76 (0.12)	0.80 (0.12)	0.73 (0.13)
	Female	0.96 (0.16)	0.99 (0.15)	0.97 (0.15)	0.97 (0.20)	0.85 (0.16)	0.81 (0.13)	0.84 (0.14)	0.83 (0.17)
TimeCont _{MIN} (s)	Male	0.74 (0.10)	0.74 (0.12)	0.75 (0.12)	0.71 (0.15)	0.60 (0.07)	0.61 (0.10)	0.64 (0.10)	0.59 (0.11)
	Female	0.76 (0.13)	0.79 (0.13)	0.78 (0.13)	0.78 (0.17)	0.68 (0.13)	0.65 (0.11)	0.67 (0.12)	0.67 (0.14)

Table A-2. Mean (SD) of COP measures during **unipedal** standing.

		Eyes Open				Eyes Closed			
		NoExo	LaevoOn	LaevoOff	SuitX	NoExo	LaevoOn	LaevoOff	SuitX
MF _{ML} (Hz)	Male	0.52 (0.31)	0.42 (0.23)	0.53 (0.24)	0.40 (0.26)	0.56 (0.16)	0.50 (0.17)	0.46 (0.14)	0.55 (0.25)
	Female	0.45 (0.21)	0.50 (0.21)	0.55 (0.17)	0.51 (0.21)	0.46 (0.15)	0.51 (0.17)	0.47 (0.14)	0.47 (0.15)
MF _{AP} (Hz)	Male	0.23 (0.21)	0.24 (0.19)	0.23 (0.15)	0.23 (0.17)	0.33 (0.19)	0.40 (0.14)	0.36 (0.10)	0.38 (0.11)
	Female	0.22 (0.15)	0.26 (0.21)	0.25 (0.19)	0.17 (0.16)	0.37 (0.13)	0.32 (0.12)	0.36 (0.14)	0.33 (0.14)
MV _{ML} (cm/s)	Male	2.95 (0.69)	2.82 (0.57)	2.89 (0.41)	2.91 (0.73)	6.03 (0.94)	5.54 (1.30)	5.47 (1.10)	5.83 (1.37)
	Female	2.32 (0.48)	2.62 (0.50)	2.53 (0.48)	2.39 (0.43)	4.71 (0.79)	4.58 (0.87)	4.61 (0.83)	4.66 (0.78)
MV _{AP} (cm/s)	Male	2.50 (0.50)	2.62 (0.57)	2.59 (0.45)	2.51 (0.61)	6.08 (1.57)	5.06 (1.33)	5.72 (1.77)	5.65 (1.98)
	Female	2.07 (0.69)	2.21 (0.51)	2.02 (0.45)	1.93 (0.52)	4.16 (1.22)	4.15 (1.20)	4.35 (1.09)	4.48 (1.31)
RMSD _{ML} (cm)	Male	0.56 (0.12)	0.55 (0.11)	0.56 (0.13)	0.59 (0.14)	1.12 (0.20)	1.03 (0.18)	1.10 (0.24)	1.05 (0.25)
	Female	0.52 (0.12)	0.52 (0.07)	0.50 (0.07)	0.52 (0.10)	0.92 (0.18)	0.92 (0.17)	0.95 (0.20)	0.92 (0.17)
RMSD _{AP} (cm)	Male	0.74 (0.20)	0.75 (0.16)	0.78 (0.25)	0.75 (0.24)	1.34 (0.38)	1.05 (0.28)	1.39 (0.43)	1.22 (0.31)
	Female	0.66 (0.22)	0.67 (0.20)	0.66 (0.25)	0.65 (0.20)	0.92 (0.25)	0.99 (0.20)	1.04 (0.21)	1.08 (0.25)
Sway (cm ²)	Male	7.66 (2.93)	7.43 (2.34)	8.06 (3.48)	8.18 (3.45)	28.83 (12.67)	20.83 (9.15)	29.35 (13.37)	24.87 (10.87)
	Female	6.40 (2.80)	6.29 (1.92)	6.32 (3.13)	6.11 (2.23)	16.38 (7.34)	17.17 (6.20)	18.75 (6.63)	18.90 (6.98)
SEn _{ML}	Male	0.25 (0.08)	0.24 (0.06)	0.25 (0.06)	0.23 (0.07)	0.25 (0.05)	0.25 (0.06)	0.23 (0.05)	0.26 (0.05)
	Female	0.21 (0.05)	0.24 (0.05)	0.24 (0.04)	0.22 (0.03)	0.25 (0.05)	0.24 (0.05)	0.23 (0.05)	0.24 (0.04)
SEn _{AP}	Male	0.15 (0.06)	0.15 (0.05)	0.15 (0.06)	0.15 (0.07)	0.20 (0.05)	0.21 (0.04)	0.17 (0.05)	0.19 (0.04)
	Female	0.14 (0.05)	0.14 (0.05)	0.13 (0.05)	0.13 (0.05)	0.20 (0.04)	0.18 (0.05)	0.18 (0.05)	0.18 (0.04)
DFA _{ML}	Male	1.22 (0.14)	1.21 (0.10)	1.23 (0.10)	1.24 (0.12)	1.26 (0.09)	1.27 (0.10)	1.29 (0.07)	1.25 (0.09)
	Female	1.28 (0.09)	1.25 (0.08)	1.26 (0.07)	1.27 (0.06)	1.30 (0.08)	1.30 (0.08)	1.31 (0.08)	1.28 (0.08)
DFA _{AP}	Male	1.36 (0.12)	1.38 (0.10)	1.36 (0.09)	1.38 (0.09)	1.36 (0.09)	1.35 (0.07)	1.41 (0.09)	1.37 (0.08)
	Female	1.39 (0.08)	1.38 (0.09)	1.42 (0.08)	1.38 (0.08)	1.38 (0.07)	1.41 (0.08)	1.39 (0.08)	1.39 (0.06)
TimeCont _{MEAN} (s)	Male	0.52 (0.09)	0.52 (0.09)	0.53 (0.06)	0.53 (0.11)	0.36 (0.04)	0.37 (0.05)	0.38 (0.05)	0.37 (0.06)
	Female	0.56 (0.08)	0.54 (0.08)	0.54 (0.08)	0.58 (0.08)	0.39 (0.05)	0.41 (0.07)	0.40 (0.06)	0.40 (0.06)
TimeCont _{MIN} (s)	Male	0.41 (0.08)	0.42 (0.07)	0.42 (0.06)	0.42 (0.10)	0.28 (0.04)	0.30 (0.05)	0.30 (0.05)	0.29 (0.05)
	Female	0.45 (0.07)	0.43 (0.07)	0.44 (0.07)	0.47 (0.07)	0.31 (0.04)	0.33 (0.05)	0.32 (0.05)	0.32 (0.05)

Chapter 3. Effects of back-support exoskeleton use on gait performance and stability during level walking

Abstract

Back-support exoskeletons (BSEs) are a promising intervention to mitigate physical demands at work. Although growing evidence indicates that BSEs can reduce low-back physical demands, there is limited understanding of potential unintended consequences of BSE use, including the risk of falls. The objective of this study was to investigate the effects of BSE use on gait performance and stability during level walking. Twenty participants (10M, 10F) completed five level walking trials and a five-minute treadmill walking trial while wearing a BSE (backXTM) with three different levels of external torque (i.e., no-torque, low-torque, and high-torque) and in a control (no-exoskeleton) condition. Spatiotemporal gait patterns, stride-to-stride gait variability measures, required coefficient-of-friction (RCoF), and minimum foot clearance (MFC) were determined to assess gait performance. Gait stability was quantified using the maximum Lyapunov exponent (LyE) of trunk kinematics and the margin-of-stability (MoS). Using the backXTM with high supportive torque decreased slip risk (7% decrease in RCoF) and slightly improved trunk stability (3% decrease in LyE). However, it also decreased step length (1%), increased step width (10%) and increased gait variability (8-19%). Changes in MoS were complex: while MoS at heel strike decreased in the AP direction, it increased in the ML direction. There was a rather large decrease in MoS (26%) in the ML direction during the swing phase. This is the first study to quantify the effects of wearing a passive BSE with multiple supportive torque levels on gait performance and stability during level walking. Our results, showing that the external torque of the BSE may adversely affect gait step width, variability, and dynamic stability, can contribute to better design and practice guidelines to facilitate the safe adoption of BSEs in the workplace.

Keywords: occupational exoskeleton; walking stability; workplace fall

3. 1. Introduction

Back-support exoskeletons (BSEs) are wearable systems that provide external torques and/or structural support about the hips/trunk. Passive (unpowered) devices can reduce physical demands in tasks such as repetitive lifting, and may thus be an effective intervention addressing overexertion injuries in the workplace (e.g., Madinei et al., 2020; So et al., 2020). However, it is also important to consider whether using BSEs contribute to any unintentional consequences. For example, walking is a fundamental activity in many workplaces, and an earlier study (Baltrusch et al., 2018) found that walking was perceived as more difficult when wearing a BSE. This perception was likely due to the external torques that the exoskeleton generates about the hip/back, since a BSE typically engages when the relative angle between the trunk and the thigh decreases (e.g., hip flexion) (Baltrusch et al., 2019; Baltrusch et al., 2018). Hip flexion torque generated during the swing phase of gait is a major contributor to the propulsive power needed for walking (Sloot and van der Krogt, 2018). External hip extension torque generated by a BSE could thus compromise gait performance and stability, and thereby increase the risk of falling.

Exoskeletons for gait rehabilitation have been extensively examined in the literature (e.g., Rodríguez-Fernández et al., 2021). These devices are designed intentionally to assist gait and are fundamentally different from industrial exoskeletons, which are designed to assist with distinct functional tasks (e.g., overhead work or lifting). A previous study found that using an arm-support exoskeleton increased velocity of the center-of-pressure during quiet standing (Kim et al., 2018), possibly due to elevation of the center of mass (CoM) of the body + exoskeleton system. Although not directly comparable, a recent study of a powered whole body exoskeleton (Kim et al., 2021) reported slower gait and more involvement of the hip joint when using the exoskeleton. Using a passive BSE led to a slower preferred gait speed and shorter stride length (Baltrusch et al., 2019). We recently investigated the effects of two different passive BSEs on postural balance during quiet upright stance and found a decrease in postural balance (Park et al., 2021).

To promote the safe adoption of a BSE, it is thus important to understand how BSE use affects gait, and whether BSE effects on gait are dependent on specific external torque characteristics (e.g., torque levels).

Therefore, the current study aimed to assess the effects of different external torque levels of a passive BSE on gait performance and stability during level walking. We studied individuals wearing a BSE during over-ground and treadmill walking, with the device set to produce different levels of external torque. Gait performance was assessed using spatiotemporal gait parameters, stride-to-stride variability, required coefficient of friction (RCoF), and minimum foot clearance (MFC). Gait stability was assessed using the maximum Lyapunov exponent (LyE) and margin of stability (MoS). LyE indicates the ability of the motor system to attenuate small perturbations (Dingwell and Marin, 2006). MoS, defined as the distance between extrapolated center of mass and the boundaries of the base-of-support (Hof et al., 2005), is used to predict the risk of falling (Bruijn et al., 2013).

Based on the rationale that the external hip extension torque generated by the BSE can interfere with walking, we hypothesized that wearing a BSE will decrease both gait performance (i.e., increase in gait variability, increase in RCoF, and decrease in MFC) and stability (i.e., increase in LyE and decrease in MoS). We also hypothesized that such changes will be more pronounced as the external BSE-generated torque increases.

3. 2. Methods

3. 2. 1. Participants

A convenience sample of 20 young and healthy participants (10 males and 10 females) was recruited from the local university and community. Respective means (SD) of age, body mass, and stature were 24.8 (4.2) years, 77.2 (11.0) kg, and 173.0 (5.8) cm for the males; and 24.1 (1.9) years, 70.3 (10.0) kg, and 169.0 (3.5) cm for the females. All reported being physically active and with no current or recent musculoskeletal disorders or injuries. All participants provided informed consent following the study protocol approved by the Institutional Review Board at Virginia Tech.

3. 2. 2. Exoskeleton selection

The backXTM AC (US Bionics Inc., USA), designed to reduce physical demands on the back during forward bending (So et al., 2020), was used in this study. This exoskeleton has a mass of 4.5 kg and consists of a waist strap, a chest support, and an external torque generator about each hip that is coupled with the waist strap and the chest support (Figure 3-1). The four exoskeleton (*EXO*) conditions included in this study were: NoExo (no exoskeleton, control condition), EXO_{OFF} (backXTM with no supportive torque), EXO_{LOW} (backXTM with low supportive torque) and EXO_{HIGH} (backXTM with high supportive torque).

3. 2. 3. Experimental design and procedures

A repeated-measures design was used to assess the effects of the four *EXO* conditions on gait performance and stability. Each participant completed over-ground walking and treadmill walking in two experimental sessions on separate days. The presentation order of *EXO* conditions was counterbalanced using multiple 4×4 Latin Squares. The over-ground walking protocol afforded collection of ground reaction forces necessary for deriving some dependent measures, whereas the treadmill protocol allowed us to collect enough strides to compute other dependent measures (e.g., LyE).

In the first session, preferred walking speed (PWS) was determined in the NoExo condition using a treadmill (T101, Horizon Fitness Inc., Cottage Grove, WI), following a standard procedure (Jordan et al., 2007). Over-ground walking was performed subsequently on a linear walking track (1.5 m wide × 15.5 m long), in the middle of which were embedded two force platforms (Bertec Corporation, Columbus, OH; and AMTI, Boston, MA, USA; Figure 3-1 (a)). In a given *EXO* condition, participants were first provided with a 5 minute walking familiarization period. They then walked across the track following the procedure reported in Allin et al. (2016). Participants completed five acceptable walking trials (i.e., foot placement on the force platform and at PWS) in each *EXO* condition. In the treadmill session, participants walked for five minutes on the treadmill at their PWS in each *EXO* condition (Figure 3-1 (b)). This walking duration was used to provide sufficient strides for subsequent analyses of gait kinematic variability and LyE (Kang and Dingwell, 2009; Toebe et al., 2012).

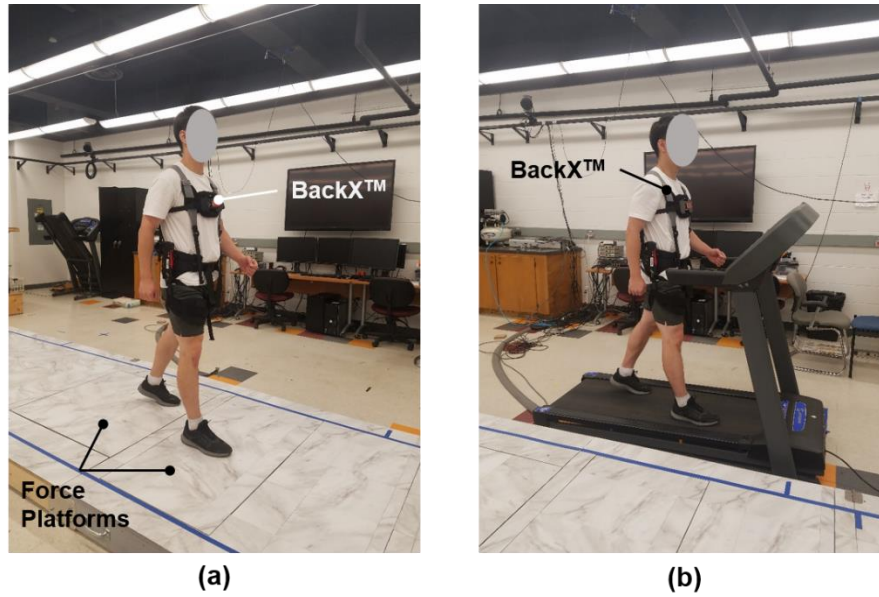


Figure 3-1. Illustrations of experimental trials during (a) over-ground walking and (b) treadmill walking.

3. 2. 4. Data collection and processing

Whole body kinematics and ground reaction forces were measured during over-ground walking trials. Thirty-three reflective markers and four rigid clusters (Figure B-1) were sampled at 120 Hz using a 12-camera optical motion capture system (Qualisys, Inc., Gothenburg, Sweden), then low-pass filtered (6 Hz cutoff; 4th-order Butterworth; bidirectional). Ground reaction forces from the two force platforms were sampled at 1200 Hz, low-pass filtered (36 Hz cutoff; 2nd-order Butterworth; bidirectional), and down-sampled to 120 Hz. Five consecutive steps in the middle of the track were used to compute spatiotemporal gait parameters, RCoF, MFC, and MoS. Bilateral foot and trunk kinematics during treadmill trials were measured using the same setup; these were low-pass filtered (10 Hz cutoff; 4th-order Butterworth; bidirectional) and used to compute gait variability and LyE. Gait events in each walking trial (i.e., heel-strike and toe-off of each foot) were detected using a coordinate-based algorithm (Zeni et al., 2008). Body segment CoMs were calculated using body segment scaling factors (Dumas et al., 2007). Three-dimensional rotations of each segment were obtained using the Cardan Y (*tilt*) – x' (*obliquity*) – z'' (*rotation*) sequence relative to the laboratory frame-of-reference system.

Means, standard deviations (across strides), and sample entropy (SaEn) of step length, step width, gait cycle time, swing time, and double support time were computed. SaEn was calculated using embedding dimension $m = 2$ and tolerance $r = 0.2$ (Yentes et al., 2013). Maximum RCoF was computed according to the method described in Chang et al. (2011). MFC was defined as the lowest elevation of the reflective markers over the foot, near the mid-swing phase of gait in a given swing phase. To obtain LyE, trunk linear and angular velocities (Kang and Dingwell, 2009) were first calculated using 3-point differences and then scaled to body size (Hof, 1996). A 12-dimensional state space was defined using linear and angular velocities and their time-delayed copies. Time delays were estimated by using the autocorrelation function (Chang et al., 2010; Rosenstein et al., 1993). Rosenstein's algorithm was used to compute the logarithmic rate of divergence (between 0 and 0.5 stride), and the latter was used to compute the short-term LyE (Dingwell and Marin, 2006; Rosenstein et al., 1993). Whole body CoM was calculated using a 13-segment model (bilateral foot, shank, thigh, upper arm, and forearm, as well as the pelvis, trunk, and head). Extrapolated CoM (XCoM) was then obtained using the equation described in Hof et al. (2005). MoS is defined as the minimum distance between XCoM and base-of-support borders (Hof et al., 2005; Sivakumaran et al., 2018). MoS measures used here included the minimum of MoSs during swing phase and the MoS at heel-strike, in both the anterior-posterior (AP) and medial-lateral (ML) directions (Figure 3-2). Positive(negative) MoS value indicate that XCoM is inside(outside) of the base-of-support boundary.

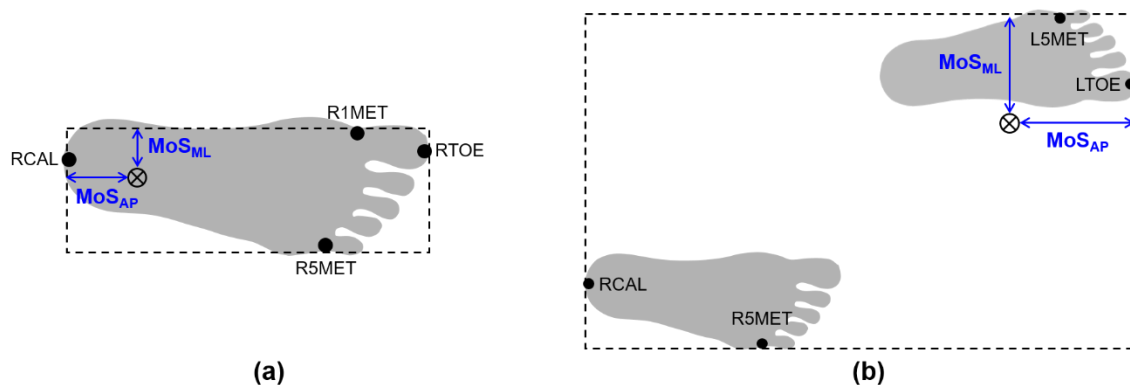


Figure 3-2. Demonstration of MoS calculated as the minimum distance between the XCoM and base-of-support boundary (dashed lines) during (a) swing phase and (b) at heel strike. TOE = 2nd distal phalanx; CAL = calcaneus; 1MET = 1st metatarsal head; 5MET = 5th metatarsal head.

3. 2. 5. Statistical analysis

Separate analyses of variance (ANOVAs) were performed on each outcome measure, including *EXO* as a within-subjects factor, gender (*GEN*) as a between-subjects factor, and walking speed as a covariate. All outcome measures were log-transformed to meet parametric model assumptions; summary statistics are provided in the original units after back transformation. Significant main effects were followed by *post hoc* pairwise comparisons using Tukey's HSD tests, and significant interaction effects were further examined using simple-effects analyses. In the case of any significant interaction effects, they were also visually examined to understand whether they were ordinal, and important. In the case of interactions being ordinal, (i.e., no cross-overs or opposing trends as a function of *GEN*), main effects of *EXO* were presented; in case of non-ordinal interactions, results of *EXO* effects were presented separately for each gender. Statistical significance was determined when $p < 0.05$, and partial eta-squared (η_p^2) was used to quantify effect sizes for main/interaction effects.

3. 3. Results

Descriptive summaries of the outcomes are included in the Appendix (Table B-1). Table 3-1 summarizes ANOVA results, regarding *EXO* and *GEN* effects; these results are present in more detail for each group of outcomes. Table B-2 presents summary of *post hoc* comparisons for the significant main effects of *EXO* on each outcome measure.

3. 3. 1. Spatiotemporal gait parameters, RCoF, and MFC

EXO had significant main and/or interactive effects on all spatiotemporal gait measures (Table 3-1 and Figure 3-3). Specifically, step length was lower in the *EXO*_{HIGH} condition compared to NoExo and *EXO*_{OFF}. Step width was higher in all exoskeleton conditions compared to the control condition. Gait cycle was shorter in the *EXO*_{HIGH} condition compared to NoExo and *EXO*_{OFF}, and was shorter in the *EXO*_{LOW} condition compared to NoExo. While swing time was longer in the *EXO*_{HIGH} condition compared to all other conditions, double support time was shorter in the *EXO*_{HIGH} condition compared to all other conditions.

Significant $EXO \times GEN$ interaction effects were found on step length and double support time. While males had greater step length in the control condition compared to all other conditions (all p -values < 0.05), females had lower step length in the EXO_{HIGH} condition compared to all other conditions (all p -values < 0.002). Males had shorter double support times in the EXO_{HIGH} condition compared to EXO_{OFF} ($p=0.002$) and EXO_{LOW} ($p=0.045$), whereas females had shorter double support times in the EXO_{HIGH} condition compared to all other conditions (all p -values < 0.001).

Table 3-1. Summary of ANOVA results for the main and interaction effects of EXO and GEN on each of the outcome measures. Table entries are p values (η_p^2), and significant effects are highlighted in bold ($p < 0.05$).

Outcome measures	Main and Interaction Effects		
	<i>EXO</i>	<i>GEN</i>	<i>EXO × GEN</i>
<i>Spatiotemporal gait parameters</i>			
Step length	< 0.001 (0.063)	0.115 (0.312)	0.032 (0.024)
Step width	< 0.001 (0.132)	0.842 (0.009)	0.218 (0.012)
Gait cycle	< 0.001 (0.105)	0.058 (0.544)	0.087 (0.018)
Swing time	< 0.001 (0.077)	0.505 (0.040)	0.143 (0.014)
Double support time	< 0.001 (0.116)	0.756 (0.014)	0.015 (0.027)
<i>Slip risk</i>			
RCoF	< 0.001 (0.049)	0.005 (0.149)	0.957 (0.001)
<i>Trip risk</i>			
MFC	< 0.001 (0.052)	0.847 (0.005)	0.013 (0.029)
<i>SD</i>			
Step length	< 0.001 (0.298)	0.175 (0.018)	0.386 (0.054)
Step width	< 0.001 (0.395)	0.075 (0.769)	0.062 (0.126)
Gait cycle	0.012 (0.183)	0.058 (0.002)	0.506 (0.042)
Swing time	< 0.001 (0.422)	0.761 (0.134)	0.634 (0.031)
Double support time	< 0.001 (0.413)	0.735 (0.293)	0.903 (0.010)
<i>SaEn</i>			
Step length	0.155 (0.092)	0.652 (0.003)	0.418 (0.051)
Step width	0.298 (0.065)	0.785 (0.009)	0.637 (0.031)
Gait cycle	0.005 (0.210)	0.067 (0.007)	0.140 (0.096)
Swing time	0.010 (0.189)	0.055 (0.001)	0.614 (0.033)
Double support time	0.586 (0.035)	0.761 (0.002)	0.466 (0.046)
<i>LyE</i>			
Trunk kinematics	< 0.001 (0.296)	0.127 (0.000)	0.160 (0.087)
<i>MoS</i>			
MoS _{AP} during swing (cm)	0.011 (0.030)	0.398 (0.151)	0.142 (0.014)
MoS _{ML} during swing (cm)	< 0.001 (0.052)	0.096 (0.109)	0.539 (0.006)
MoS _{AP} at heel strike (cm)	< 0.001 (0.287)	0.003 (0.690)	0.098 (0.017)
MoS _{ML} at heel strike (cm)	< 0.001 (0.097)	< 0.001 (0.606)	0.038 (0.023)

RCoF and MFC differed significantly across *EXO* conditions (Table 3-1 and Figure 3-3). RCoF was lower in the *EXO*_{HIGH} condition compared to all other conditions. MFC was lower in the *EXO*_{OFF} compared to NoExo and *EXO*_{LOW}. However, MFC was significantly lower in the *EXO*_{OFF} condition compared to all other *EXO* conditions (all *p*-values<0.025) only among males. *GEN* had a significant main effect on RCoF (Table 3-1), such that RCoF was higher in males than in females.

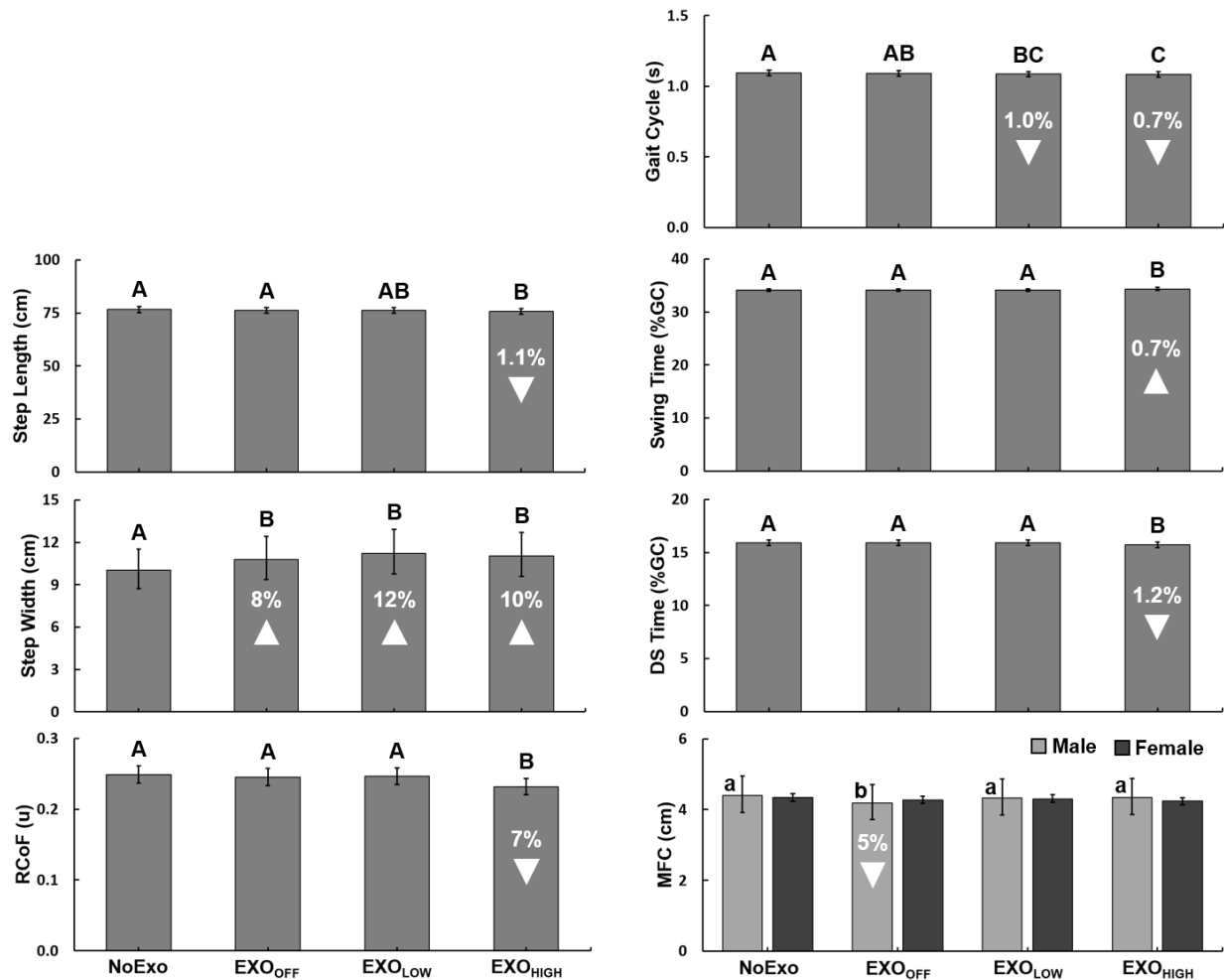


Figure 3-3. Spatiotemporal gait parameters, RCoF, and MFC in four *EXO* conditions during level walking. Data are presented as least-squares means with error bars indicating 95% confidence intervals. Upper case letters specify groupings obtained from pairwise comparisons between *EXO* conditions. Lower case letters show results from pairwise comparisons between *EXO* conditions within *GEN*. The triangle (or inverted triangle) symbols and accompanying numbers indicate a significant increase (or decrease) from the control no-exoskeleton condition, and the corresponding percentage change in dependent measure.

3.3.2. Gait variability

EXO had significant main effects on all SDs and SaEn of the gait cycle and swing time (Table 3-1 and Figure 3-4). Stride-to-stride gait variability was typically lower without the exoskeleton.

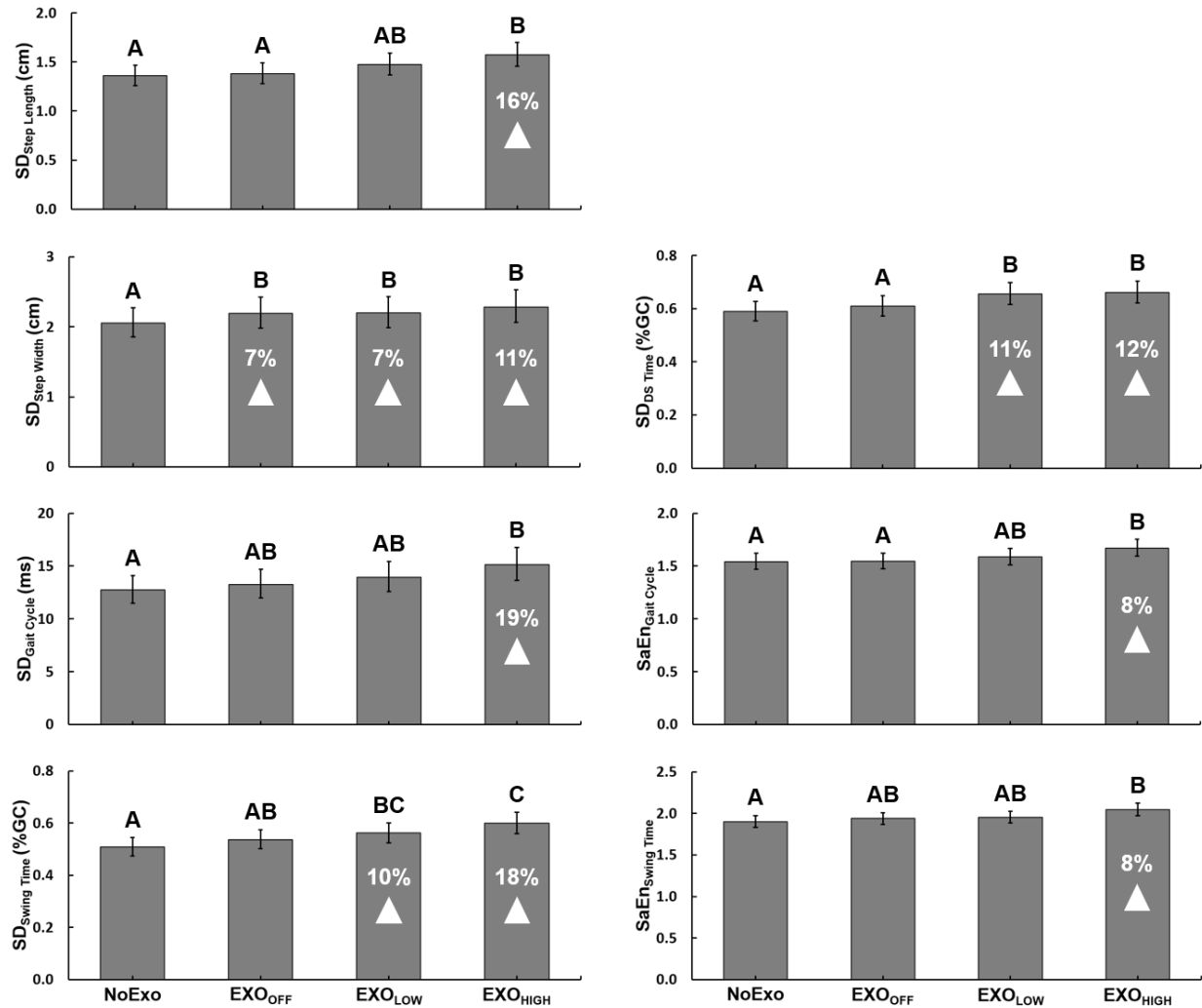


Figure 3-4. Measures of gait variability in the four EXO conditions during level walking. Data are presented as least-squares means with error bars indicating 95% confidence intervals. Upper case letters specify groupings obtained from pairwise comparisons between EXO conditions. The triangle (or inverted triangle) symbols and accompanying numbers indicate a significant increase (or decrease) from the control no-exoskeleton condition, and the corresponding percentage change in dependent measure.

Specifically: $SD_{StepLength}$ was higher in the EXO_{HIGH} condition compared to NoExo and EXO_{OFF}; $SD_{StepWidth}$ in the NoExo condition was lower than all exoskeleton conditions; $SD_{CycleTime}$ was higher in the EXO_{HIGH} as compared to NoExo; $SD_{SwingTime}$ was higher in the EXO_{HIGH} condition than NoExo and EXO_{OFF}, and was

also higher in the EXO_{LOW} condition than NoExo; SD_{DoubleSupportTime} was higher in both the EXO_{LOW} and EXO_{HIGH} conditions than NoExo and EXO_{OFF}; SaEn_{CycleTime} was higher in the EXO_{HIGH} condition as compared to NoExo and EXO_{OFF}; and SaEn_{SwingTime} in the EXO_{HIGH} condition was higher than NoExo.

3.3.3. Lyapunov exponent (LyE)

LyE of trunk kinematics during level walking differed significantly between EXO conditions (Table 3-1 and Figure 3-5). Specifically, LyE was lower in the EXO_{HIGH} condition as compared to NoExo and EXO_{OFF}, and was also lower in the EXO_{LOW} condition as compared to EXO_{OFF}.

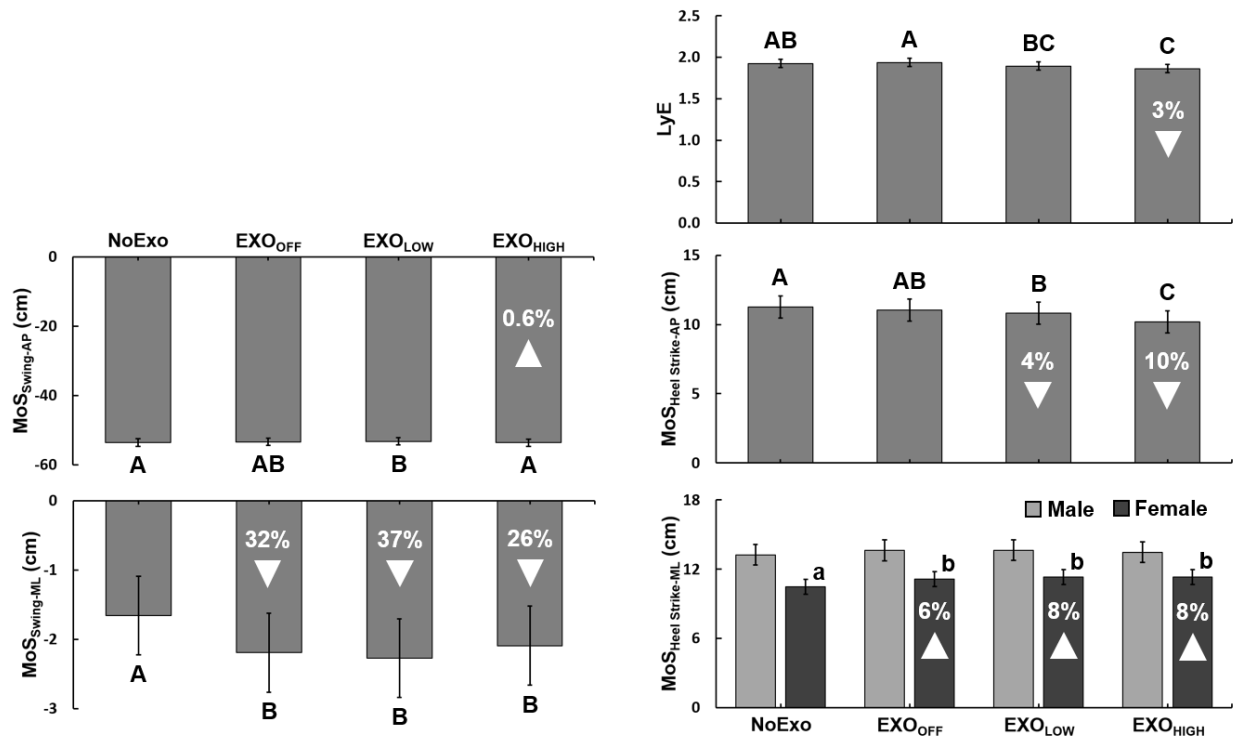


Figure 3-5. Measures of gait stability in the four EXO conditions during level walking. Data are presented as least-squares means with error bars signifying 95% confidence intervals. Upper case letters specify groupings obtained from pairwise comparisons between EXO conditions, and lower case letters show results from pairwise comparisons between EXO conditions within GEN. The triangle (or inverted triangle) symbols and accompanying numbers indicate a significant increase (or decrease) from the control no-exoskeleton condition, and the corresponding percentage change in dependent measure.

3. 3. 4. Margin of stability (MoS)

All MoS measures showed a significant main effect of *EXO* (Table 3-1 and Figure 3-5). During swing phase, MoS_{AP} was lower in the EXO_{LOW} condition than NoExo and EXO_{HIGH}, while MoS_{ML} was lower in NoExo compared to all other conditions. At heel strike, MoS_{AP} was lower in the EXO_{LOW} condition compared to NoExo, and was lower in the EXO_{HIGH} condition compared to all other conditions. Both the main effect of *EXO* and the *EXO*×*GEN* interaction effect were significant for MoS_{ML} at heel strike (Table 3-1 and Figure 3-5). Although all exoskeleton conditions led to a higher MoS than NoExo, simple effects analysis showed that these differences were significant only among females (all *p*-values<0.001). *GEN* had a significant main effect on both MoS_{AP} and MoS_{ML} at heel strike (Table 3-1), and both measures were higher among males.

3. 4. Discussion

3. 4. 1. Gait performance

Walking with the backXTM with high supportive torque (EXO_{HIGH}) led to statistically significant changes compared to the no exoskeleton (control) condition, which included decreases in step length, gait cycle time, and double support time and an increased swing time (Figure 3-3). These changes may have occurred because the external hip extension torque of the BSE impeded forward swing, but helped the participants move forward during double support. It is unclear whether these changes, which were rather small (0.7-1.2%), are practically meaningful. More substantial changes, though, were evident for step width. Compared to the control condition, BSE use increased step width by 8-12% (Figure 3-3). Participants may have adopted different hip lateral angles due to structural BSE constraints at the hip, and wider step widths could have enhanced lateral stability by increasing the base-of-support (Hak et al., 2012; McAndrew Young and Dingwell, 2012). However, this strategy might be sub-optimal in longer-term walking, with higher energetic costs (Baltrusch et al., 2019; Maxwell Donelan et al., 2001).

The EXO_{HIGH} condition caused a 6-7% decrease in RCoF (i.e., decreased slip risk) compared to the other conditions (Figure 3-3). When the maximum RCoF occurred, the EXO_{HIGH} condition led to significant changes in foot angle (36% decrease), normal force (46% increase), and shear force (36% increase) compared to the control condition. Hence, it appears that the EXO_{HIGH} condition reduced foot angle between heel-strike and mid-stance, which in turn resulted in a larger increase in normal vs. shear force, thereby decreasing RCoF. There were no significant changes in MFC (i.e., trip risk) with high external torque.

Wearing the backXTM substantially increased gait variability (7-19%), and the change in gait variability increased with the magnitude of external device torque (Figure 3-4). An increase in gait variability with BSE use may have stemmed from an inconsistent level of external torque being applied on the hip joint across strides, due to relative motion and misalignment between the device and human anatomy (Näf et al., 2018). It may also be the case that participants were simply not adequately adapted to the exoskeleton, or additional familiarization/practice is needed. An increase in stride-to-stride gait variability is associated with an increased risk of falling among older adults (Hausdorff et al., 2001; Maki, 1997). Our findings among young adults may not be directly comparable to older adults investigated in previous studies, yet changes in gait variability with BSE use here are of a similar magnitude, and hence might be of practical significance.

3. 4. 2. Gait stability

The EXO_{HIGH} condition significantly lowered the LyE values of the trunk compared to both the control and no-torque conditions (Figure 3-5). Although this result is inconsistent with our hypothesis, there are some explanations to consider. First, given that step width is important for trunk control during walking (Shih et al., 2021), the increased step width noted above with BSE use may have had beneficial effects on trunk stability. Second, participants may have increased trunk muscle activity to counteract perturbations (i.e., external torques applied to the hip joint) during swing phase, and this in turn could have contributed to increased trunk stability during walking. Third, Kang and Dingwell (Kang and Dingwell, 2009) noted that the effects of small perturbations on a body segment during walking could be attenuated by a greater inertia

of that segment. The total inertia of the trunk (human+BSE) most likely increased when wearing the BSE, and this increase could have reduced the divergence rate of the trunk motion for small perturbations.

At heel-strike, the EXO_{HIGH} condition decreased MoS_{AP} (6-10%) compared to other conditions, and the EXO_{LOW} condition also decrease MoS_{AP} compared to the control condition (Figure 3-5), both in agreement with our hypothesis. We assume that the BSE external torque hindered hip flexion during forward swing, which resulted in reduced step length (i.e., reduced base-of-support in AP direction), and thereby decreased MoS_{AP} at heel strike. Step length was indeed significantly and inversely correlated with MoS_{AP} at heel strike ($r = -0.52, p < 0.001$). BSE use increased MoS_{ML} (6-8%) at heel-strike among females (Figure 3-5), which may have resulted from an increased step width; in fact, step width was significantly correlated with MoS_{ML} at heel-strike ($r = 0.49, p < 0.001$). MoS_{ML} decreased (26-37%; i.e., more unstable compared to control condition; Figure 3-5), however, typically at 95-97% of swing phase (i.e., just before the heel strike). This apparent decrease in ML gait stability may have occurred because the whole body CoM travelled more in the lateral direction to increase step width, while the base-of-support remained unchanged.

3. 4. 3. Limitations and future work

Individual differences in BSE fit/adjustments and relative motions between the BSE and wearer may have caused misalignment between the exoskeleton and human hip-joint axes, which may have resulted in some external hip extension torque to be transferred as undesired hip abduction torque. We only investigated the short-term effects of BSE use. As such, our results may not be generalizable to long-term users whose postural control may evolve with repeated BSE use during walking. Our participants were all relatively young, and the generalizability of our findings to diverse populations is unclear. Future work should include more diverse populations and longer testing durations, to help understand how learning effects influence the findings reported above, and to more broadly quantify associations between exoskeleton use and fall risk, including whether exoskeletons affect balance recovery following induced perturbations.

3. 4. 4. Conclusions

Among a young and healthy participant group, using a BSE with high supportive torque decreased step length, increased step width, increased gait kinematic variability, and slightly improved trunk motion control during walking. A decrease in step length may have contributed to the observed decrease in slip risk and the margin of stability in the AP direction at heel strike, and an increase in step width may have resulted in the observed changes in margin of stability in the ML direction. Overall, changes in step width and increased gait variability were the most substantial effects found from the BSE, suggesting that using such an exoskeleton with high external torques could adversely affect fall risk and walking energetics.

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Appendix B

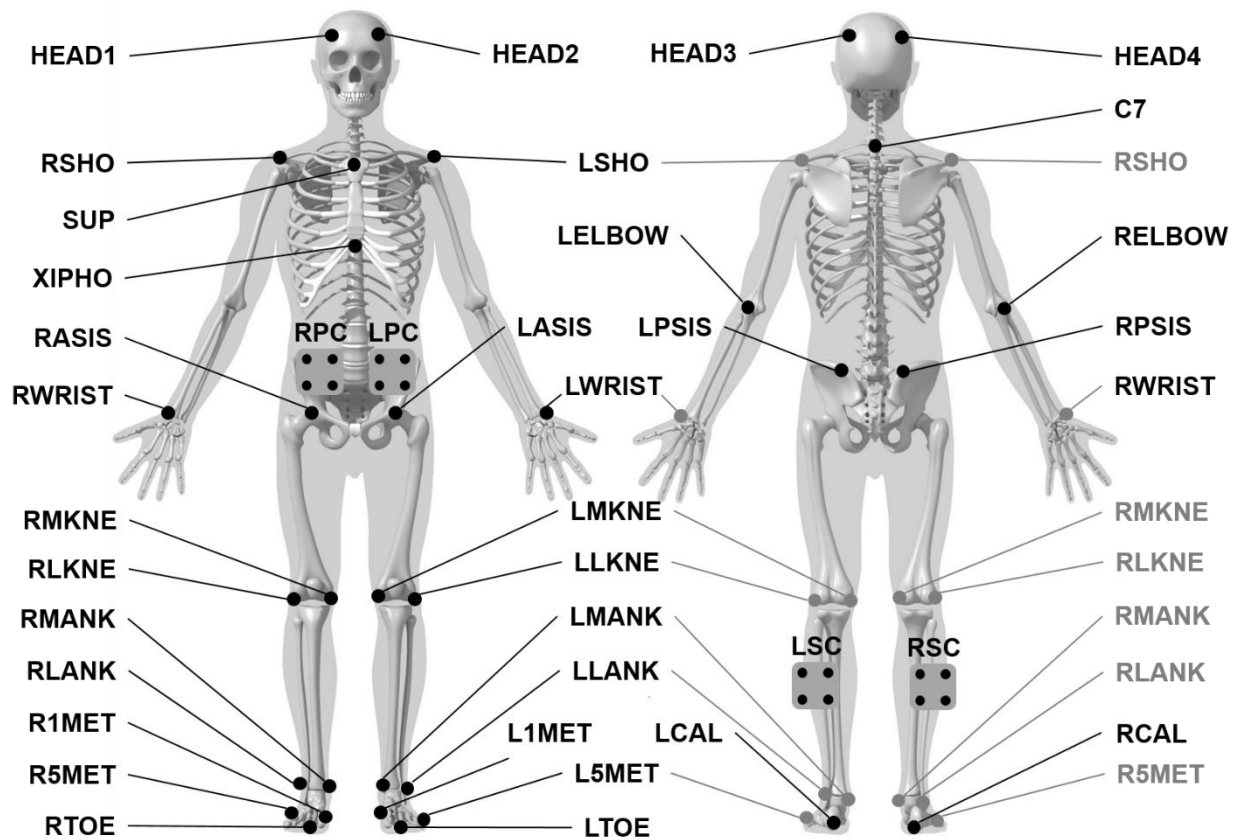


Figure B-1. Anterior (left) and posterior (right) views of the marker set for gait kinematics (TOE = 2nd distal phalanx; CAL = calcaneus; 1MET = 1st metatarsal head; 5MET = 5th metatarsal head; ANK = malleoli; KNE = femoral epicondyle; ASIS = anterior superior iliac spine; PSIS = posterior superior iliac spine; SUP = suprasternale; XIPHO = xiphoid process; C7 = spinous process of the 7th cervical vertebra; SHO = acromion; ELBOW = olecranon; WRIST = radial styloid; PC = pelvis cluster; SC = shank cluster). Participants wore a hair-band where four reflective markers (i.e., HEAD1-4) were attached. Wearing the backXTM covered anatomical landmarks on pelvis (i.e., LASIS, RASIS, LPSIS, and RPSIS). Hence, pelvis clusters (i.e., LPC and RPC) attached on the pelvis belt of the backXTM were used in exoskeleton conditions (i.e., EXO_{OFF}, EXO_{LOW}, and EXO_{HIGH}). Pelvis markers were reconstructed in subsequent data processing based on their relative location with pelvis clusters obtained in static calibration trials using a pointer tip. Skeleton illustration was obtained from Visual 3D ProfessionalTM (C-Motion Inc., Germantown, MD).

Table B-1. Least squares mean (95% confidence intervals) of all outcome measures.

Outcome measures	EXO condition			
	NoExo	EXO _{OFF}	EXO _{LOW}	EXO _{HIGH}
<i>Spatiotemporal gait parameters</i>				
Step length (cm)	76.66 (75.30, 78.05)	76.29 (74.93, 77.67)	76.25 (74.89, 77.63)	75.82 (74.48, 77.20)
Step width (cm)	10.02 (8.71, 11.53)	10.80 (9.39, 12.42)	11.22 (9.76, 12.91)	11.04 (9.60, 12.70)
Gait cycle (ms)	1094 (1075, 1113)	1090 (1071, 1110)	1086 (1067, 1105)	1083 (1064, 1102)
Swing time (%GC)	34.13 (33.87, 34.40)	34.13 (33.87, 34.40)	34.14 (33.88, 34.41)	34.36 (34.09, 34.63)
Double support time (%GC)	15.90 (15.64, 16.17)	15.93 (15.67, 16.20)	15.93 (15.67, 16.19)	15.71 (15.46, 15.97)
<i>Slip risk</i>				
RCoF (μ)	0.249 (0.237, 0.261)	0.245 (0.234, 0.258)	0.246 (0.235, 0.259)	0.232 (0.221, 0.243)
<i>Trip risk</i>				
MFC (cm)	4.37 (4.13, 4.62)	4.23 (4.00, 4.47)	4.32 (4.08, 4.57)	4.29 (4.06, 4.53)
<i>SD</i>				
Step length (cm)	1.36 (1.26, 1.47)	1.38 (1.28, 1.49)	1.47 (1.37, 1.59)	1.57 (1.46, 1.70)
Step width (cm)	2.05 (1.86, 2.27)	2.19 (1.98, 2.43)	2.20 (1.99, 2.43)	2.28 (2.06, 2.53)
Gait cycle (ms)	12.73 (11.50, 14.10)	13.25 (11.97, 14.68)	13.93 (12.58, 15.43)	15.12 (13.66, 16.75)
Swing time (%GC)	0.51 (0.47, 0.54)	0.54 (0.50, 0.57)	0.56 (0.52, 0.60)	0.60 (0.56, 0.64)
Double support time (%GC)	0.59 (0.55, 0.63)	0.61 (0.57, 0.65)	0.66 (0.62, 0.70)	0.66 (0.62, 0.70)
<i>SaEn</i>				
Step length	2.06 (2.00, 2.11)	2.15 (2.09, 2.21)	2.09 (2.03, 2.15)	2.07 (2.02, 2.13)
Step width	2.15 (2.10, 2.21)	2.09 (2.04, 2.15)	2.12 (2.06, 2.17)	2.15 (2.10, 2.21)
Gait cycle	1.54 (1.47, 1.62)	1.55 (1.47, 1.62)	1.59 (1.51, 1.67)	1.67 (1.59, 1.75)
Swing time	1.90 (1.83, 1.97)	1.94 (1.87, 2.01)	1.95 (1.88, 2.02)	2.05 (1.98, 2.12)
Double support time	2.05 (1.92, 2.20)	1.96 (1.83, 2.10)	2.00 (1.87, 2.14)	2.01 (1.88, 2.15)
<i>LyE</i>				
Trunk kinematics	1.92 (1.87, 1.97)	1.94 (1.89, 1.99)	1.89 (1.84, 1.94)	1.86 (1.82, 1.91)
<i>MoS</i>				
MoS _{AP} during swing (cm)	-53.55 (-54.61, -52.50)	-53.39 (-54.45, -52.33)	-53.25 (-54.30, -52.19)	-53.59 (-54.65, -52.54)
MoS _{ML} during swing (cm)	-1.66 (-2.23, -1.09)	-2.19 (-2.76, -1.62)	-2.27 (-2.84, -1.70)	-2.09 (-2.66, -1.52)
MoS _{AP} at heel strike (cm)	11.25 (10.46, 12.05)	11.04 (10.24, 11.83)	10.83 (10.04, 11.62)	10.18 (9.39, 10.97)
MoS _{ML} at heel strike (cm)	11.87 (11.35, 12.39)	12.40 (11.88, 12.91)	12.49 (11.98, 13.01)	12.38 (11.86, 12.90)

Table B-2. Summary of *post hoc* pairwise comparisons (Tukey's HSD tests) for the significant main effects of *EXO* on each of the outcome measures. Table entries are *p* values, and significant differences are highlighted in bold ($p < 0.05$).

Outcome measures	<i>EXO</i> condition					
	NoExo vs. EXO _{HIGH}	EXO _{OFF} vs. EXO _{HIGH}	EXO _{LOW} vs. EXO _{HIGH}	NoExo vs. EXO _{LOW}	EXO _{OFF} vs. EXO _{LOW}	NoExo vs. EXO _{OFF}
<i>Spatiotemporal gait parameters</i>						
Step length	< 0.001	0.029	0.057	0.069	0.124	0.995
Step width	< 0.001	0.518	0.741	< 0.001	0.084	< 0.001
Gait cycle	< 0.001	< 0.001	0.346	< 0.001	0.067	0.197
Swing time	< 0.001	< 0.001	< 0.001	0.999	0.999	1.000
Double support time	< 0.001	< 0.001	< 0.001	0.925	0.998	0.850
<i>Slip risk</i>						
RCoF	< 0.001	0.009	0.005	0.940	0.995	0.846
<i>Trip risk</i>						
MFC	0.055	0.234	0.795	0.371	0.028	< 0.001
<i>SD</i>						
Step length	< 0.001	0.002	0.233	0.099	0.242	0.968
Step width	< 0.001	0.134	0.166	0.003	1.000	0.003
Gait cycle	0.009	0.067	0.402	0.324	0.778	0.868
Swing time	< 0.001	< 0.001	0.088	0.003	0.360	0.199
Double support time	< 0.001	0.003	0.984	< 0.001	0.009	0.452
<i>SaEn</i>						
Gait cycle	0.009	0.010	0.156	0.643	0.678	1.000
Swing time	0.007	0.067	0.147	0.604	0.983	0.818
<i>LyE</i>						
Trunk kinematics	0.004	< 0.001	0.286	0.255	0.049	0.856
<i>MoS</i>						
MoS _{AP} during swing	0.984	0.300	0.016	0.044	0.601	0.508
MoS _{ML} during swing	0.019	0.899	0.610	< 0.001	0.951	0.002
MoS _{AP} at heel strike	< 0.001	< 0.001	< 0.001	< 0.001	0.128	0.089
MoS _{ML} at heel strike	< 0.001	0.998	0.721	< 0.001	0.814	< 0.001

Chapter 4. Effects of back-support exoskeleton use on lower limb joint kinematics and kinetics during level walking

Abstract

As increasing evidence is emerging for the effectiveness of occupational back support exoskeletons (BSE) in reducing low-back physical demands, there is currently limited knowledge on how BSE use affects balance and postural control. We assessed the effects of using a BSE on lower limb joint kinematics and kinetics during level walking. Twenty young, healthy participants completed level walking trials while wearing a BSE (backXTM) with three different levels of supporting torque (i.e., no torque, low, and high) and in a control condition (no-exoskeleton). The torque profile of the backXTM was measured using a dynamometer, and used for assessing hip torque generated by the participant only. When hip extension torques were required for walking (i.e., the initial 0-10% and final 75-100% of the gait cycle), the backXTM with high supportive torque generated substantial external hip extension torque, and participants showed adapted gait patterns (i.e., decrease in hip extension torque and extensor power, to compensate for the external hip extension torque provided by the BSE, so that the total human+EXO hip extension torque and power remained the same across conditions). However, participants increased hip extensor power and ankle dorsiflexor power in the stance phase in those conditions when the BSE did not provide external torque, to compensate for the added mass and inertial effects of the device. When hip flexion torques were required for walking (i.e., the middle 15-70% of the gait cycle), although the backXTM with high supportive torque did not generate substantial hip extension torque, participants showed changes in lower limb kinematics (i.e., increase in ankle plantarflexion and dorsiflexion velocities) and kinetics (i.e., increase in hip flexor, knee extensor, and ankle plantarflexor powers) to increase mechanical energy for propelling the leg into the swing phase. Wearing the backXTM also impeded hip flexion during walking by decreasing hip range of motion and peak hip flexion velocity. Hence, wearing the backXTM may increase metabolic costs associated with walking. Whether such changes in joint kinematics and kinetics will in turn further lead to muscle fatigue and/or postural instability in long-distance walking are yet to be determined. Future work

should also extend such analyses of lower-limb kinetics to other tasks that may require greater hip flexion, such as obstacle crossing/negotiation and stair climbing.

Keywords: occupational exoskeleton, gait performance, walking kinematics, walking kinetics

4. 1. Introduction

A back-support exoskeleton (BSE) is a wearable system that can assist the wearer by providing external torque and/or structural support about the hip/trunk (Kermavnar et al., 2021; Toxiri et al., 2019). As work-related musculoskeletal disorders (WMSDs) of the low back continue to be a serious health problem in the workplace (Ferguson et al., 2019), there is growing interest in exoskeleton technologies as a promising ergonomic intervention to reduce the risk of low-back WMSDs. Earlier investigations have shown that BSE use can reduce physical loads on the low back (e.g., Madinei et al., 2020; Motmans et al., 2018). However, there also are concerns regarding unexpected or unintended challenges, such as increased discomfort (Giustetto et al., 2021), decreased agility (Kermavnar et al., 2021), and more rigid movements (Koopman et al., 2019) when wearing a BSE.

Given that BSEs are designed primarily to assist during physical tasks requiring lifting/lowering and/or stooped posture (Kozinc et al., 2020; Toxiri et al., 2019), BSE use may have adverse effects on performing non-stationary tasks such as walking and carrying. Recent studies have shown that walking was perceived as more difficult when wearing a BSE (Baltrusch et al., 2018), and that both preferred speed and stride length decreased during walking with a BSE (Baltrusch et al., 2019). These effects may be due to the external torque (in hip extension) provided by the BSE when the relative angle between the thigh and the trunk decreases. Park et al. (2020) similarly reported that wearing a BSE altered gait performance during level walking (i.e., decrease in both stride length and double support time), and that such changes were more pronounced when the external torque from the BSE increased.

In addition to spatiotemporal parameters, lower limb kinematics (e.g., joint angle and angular velocity) and kinetics (e.g., joint torque and power) are critical to understand gait. Previous studies have found that

abnormal joint kinematics and/or kinetics are associated with gait instability. For example, during walking, the peak hip extension angle and the peak hip flexion torque were smaller in patients with hip dysplasia than in healthy people (Jacobsen et al., 2014). Compared to young adults, healthy older adults exhibited different joint kinematics (e.g., decrease in late-stance ankle plantarflexion angle, increase in hip flexion angle during stance phase, decrease in foot angle at heel-strike) and kinetics (e.g., increase in early-stance hip extensor power generation, late-stance hip flexor power generation, and late-stance knee power absorption) during walking (DeVita and Hortobagyi, 2000; McGibbon and Krebs, 2004; Winter et al., 1990). By engaging during hip flexion and generating an extension torque about the hip joints, BSE use may thus alter both lower limb joint kinematics and kinetics in an unexpected way.

Although studies in other contexts (e.g., rehabilitation devices or gait assistive devices) suggest that external torque provided by an exoskeleton at the hip may not substantially alter gait kinematics and kinetics during walking, these results are not directly applicable to industrial BSEs. For example, Lewis and Ferris (2011) demonstrated that healthy individuals showed similar knee/ankle angle and hip torque profiles over the gait cycle regardless of use of a hip exoskeleton. Their exoskeleton was designed to provide hip flexion assistance approximately between 1/3rd to 1/2nd of the gait cycle. However, little is known regarding the effects of industrial/occupational BSEs on joint kinematics and kinetics, given the differences in design and/or supportive torque magnitudes between industrial BSEs and rehabilitation devices; and, due to the fact that as opposed to gait-assistance exoskeletons, BSEs are not explicitly designed to follow the human gait cycle.

In this study, we investigated the effects of different external torque levels of a BSE on lower limb joint (i.e., hip, knee, and ankle) kinematics and kinetics. Joint kinematics were explored using range of motion (RoM) and angular velocities, while joint kinetics were investigated using torque and power, during level walking. The torque profile of the BSE was measured using a dynamometer and used for computing net hip kinetics generated by the participant only. We hypothesized that wearing a BSE will hinder hip joint motion (i.e., decrease in hip RoM), and lead to an increase in hip flexion torque, due to the external hip

extension torque generated by the device. We also hypothesized that, although the external torque constraints of a BSE are associated primarily with the hip joint, these constraints will also cause synchronized or compensatory changes in knee/ankle kinematics and kinetics.

4. 2. Methods

4. 2. 1. Exoskeleton

The backXTM AC (US Bionics Inc., USA), designed to reduce physical demands on the back during forward bending, was used in this study. This exoskeleton has mass of 4.5 kg and consists of a waist strap, chest support, and an external torque generator about each hip that is coupled with the waist strap and the chest support. The four exoskeleton (*EXO*) conditions included in this study were: NoExo (no exoskeleton, control condition), EXO_{OFF} (backXTM with no supportive torque), EXO_{LOW} (backXTM with low supportive torque) and EXO_{HIGH} (backXTM with high supportive torque). The EXO_{OFF} condition was included as a way to isolate the effects of the additional mass and joint movement restriction effects associated with a BSE on walking performance, and to delineate these from the effects of external hip extension torque, which was only present in the other *EXO* conditions (EXO_{LOW} and EXO_{HIGH}).

4. 2. 2. Exoskeleton torque profile measurement

The backXTM torque profile was assessed using the Humac NormTM isokinetic dynamometer (Computer Sports Medicine Inc., Stoughton, MA), with the backXTM attached to a mannequin (MZ-HM01, RoxyDisplay, East Brunswick, NJ). This mannequin had a hinge joint at the hip and was utilized to mimic hip extension/flexion, to avoid any active voluntary or reflexive movements that may have resulted from involving human subjects. The backXTM was first fitted to the mannequin, the mannequin was positioned and secured on the dynamometer bed, and then the left thigh of the mannequin was connected to the hip adapter of the dynamometer (Figure 4-1).



Figure 4-1. Demonstration of the torque profile measurements for the backX™ using a dynamometer.

The mannequin was repeatedly moved into isokinetic hip extension/flexion using the continuous passive mode of the Humac Norm™. Hip torque was recorded via the dynamometer software (HUMAC 2016, v.10.7.1) at 100 Hz. Alignment was maintained for the hip joint center of the mannequin, the center of the spring-loaded joint of the backX™, and the rotation center of the dynamometer. Hip angle was gradually varied between 0 and 120 degrees at several constant speeds: 0 (static), 20, 40, 60, 80, 100, and 120 deg./s. For each speed, 14 extension/flexion trials were completed, and the mean was obtained for subsequent processing.

Hip torque profile was assessed for three *EXO* conditions (i.e., the mannequin wearing *EXO_{OFF}*, *EXO_{LOW}*, and *EXO_{HIGH}*). The mannequin-only condition (i.e., no backX™) was also measured to obtain offset torque data. Hip torque profile for *EXO_{OFF}*, *EXO_{LOW}*, and *EXO_{HIGH}* were finally obtained by subtracting hip torque data for the mannequin-only condition from the mannequin in each respective *EXO* condition. Figure 4-2 presents sample torque profiles of the BSE at 40 deg./s. The backX™ generated higher hip extension torque during hip flexion than during hip extension. Maximum torques for *EXO_{OFF}*, *EXO_{LOW}*, and *EXO_{HIGH}* were 5.5, 17.8, 27.4 Nm, respectively.

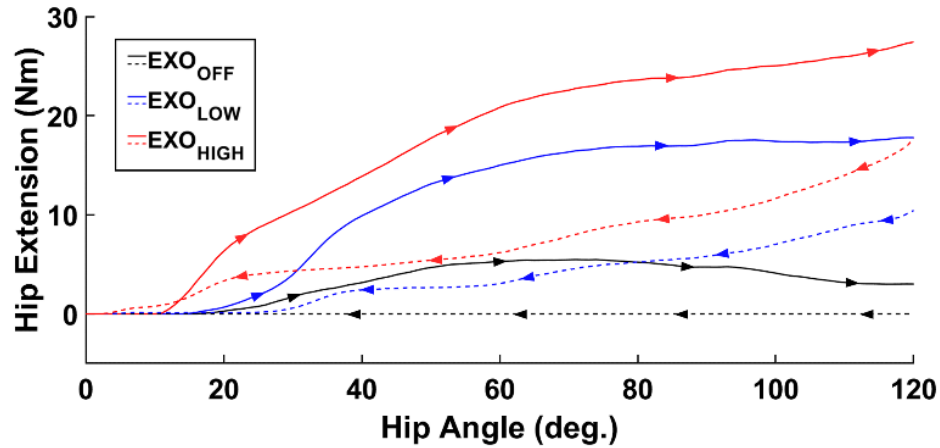


Figure 4-2. Torque profiles in EXO_{OFF} (black), EXO_{LOW} (blue), and EXO_{HIGH} (red) conditions during isokinetic hip flexion (solid lines) and extension (dashed lines) at 40 deg./s. Note that torque profiles depicted here were obtained from one smart joint (i.e., joint on one side of the BSE).

4. 2. 3. Participants

A convenience sample of 20 young and healthy participants was recruited from the university and local community. Participants included 10 males with mean (SD) age of 24.8 (4.2) years, stature of 173.0 (5.8) cm, and body mass of 77.2 (11.0) kg; corresponding values for 10 females were 24.1 (1.9) years, 169.0 (3.5) cm, and 70.3 (10.0) kg, respectively. All reported being physically active (i.e., exercising 2-3 times per week), and had no self-reported current or recent (past 12 months) musculoskeletal disorders or injuries. This research complied with the tenets of the Declaration of Helsinki and was approved by the Institutional Review Board at Virginia Tech. Written informed consent was obtained from each participant prior to data collection.

4. 2. 4. Experimental design and procedures

We used a repeated-measures design to assess the effects of different external torque levels of the BSE on lower limb joint kinematics and kinetics during level walking. Level walking was performed on a linear walking track (1.5 m wide $\times$ 15.5 m long), in the middle of which a force platform was embedded (AMTI, Boston, MA), to be level with the rest of the track. Participants were asked to walk across the track at a predetermined preferred walking speed (PWS). The presentation order of EXO conditions was counterbalanced using 4 $\times$ 4 Balanced Latin Squares.

Participants completed one experimental session (~2 hours). They were first fitted with the backX™ following the manufacturer's recommendations and then were familiarized with the device. PWS was determined, without wearing the backX™, following the procedure reported by Jordan et al. (2007). During the level walking task, for a given *EXO* condition, participants were asked to walk on the track for ~5 min. to become familiarized with their PWS; we used a stopwatch and gave verbal feedback to ensure they matched their PWS within ± 0.1 m/s. After a 1-min rest break, participants completed five acceptable walking trials (i.e., proper foot placement on the force platform and PWS) in each experimental condition. No rest was given between walking trials. Participants were not aware of where the force platforms were located (so as to not disturb their natural walking).

4. 2. 5. Instrumentation and outcome measures

Body segment (bilateral foot, shank, and thigh, as well as the pelvis and trunk) kinematics and ground reaction forces were measured during over-ground walking trials. Reflective markers (n=33) and four rigid marker clusters (Figure B-1) were sampled at 120 Hz using a 12-camera optical motion capture system (Qualisys, Inc., Gothenburg, Sweden), then low-pass filtered (6 Hz cutoff; 4th-order Butterworth; bidirectional). Triaxial ground reaction forces and torques from the force platform were sampled at 1200 Hz, subsequently down-sampled to 120 Hz, and low-pass filtered (10 Hz cutoff; 2nd-order Butterworth; bi-directional). From each time the participants walked across the track, one stride starting from the heel-strike on the force platform was used for further analysis. Three-dimensional rotations of each segment were defined using the Cardan $Y - x' - z''$ (tilt–obliquity–rotation) convention relative to the laboratory reference frame. In addition to joint angles and angular velocities, linear and angular accelerations of each body segment were calculated using a standard three-point difference equation.

Sagittal plane joint kinetics for hip, knee, and ankle were assessed from the limb that contacted the force platform only, assuming left/right symmetry. Body segment center-of-masses were calculated using scaling factors provided in Dumas et al. (2007) and used for representing the linear motions of each segment. Joint torques were calculated based on the ground reaction forces/torques and body segment kinematics, using a

bottom-up inverse dynamics model (Kingma et al., 1996) implemented using custom code in MATLAB (MathWorks, Natick, MA). The calculated hip torque ($T_{\text{HIP-TOTAL}}$) includes torque generated by participant plus the backXTM. Hence, hip torque generated by the participant only ($T_{\text{HIP-HUMAN}}$) was calculated by subtracting the torque generated by the device ($T_{\text{HIP-BSE}}$) from $T_{\text{HIP-TOTAL}}$. $T_{\text{HIP-BSE}}$ values for any given hip angle and angular velocity were estimated using a two-dimensional interpolation technique, applied on the data collected using the dynamometer (section 4. 2. 2). Joint power was calculated as the product of the joint torque and angular velocity. Hip, knee, and ankle joint powers were defined using terminology adopted from Winter (2009) (Figure C-1): Power generation occurs via the hip extensors (H1), hip flexors (H3), knee extensors (K2), and ankle plantarflexors (A2), while power absorption occurs via the hip flexors (H2), knee extensors (K1 and K3), knee flexors (K4), and ankle dorsiflexors (A1). Hip joint power generated by the participant were calculated using $T_{\text{HIP-HUMAN}}$. Joint torque and power quantities were normalized to body mass.

Outcome measures for joint kinematics included RoM at the hip (thigh to trunk), knee, and ankle, and peak velocities in flexion and extension at the hip (thigh to trunk), knee, and ankle. Outcome measures for joint torque included peak hip extension and flexion torques, peak knee extension and flexion torques, and peak ankle plantarflexion and dorsiflexion torques. Outcome measures for joint power included peak value in H1 ($H1_{\text{GENERATION}}$), H2 ($H2_{\text{ABSORPTION}}$), H3 ($H3_{\text{GENERATION}}$), K1 ($K1_{\text{ABSORPTION}}$), K2 ($K2_{\text{GENERATION}}$), K3 ($K3_{\text{ABSORPTION}}$), K4 ($K4_{\text{ABSORPTION}}$), A1 ($A1_{\text{ABSORPTION}}$), and A2 ($A2_{\text{GENERATION}}$).

4. 2. 6. Statistical analysis

Separate two-way, mixed-factor analyses of covariance (ANCOVAs) were used to assess the effects of *EXO* and gender (*GEN*) on each set of outcome measures, with walking speed as a covariate. Parametric model assumptions were evaluated, and data transformations were performed to meet these assumptions as required. Order of condition presentation effects was also explored for significance in the initial models. Significant effects were followed by *post hoc* pairwise comparisons using Tukey's HSD tests, and significant interaction effects were further examined using simple-effects analyses. In the case of any

significant interaction effects, they were also visually examined to understand whether they were ordinal, and important. In the case of interactions being ordinal, (i.e., no cross-overs or opposing trends as a function of *GEN*), main effects of *EXO* were presented; in case of non-ordinal interactions, results of *EXO* effects were presented separately for each gender. Partial eta-squared (η_p^2) was used to quantify effect sizes for significant main/interaction effects. After analysis, summary results were back-transformed and are presented in the original units in text and figures. Summary data are presented as least-squares means (95% confidence intervals) unless stated otherwise. All statistical analyses were completed using JMP® Pro 15.0 (SAS Institute Inc., Cary, NC), using the restricted maximum likelihood (REML) method, with statistical significance concluded when $p < 0.05$.

4. 3. Results

Figure C-1 shows summary data (from all walking trials across all participants) of sagittal plane angles, angular velocities, torques, and powers for the hip, knee, ankle in NoExo and EXO_{HIGH} conditions (normalized and represented from 0-100% gait cycle). Table 1 summarizes ANCOVA results of *EXO* and *GEN* effects; these results are presented in more detail for each group of outcomes. Tables C-1 and C-2 in the appendix include descriptive summaries of the outcomes and summary of *post hoc* comparisons for the significant main effects of *EXO* on each outcome measure, respectively.

4. 3. 1. Joint RoM

A significant main effect of *EXO* was seen in all joint RoM measures (Table 1). Specifically, hip RoM significantly decreased from no-exoskeleton control condition to EXO_{OFF}, from EXO_{OFF} to EXO_{LOW}, and from EXO_{LOW} to EXO_{HIGH} (Figure 4-3). Simple effects analysis of the *EXO*×*GEN* interaction showed that while all pairwise differences were significant for males (p -values<0.015), hip RoM was lower in EXO_{HIGH} than in EXO_{LOW}, and also lower in EXO_{LOW} than in NoExo and EXO_{OFF} conditions among females (p -values<0.001). Knee RoM was lower in EXO_{HIGH} condition compared to NoExo (Figure 4-3). Ankle RoM was lower in EXO_{HIGH} than in EXO_{OFF} and EXO_{LOW} conditions (Figure 4-3).

Table 4-1. Summary of ANCOVA results for the main and interaction effects of *EXO* and *GEN* on each set of outcome measures. Any transformations used on the outcome measures are specified. For each effect, entries are p values (η_p^2), and significant effects are highlighted in bold ($p < 0.05$).

Outcome measures	Data transformation performed	Main and interaction effects		
		<i>EXO</i>	<i>GEN</i>	<i>EXO</i> × <i>GEN</i>
<i>RoM</i>				
Hip	Log	< 0.001 (0.518)	0.140 (0.230)	< 0.001 (0.078)
Knee	Log	0.005 (0.034)	0.657 (0.020)	0.337 (0.009)
Ankle	Log	0.002 (0.038)	0.702 (0.018)	0.841 (0.003)
<i>Peak angular velocity</i>				
Hip extension	Log	< 0.001 (0.202)	0.660 (0.012)	0.333 (0.009)
Hip flexion	Log	< 0.001 (0.226)	0.002 (0.507)	0.130 (0.014)
Knee extension	Log	< 0.001 (0.231)	0.094 (0.170)	0.094 (0.017)
Knee flexion	Log	0.095 (0.016)	0.933 (0.003)	0.156 (0.013)
Ankle plantarflexion	Log	< 0.001 (0.143)	0.024 (0.446)	0.048 (0.021)
Ankle dorsiflexion	—	< 0.001 (0.225)	0.003 (0.505)	0.012 (0.028)
<i>Peak torque</i>				
Hip extension	—	< 0.001 (0.144)	0.661 (0.021)	0.871 (0.002)
Hip flexion	Square root	0.034 (0.023)	0.790 (0.012)	0.773 (0.003)
Knee extension	Square root	0.027 (0.024)	0.488 (0.038)	0.199 (0.013)
Knee flexion	Square root	< 0.001 (0.156)	0.005 (0.389)	0.711 (0.004)
Ankle plantarflexion	—	0.022 (0.025)	0.005 (0.415)	0.023 (0.025)
Ankle dorsiflexion	Log	0.432 (0.007)	0.327 (0.090)	0.716 (0.004)
<i>Peak power</i>				
H1 _{GENERATION}	Log	< 0.001 (0.107)	0.715 (0.011)	0.073 (0.016)
H2 _{ABSORPTION}	Square root	< 0.001 (0.078)	0.234 (0.120)	0.974 (0.001)
H3 _{GENERATION}	Log	< 0.001 (0.091)	0.118 (0.157)	< 0.001 (0.099)
K1 _{ABSORPTION}	Square root	0.161 (0.014)	0.951 (0.000)	0.323 (0.009)
K2 _{GENERATION}	Log	< 0.001 (0.052)	0.286 (0.029)	0.558 (0.006)
K3 _{ABSORPTION}	Square root	< 0.001 (0.129)	0.030 (0.474)	< 0.001 (0.050)
K4 _{ABSORPTION}	—	< 0.001 (0.489)	0.190 (0.141)	0.330 (0.009)
A1 _{ABSORPTION}	Log	< 0.001 (0.079)	0.012 (0.403)	0.220 (0.012)
A2 _{GENERATION}	Log	< 0.001 (0.132)	0.612 (0.022)	0.012 (0.028)

4. 3. 2. Joint angular velocity

EXO showed a significant main effect on both peak hip extension and flexion velocities (Table 1). *Post hoc* analysis showed that peak hip extension velocity was least in the NoExo condition and highest in the EXO_{HIGH} condition (Figure 4-4). In contrast, peak hip flexion velocity was higher in the NoExo condition compared to other conditions and was also higher in EXO_{OFF} and EXO_{LOW} than in EXO_{HIGH} (Figure 4-4). While peak knee flexion velocity did not show any significant main or interaction effects, significant main effect of *EXO* was observed in peak knee extension velocity (Table 1). Peak knee flexion velocity was

higher in NoExo compared to EXO_{OFF} and EXO_{HIGH} conditions, and was also higher in EXO_{OFF} and EXO_{LOW} compared to EXO_{HIGH} condition (Figure 4-4).

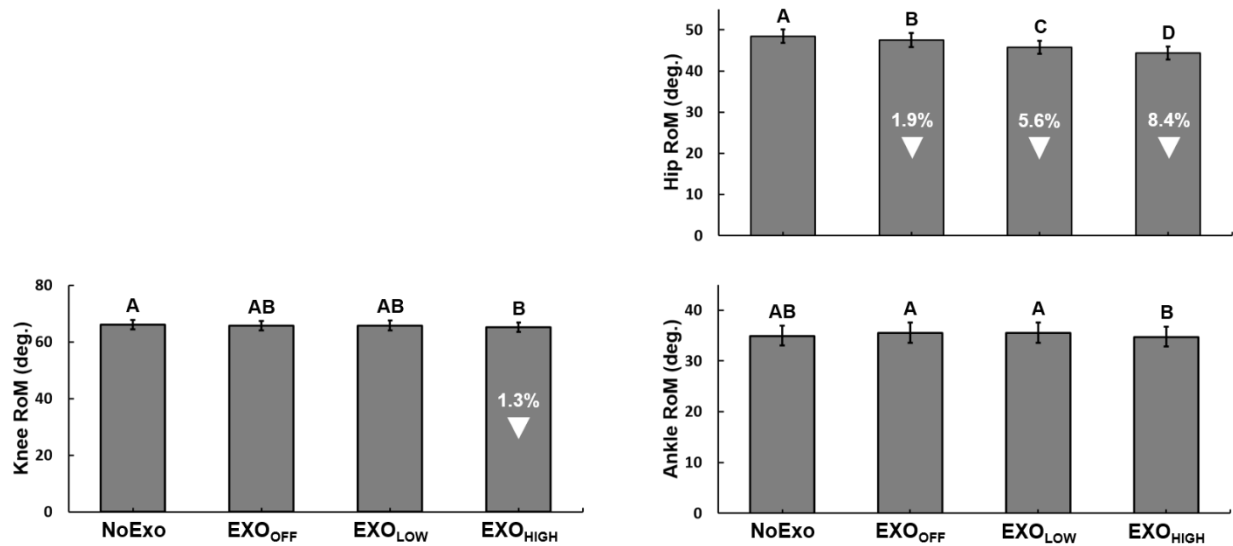


Figure 4-3. Hip, knee, and ankle RoM in four *EXO* conditions during level walking. Data are presented as least-squares means with error bars indicating 95% confidence intervals. Upper case letters specify groupings obtained from pairwise comparisons between *EXO* conditions. The inverted triangle symbols and accompanying numbers indicate a significant decrease from the control no-exoskeleton condition, and the corresponding percentage change in dependent measure.

Both the main effect of *EXO* and the interaction effect of *EXO*×*GEN* were significant for both peak ankle plantarflexion and dorsiflexion velocities (Table 1). Specifically, peak ankle plantarflexion velocity increased in all exoskeleton conditions compared to the NoExo condition (Figure 4-4). Simple effects analysis showed that such changes were significant for both males and females (all p -values<0.05), but females showed relatively more increase (6-7%) than males (3-4%). Peak ankle dorsiflexion velocity was least in NoExo, higher in EXO_{OFF}, still higher in EXO_{LOW}, and highest in EXO_{HIGH} (Figure 4-4). Peak ankle dorsiflexion velocity was lower in NoExo than in EXO_{LOW} ($p=0.001$) and EXO_{HIGH} ($p<0.001$), and was also lower in EXO_{OFF} than in EXO_{HIGH} ($p<0.001$) among males. However, peak ankle dorsiflexion velocity was lower in the NoExo condition compared to other conditions (p -values<0.02), and was also lower in EXO_{OFF} and EXO_{LOW} than in EXO_{HIGH} (p -values<0.03) among females. A significant main effect of *GEN* was seen

in peak hip flexion, peak ankle plantarflexion, and peak ankle dorsiflexion velocities (Table 1), and all were higher among females.

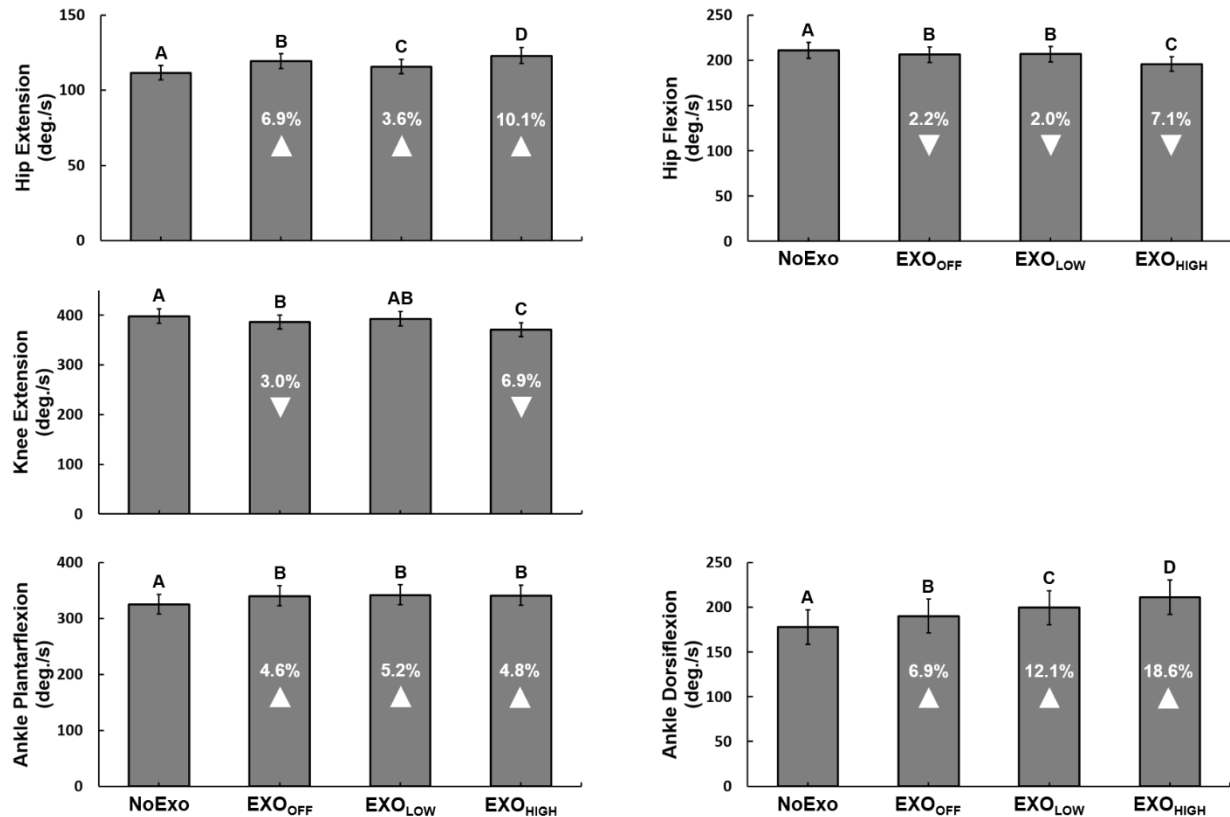


Figure 4-4. Peak angular velocities of hip, knee, and ankle in four *EXO* conditions during level walking. Data are presented as least-squares means with error bars indicating 95% confidence intervals. Upper case letters specify groupings obtained from pairwise comparisons between *EXO* conditions. The triangle (or inverted triangle) symbols and accompanying numbers indicate a significant increase (or decrease) from the control no-exoskeleton condition, and the corresponding percentage change in dependent measure.

4. 3. 3. Joint torque

Significant main effects of *EXO* were observed on both peak hip extension and flexion torques (Table 1). Specifically, peak hip extension torque was lower in *EXO_{HIGH}* condition compared to other conditions, while peak hip flexion torque was lower in *EXO_{LOW}* compared to no exoskeleton control condition (Figure 4-5). The main effect of *EXO* was significant for both peak knee extension and flexion torques (Table 1). *Post hoc* analysis revealed that peak knee extension torque was higher in *EXO_{OFF}* than in *EXO_{HIGH}*.

However, peak knee flexion torque was higher in control condition as compared to other conditions, and was also higher in EXO_{OFF} than in EXO_{LOW} and EXO_{HIGH} (Figure 4-5).

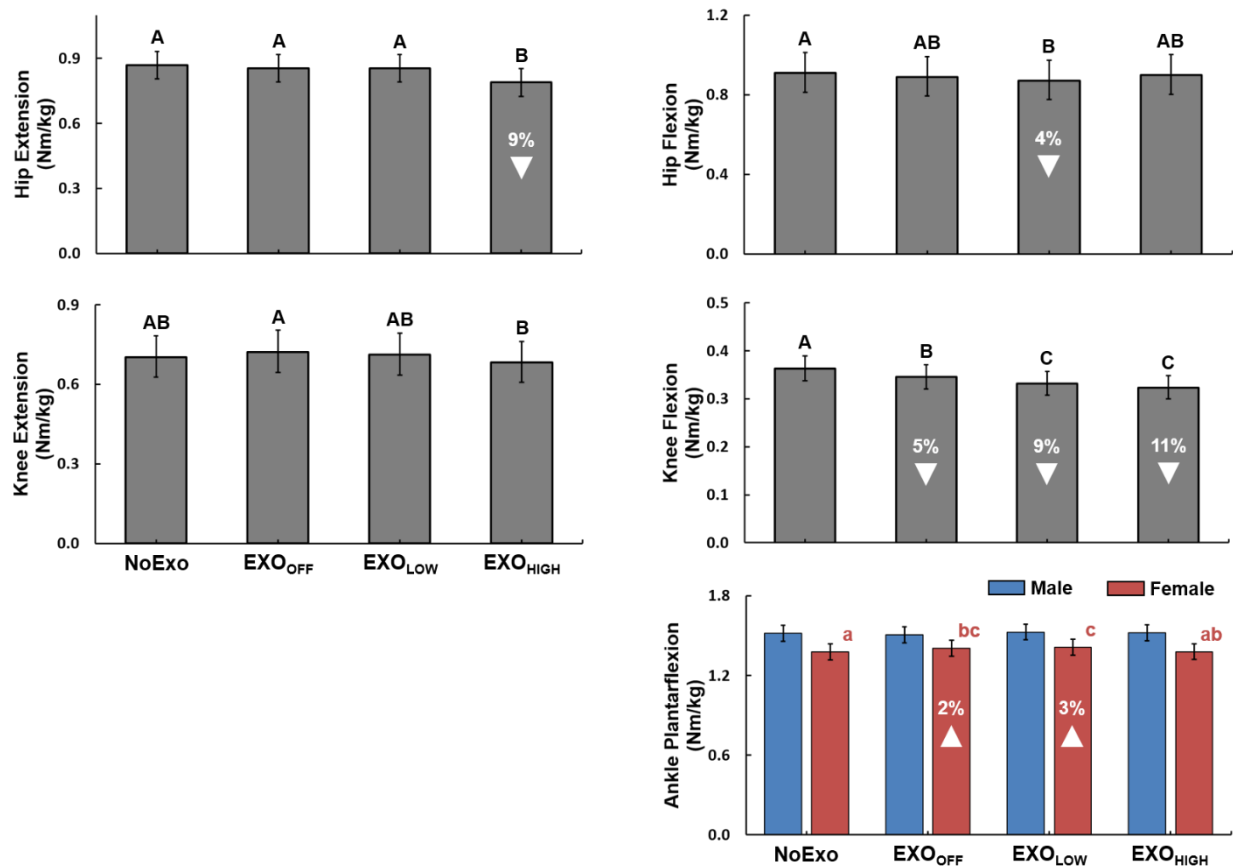


Figure 4-5. Peak torques at hip, knee, and ankle in four *EXO* conditions during level walking. Data are presented as least-squares means with error bars indicating 95% confidence intervals. Upper case letters specify groupings obtained from pairwise comparisons between *EXO* conditions. Lower case letters show results from pairwise comparisons between *EXO* conditions within *GEN*. The triangle (or inverted triangle) symbols and accompanying numbers indicate a significant increase (or decrease) from the control no-exoskeleton condition, and the corresponding percentage change in dependent measure.

While peak ankle dorsiflexion torque did not show any significant effects, both the main effect of *EXO* and the *EXO*×*GEN* interaction effect were significant for peak ankle plantarflexion torque (Table 1). Peak ankle plantarflexion torque was lower in the control condition than EXO_{LOW}. While males did not show any significant differences between *EXO* conditions, peak ankle plantarflexion torque was lower in control condition than in both EXO_{OFF} and EXO_{LOW}, and was also lower in EXO_{HIGH} compared to EXO_{LOW}.

condition among females (all p -values<0.05; Figure 4-5). A significant main effect of *GEN* was found on peak knee flexion and peak ankle plantarflexion torques (Table 1), and both were higher among males.

4. 3. 4. Joint power

EXO had significant main effect on both $H1_{\text{GENERATION}}$ and $H2_{\text{ABSORPTION}}$ peak powers (Table 1). Specifically, $H1_{\text{GENERATION}}$ peak power was higher in EXO_{OFF} compared to other conditions, and was lower in EXO_{HIGH} than in other conditions (Figure 4-6). $H2_{\text{ABSORPTION}}$ peak power was higher in NoExo than in EXO_{OFF} and EXO_{LOW} , and was also higher in EXO_{HIGH} compared to EXO_{OFF} (Figure 4-6). Both the main effect of *EXO* and the *EXO*×*GEN* interaction effect were significant for $H3_{\text{GENERATION}}$ peak power. Although $H3_{\text{GENERATION}}$ peak power was lower in both NoExo and EXO_{OFF} conditions than EXO_{LOW} and EXO_{HIGH} , simple effects analysis showed that these differences were significant only among females (all p -values<0.001), whereas $H3_{\text{GENERATION}}$ peak power in males was lower in EXO_{OFF} compared to NoExo (p <0.001) and EXO_{HIGH} (p =0.032) conditions (Figure 4-6).

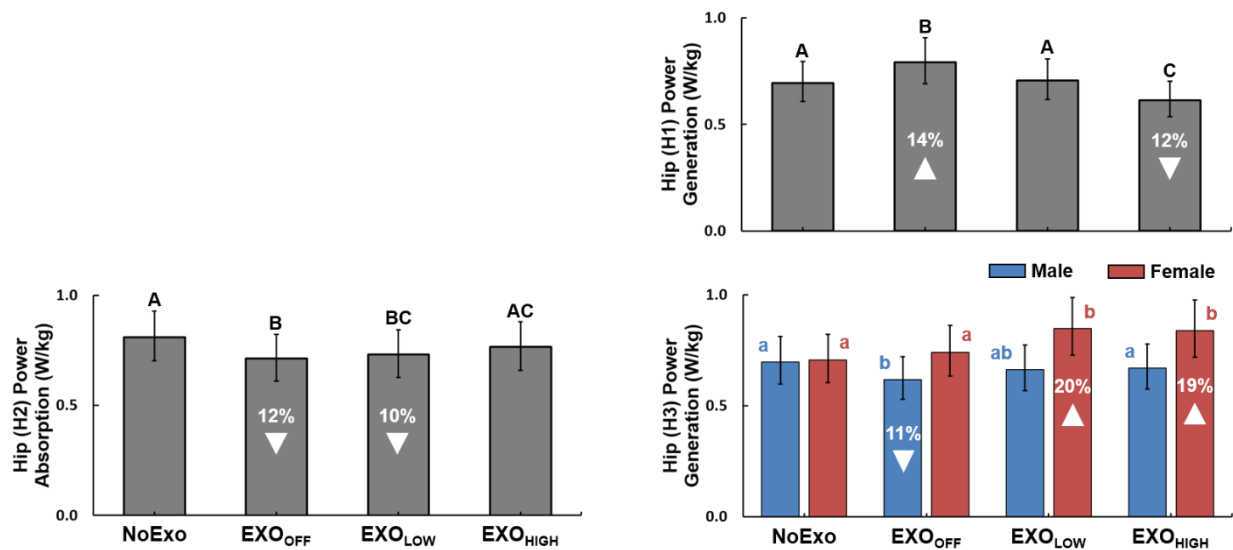


Figure 4-6. Peak hip powers in four *EXO* conditions during level walking. Data are presented as least-squares means with error bars indicating 95% confidence intervals. Upper case letters specify groupings obtained from pairwise comparisons between *EXO* conditions. Lower case letters show results from pairwise comparisons between *EXO* conditions within *GEN*. The triangle (or inverted triangle) symbols and accompanying numbers indicate a significant increase (or decrease) from the control no-exoskeleton condition, and the corresponding percentage change in dependent measure.

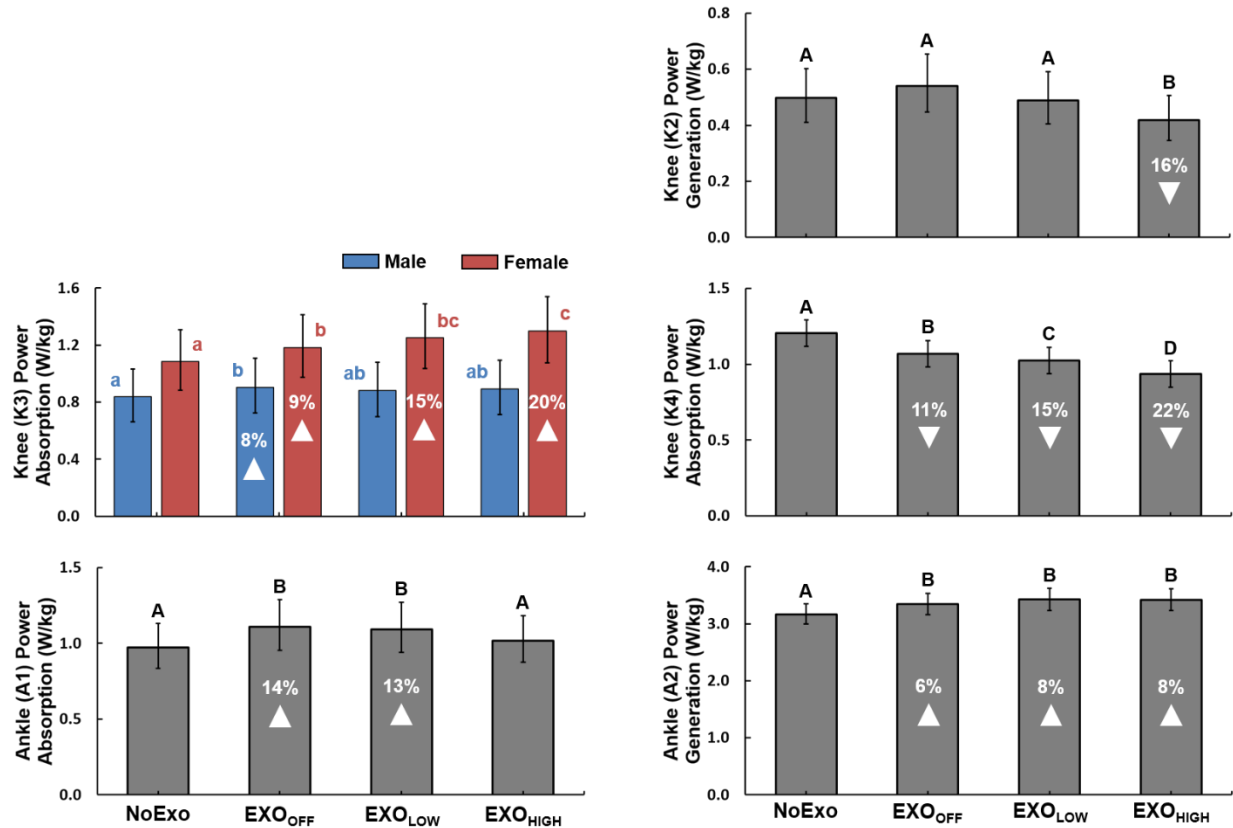


Figure 4-7. Peak knee and ankle powers in four *EXO* conditions during level walking. Data are presented as least-squares means with error bars indicating 95% confidence intervals. Upper case letters specify groupings obtained from pairwise comparisons between *EXO* conditions. Lower case letters show results from pairwise comparisons between *EXO* conditions within *GEN*. The triangle (or inverted triangle) symbols and accompanying numbers indicate a significant increase (or decrease) from the control no-exoskeleton condition, and the corresponding percentage change in dependent measure.

While $K1_{\text{GENERATION}}$ peak power did not show any significant main or interaction effects, significant main effects of *EXO* were observed in both $K2_{\text{GENERATION}}$ and $K4_{\text{ABSORPTION}}$ peak powers (Table 1). *Post hoc* analysis showed that $K2_{\text{GENERATION}}$ peak power was lower in EXO_{HIGH} compared to other conditions (Figure 4-7). $K4_{\text{ABSORPTION}}$ peak power was higher in control condition compared to other conditions, was higher in EXO_{OFF} than in EXO_{LOW} and EXO_{HIGH} , was also higher in EXO_{LOW} than in EXO_{HIGH} (Figure 4-7). Both the main effect of *EXO* and the *EXO*×*GEN* interaction effect were significant for $K3_{\text{ABSORPTION}}$ peak power (Table 1). Specifically, $K3_{\text{ABSORPTION}}$ peak power was lower in control condition compared to other conditions (Figure 4-7). However, simple effect analysis revealed that $K3_{\text{ABSORPTION}}$ peak power was lower in control condition than in EXO_{OFF} only among males ($p=0.021$), whereas $K3_{\text{ABSORPTION}}$ peak power in

females was lower in control condition compared to other conditions, and was also lower in EXO_{OFF} than in EXO_{HIGH} (all p -values<0.005).

Significant main effects of *EXO* were shown in both A1_{ABSORPTION} and A2_{GENERATION} peak powers (Table 1). Specifically, A1_{ABSORPTION} peak power was lower in both NoExo and EXO_{HIGH} conditions than in EXO_{OFF} and EXO_{LOW} (Figure 4-7). A2_{GENERATION} peak power was lower in control condition compared to other conditions (Figure 4-7). Simple effects analysis of the *EXO*×*GEN* interaction showed that such differences were significant for both males and females (all p -values<0.05), but the difference in A2_{GENERATION} peak power between NoExo and EXO_{OFF} was not significant for males (p =0.216). A significant main effect of *GEN* was found on K3_{ABSORPTION} and A1_{ABSORPTION} peak powers (Table 1), and both were higher among males.

4. 4. Discussion

4. 4. 1. Changes in hip joint

Wearing the backXTM with high supportive torque (EXO_{HIGH}) led to a decrease in peak hip extension torque (9%) compared to the NoExo Condition (Figure 4-5). From Figure C-1, it can be observed that these changes occurred primarily while the hip was flexed more than ~20° (i.e., in the initial 0-10% and final 75-100% of the gait cycle). During these phases, the backXTM generated substantial external hip extension torque, and hence participants decreased their generated hip extension torque accordingly, such that the total extension torque generated by participants + BSE did not differ across exoskeleton conditions (p =0.26). These results suggest that participants adapted their walking patterns to the given exoskeleton setting.

However, the EXO_{HIGH} condition also led to a decrease in hip RoM (8%) compared to the NoExo condition (Figure 4-3), by decreasing both peak hip extension (13%, p <0.001) and flexion (7%, p <0.001) angles. Furthermore, our results on hip RoM (i.e., decrease in hip flexion angle) and reduced peak hip flexion velocity suggest impeded hip flexion due to wearing the exoskeleton (elaborated more below). However, peak hip flexion torque did not differ between NoExo and EXO_{HIGH} conditions (Figure 4-5). Figure C-1

shows that, during the times when hip flexion torque contributes to gait (i.e., during 15-70% of the gait cycle), the hip was flexed less than $\sim 10^\circ$ or mostly extended. At this low degree of flexion, the backXTM did not generate substantial external hip extension torque for that combination of angle and angular velocity (Figure 4-2). Hence, wearing the backXTM had minimal effect on flexion torque at the hip during walking. It is important to note, however, that these results might not be similar for other activities such as obstacle crossing, walking upstairs, or stepping for balance recovery from perturbations, in which hip flexion torque is typically generated when the hip is significantly more flexed than while walking.

There was a 12-22% decrease in H1_{GENERATION} peak power in the EXO_{HIGH} condition (Figure 4-6). This can also be explained by the adapted walking patterns (i.e., decrease in peak hip extension torque) when the backXTM generated substantial hip extension torque. There was, however, a 11-19% increase in H1_{GENERATION} peak power in the EXO_{OFF} condition (Figure 4-6). In prior work, DeVita and Hortobagyi (2000) reported that older adults walked with an increase in H1_{GENERATION} power compared to young adults, and that this was an altered motor strategy that occurred with old age since the increase in H1_{GENERATION} was compensated by a corresponding decrease in ankle power, thereby keeping the overall power to be constant but changing the distribution of power generation across the lower-limb joints. In our case, increase in H1_{GENERATION} in the EXO_{OFF} condition could not be interpreted as an altered motor strategy, because contrary to DeVita and Hortobagyi (2000), we also observed an increase in A1_{ABSORPTION} power (Figure 4-7). Thus, this may be because participants generated more H1 power when the backXTM was worn but was not generating any hip extension torque, to compensate for the added mass and inertial effects of the device during the stance phase. Furthermore, a decrease (10-12%) in H2_{ABSORPTION} peak power in EXO_{OFF} and EXO_{LOW} may also indicate synchronized changes with H1_{GENERATION} peak power.

H3_{GENERATION} peak power increased in both EXO_{LOW} and EXO_{HIGH} (7-11%) conditions, compared to NoExo and EXO_{OFF} conditions (Figure 4-6). Given that hip flexor power (H3) during gait is primarily generated to propel the leg into the swing phase by adding mechanical energy to the limb (Eng and Winter, 1995; Sadeghi et al., 1997), participants may have increased H3_{GENERATION} power in both EXO_{LOW} and EXO_{HIGH}

conditions, in anticipation of external hip extension torque that will be provided by the backXTM from the ~65% of the gait cycle (i.e., when hip is flexed; Figure C-1). We assume that the familiarization procedure (i.e., ~5 min. walking) provided for each *EXO* condition may have helped participants to learn when the external torque is engaged during walking.

4. 4. 2. Changes in knee joint

EXO_{HIGH} led to a slight (6%) decrease in peak knee extension torque (Figure 4-5); this change may reflect the co-dependence between the hip and knee as they go into extension during the initial phase of gait. Since peak knee extension torque occurred at about 10% of gait cycle (Figure C-1), decrease in peak knee extension torque may have subsequently decreased *K2_{GENERATION}* peak power (15-23%) in the *EXO_{HIGH}* condition (Figure 4-7). Previous studies found that *K2_{GENERATION}* power is mainly produced by concentric contraction of the quadriceps to extend the knee prior to heel-off (Sadeghi et al., 1997; Sadeghi et al., 2001), which is in agreement with our results that the knee was 15-20% less extended prior to heel-off (i.e., at ~40% of the gait cycle) in *EXO_{HIGH}* compared to *NoExo* ($p<0.001$) and *EXO_{OFF}* ($p<0.001$).

K3_{ABSORPTION} peak power increased by 9-13% in all exoskeleton conditions (Figure 4-7). The role of *K3_{ABSORPTION}* power is to prevent knee collapse in late stance and early swing through eccentric contraction of the rectus femoris (Eng and Winter, 1995). Hence, increase in *K3_{ABSORPTION}* peak power indicates that wearing the backXTM led to greater energy transfer from the knee to the hip at around 60% of the gait cycle, resulting in increased hip flexor power to propel the leg into swing (Neptune et al., 2008). Our results also show that *H3_{GENERATION}* peak power increased in *EXO_{LOW}* and *EXO_{HIGH}* conditions compared to the control condition (Figure 4-6).

Compared to the control condition, wearing the backXTM decreased peak knee flexion torque by 5-11%. *EXO_{LOW}* and *EXO_{HIGH}* also led to decrease in peak knee flexion torque, compared to the *EXO_{OFF}* condition (Figure 4-5). We believe that the adapted walking patterns (i.e., decrease in peak hip extension torque), when the backXTM generated hip extension torque, resulted in a synchronized change in the knee, through

a decrease in peak knee flexion torque. This can be further supported by the fact that the decreased activation in biceps femoris (a biarticular muscle) to decrease hip extension torque can subsequently lead to decrease in knee flexion torque during walking. Knee flexion torque, especially during terminal swing, has a role to reduce knee extension velocity, resulting in energy absorption (Sloot and van der Krogt, 2018). This was also observed in our results, that peak knee extension velocity decreased in EXO_{OFF} and EXO_{HIGH} than in NoExo condition (Figure 4-4), and that wearing the BSE substantially decreased K4_{ABSORPTION} peak power (by 11-22%) compared to the NoExo condition, and that the change in K4_{ABSORPTION} peak power increased with the magnitude of external device torque (Figure 4-7).

4. 4. 3. Changes in ankle joint

The EXO_{OFF} and EXO_{LOW} conditions slightly increased peak ankle plantarflexion torque (by 2% and 3% respectively), compared to the control condition, but only among females, and the EXO_{HIGH} decreased peak ankle plantarflexion torque by 2% compared to the EXO_{LOW} condition, whereas males did not present such differences between EXO conditions (Figure 4-5). Ankle plantarflexion torque is the most important energy generation source during the gait cycle (Bovi et al., 2011). However, whether this 2-3% increase among females is practically meaningful (albeit statistically significant), and what its long-term consequences may be, are not currently clear.

The EXO_{OFF} and EXO_{LOW} conditions caused 8-14% increase in A1_{ABSORPTION} peak power compared to both NoExo and EXO_{HIGH} (Figure 4-7). As discussed in the hip joint section, this may be because participants needed to absorb more A1 power when the backXTM was worn but generating no or low torque (EXO_{OFF} and EXO_{LOW} conditions), to compensate for the added mass and inertial effects of the device. Increased A1_{ABSORPTION} peak power can lead to an increase in A2_{GENERATION} peak power, as A2_{GENERATION} power is mainly produced by the release of the elastic energy stored during the A1 absorption period (Stewart et al., 2007).

All exoskeleton conditions led to increase in both peak ankle plantarflexion (4-5%) and dorsiflexion (7-19%) velocities compared to control condition (Figure 4-4). Such increases may have caused by increase in $A2_{\text{GENERATION}}$ peak power (6-8%) in all exoskeleton conditions (Figure 4-7). $A2_{\text{GENERATION}}$ power, produced by the ankle plantar flexors (during push-off), provides the majority of the total mechanical work during gait, and thereby has an important role for forward progression of the body and leg swing initiation (DeVita and Hortobagyi, 2000; Winter, 2009). Hence, we consider that increased $A2_{\text{GENERATION}}$ peak power may indicate synchronized changes with increased $K3_{\text{ABSORPTION}}$ and $H3_{\text{GENERATION}}$ peak powers when wearing the backXTM. This suggests that wearing the backXTM leads to more mechanical energy generation (by hip and ankle) and absorption (by knee) at ~60% gait cycle to swing forward, and explains why earlier studies may have reported that wearing a BSE increased metabolic costs during walking (Baltrusch et al., 2019).

4. 4. 4. Gender differences

A significant $EXO \times GEN$ interaction effect was shown in peak ankle plantarflexion torque, $H3_{\text{GENERATION}}$, $K3_{\text{ABSORPTION}}$, and $A2_{\text{GENERATION}}$ peak powers. Overall, the effects of the external torque level of the backXTM on these measures were larger among females (Figures 4-5, 4-6, and 4-7), indicating that females generate and absorb relatively more mechanical energy for level walking while wearing a BSE, compared to males. Given that there were no significant gender differences in anthropometric measures (i.e., stature, mass, chest circumference, waist-hip ratio, and thigh circumference) and walking speed, gender differences in strength (e.g., maximum hip extension torque) may have influenced our findings. While previous investigations have shown gender differences in lower limb kinetics during walking (e.g., Cho et al., 2004; Rowe et al., 2021), further gender differences when wearing a BSE need to be clarified in follow-up studies.

4. 4. 5. Limitations and future work

This study has some limitations common with most laboratory-based research conducted using exoskeletons such as individual differences in BSE fit/adjustments may have influenced our findings, we focused only on the short-term effects of BSE use, and included only rather healthy and young individuals.

Level walking was studied using only straight-walking conditions. In addition, specifically, the backX™ torque profile was obtained using a dynamometer and a manikin, over 0-120 deg. of hip angle, covering the entire range of hip excursions that are possible during gait, but only at limited, discrete speeds in steps of 20 deg./s. Thus, there may have been some errors in our torque estimates at the interpolated velocities. Additionally, lower limb muscle activities were not measured in this study. Hence, while our interpretations on the results of joint kinematics and kinetics are valuable, changes in muscle coordination strategies or co-contractions could not be investigated. Future work extending kinetic analyses to other tasks that may be encountered in a workplace, such as change in directions during level walking, obstacle crossing, stairclimbing, or recovering from perturbations using step recovery, i.e., activities that require greater hip flexion than level walking, would offer more complete and compelling evidence of how a BSE may impact postural control and fall risk.

4. 4. 6. Conclusions

We evaluated the effects of different external torque assistance levels of a BSE on lower limb joint kinematics and kinetics during level walking, among a young and healthy participant group. Overall, wearing the backX™ impeded hip flexion by decreasing hip RoM and hip flexion velocity. When hip extension torques were required for walking (i.e., the initial 0-10% and final 75-100% of the gait cycle), the backX™ with high supportive torque generated substantial external hip extension torque, and participants showed adapted gait patterns (i.e., decrease in hip extension torque and extensor power), and this in turn may have led to decrease in knee extension velocity, flexion torque, and extensor power. However, participants increased hip extensor power and ankle dorsiflexor power, while the backX™ did not provide high external torque, to compensate for the added mass and inertial effects of the device during the stance phase. When hip flexion torques were required for walking (i.e., the middle 15-70% of the gait cycle), although the backX™ with high supportive torque did not generate substantial hip extension torque, participants showed changes in lower limb kinematics (i.e., increase in ankle plantarflexion and dorsiflexion velocities) and kinetics (i.e., increase in hip flexor, knee extensor, and ankle plantarflexor powers) to

increase mechanical energy generation for propelling the leg into the swing phase, probably in anticipation of external hip extension torque provided by the device from ~70% of the gait cycle. Hence, wearing a back-support exoskeleton may increase metabolic costs during walking. Whether this further leads to muscle fatigue and/or postural instability in long-distance walking is to be confirmed in future work.

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Appendix C

Table C-1. Least squares mean (95% confidence intervals) of all outcome measures.

Outcome measures	EXO condition			
	NoExo	EXO _{OFF}	EXO _{LOW}	EXO _{HIGH}
<i>RoM (deg.)</i>				
Hip	48.43 (46.78, 50.13)	47.49 (45.87, 49.16)	45.70 (44.15, 47.30)	44.35 (42.84, 45.91)
Knee	66.10 (64.45, 67.79)	65.80 (64.15, 67.48)	65.86 (64.22, 67.55)	65.24 (63.61, 66.91)
Ankle	34.92 (33.02, 36.94)	35.51 (33.57, 37.56)	35.51 (33.57, 37.56)	34.70 (32.80, 36.70)
<i>Peak angular velocity (deg./s)</i>				
Hip extension	111.56 (106.89, 116.42)	119.21 (114.23, 124.42)	115.61 (110.77, 120.66)	122.86 (117.73, 128.22)
Hip flexion	210.96 (202.42, 219.86)	206.28 (197.93, 214.99)	206.82 (198.45, 215.55)	195.93 (188.00, 204.20)
Knee extension	398.11 (383.63, 413.13)	386.13 (372.08, 400.70)	392.82 (378.53, 407.65)	370.71 (357.23, 384.69)
Knee flexion	332.42 (320.83, 344.43)	334.26 (322.60, 346.34)	332.19 (320.61, 344.20)	329.35 (317.87, 341.25)
Ankle plantarflexion	325.23 (308.56, 342.80)	340.27 (322.83, 358.65)	342.04 (324.50, 360.52)	340.93 (323.46, 359.35)
Ankle dorsiflexion	177.99 (158.98, 197.01)	190.21 (171.19, 209.23)	199.48 (180.46, 218.50)	211.17 (192.16, 230.19)
<i>Peak torque (Nm/kg)</i>				
Hip extension	0.87 (0.80, 0.93)	0.85 (0.79, 0.92)	0.85 (0.79, 0.92)	0.79 (0.73, 0.85)
Hip flexion	0.91 (0.81, 1.01)	0.89 (0.79, 0.99)	0.87 (0.78, 0.97)	0.90 (0.80, 1.00)
Knee extension	0.70 (0.63, 0.78)	0.72 (0.65, 0.80)	0.71 (0.64, 0.79)	0.68 (0.61, 0.76)
Knee flexion	0.36 (0.34, 0.39)	0.35 (0.32, 0.37)	0.33 (0.31, 0.36)	0.32 (0.30, 0.35)
Ankle plantarflexion	1.45 (1.41, 1.49)	1.46 (1.41, 1.50)	1.47 (1.43, 1.51)	1.45 (1.41, 1.49)
Ankle dorsiflexion	0.17 (0.15, 0.19)	0.16 (0.14, 0.18)	0.16 (0.14, 0.18)	0.16 (0.14, 0.18)
<i>Peak power (W/kg)</i>				
H1 _{GENERATION}	0.70 (0.61, 0.80)	0.79 (0.69, 0.91)	0.71 (0.62, 0.81)	0.61 (0.54, 0.70)
H2 _{ABSORPTION}	-0.81 (-0.93, -0.70)	-0.71 (-0.82, -0.61)	-0.73 (-0.84, -0.63)	-0.76 (-0.88, -0.66)
H3 _{GENERATION}	0.70 (0.63, 0.78)	0.68 (0.61, 0.75)	0.75 (0.67, 0.84)	0.75 (0.67, 0.83)
K1 _{ABSORPTION}	-0.81 (-0.97, -0.67)	-0.87 (-1.03, -0.72)	-0.87 (-1.03, -0.72)	-0.86 (-1.02, -0.72)
K2 _{GENERATION}	0.50 (0.41, 0.60)	0.54 (0.45, 0.65)	0.49 (0.40, 0.59)	0.42 (0.35, 0.51)
K3 _{ABSORPTION}	-0.96 (-1.10, -0.82)	-1.04 (-1.19, -0.90)	-1.06 (-1.21, -0.92)	-1.09 (-1.24, -0.94)
K4 _{ABSORPTION}	-1.21 (-1.29, -1.12)	-1.07 (-1.15, -0.98)	-1.02 (-1.11, -0.94)	-0.94 (-1.02, -0.85)
A1 _{ABSORPTION}	-0.97 (-1.13, -0.84)	-1.11 (-1.29, -0.95)	-1.09 (-1.27, -0.94)	-1.02 (-1.18, -0.87)
A2 _{GENERATION}	3.17 (2.99, 3.35)	3.34 (3.16, 3.53)	3.42 (3.24, 3.62)	3.42 (3.23, 3.62)

Table C-2. Summary of *post hoc* pairwise comparisons (Tukey's HSD tests) for the significant main effects of *EXO* on each of the outcome measures. Table entries are *p* values, and significant differences are highlighted in bold ($p < 0.05$).

Outcome measures	<i>EXO</i> condition					
	NoExo vs. EXO _{OFF}	NoExo vs. EXO _{LOW}	NoExo vs. EXO _{HIGH}	EXO _{OFF} vs. EXO _{LOW}	EXO _{OFF} vs. EXO _{HIGH}	EXO _{LOW} vs. EXO _{HIGH}
<i>RoM</i>						
Hip	< 0.001	< 0.001	< 0.001	< 0.001	< 0.001	< 0.001
Knee	0.613	0.782	0.003	0.993	0.108	0.055
Ankle	0.125	0.119	0.822	1.000	0.012	0.012
<i>Peak angular velocity</i>						
Hip extension	< 0.001	0.004	< 0.001	0.019	0.022	< 0.001
Hip flexion	0.015	0.041	< 0.001	0.985	< 0.001	< 0.001
Knee extension	< 0.001	0.248	< 0.001	0.083	< 0.001	< 0.001
Ankle plantarflexion	< 0.001	< 0.001	< 0.001	0.900	0.994	0.972
Ankle dorsiflexion	0.002	< 0.001	< 0.001	0.029	< 0.001	0.003
<i>Peak torque</i>						
Hip extension	0.561	0.575	< 0.001	1.000	< 0.001	< 0.001
Hip flexion	0.505	0.026	0.898	0.481	0.899	0.146
Knee extension	0.500	0.914	0.421	0.881	0.019	0.131
Knee flexion	0.004	< 0.001	< 0.001	0.035	< 0.001	0.387
Ankle plantarflexion	0.758	0.025	0.992	0.254	0.897	0.054
<i>Peak power</i>						
H1 _{GENERATION}	0.004	0.979	0.006	0.014	< 0.001	0.002
H2 _{ABSORPTION}	< 0.001	< 0.001	0.080	0.758	0.025	0.259
H3 _{GENERATION}	0.295	0.006	0.008	< 0.001	< 0.001	1.000
K2 _{GENERATION}	0.466	0.992	0.017	0.308	< 0.001	0.039
K3 _{ABSORPTION}	< 0.001	< 0.001	< 0.001	0.769	0.064	0.431
K4 _{ABSORPTION}	< 0.001	< 0.001	< 0.001	0.018	< 0.001	< 0.001
A1 _{ABSORPTION}	0.962	< 0.001	0.341	< 0.001	0.008	0.034
A2 _{GENERATION}	< 0.001	< 0.001	< 0.001	0.191	0.216	1.000

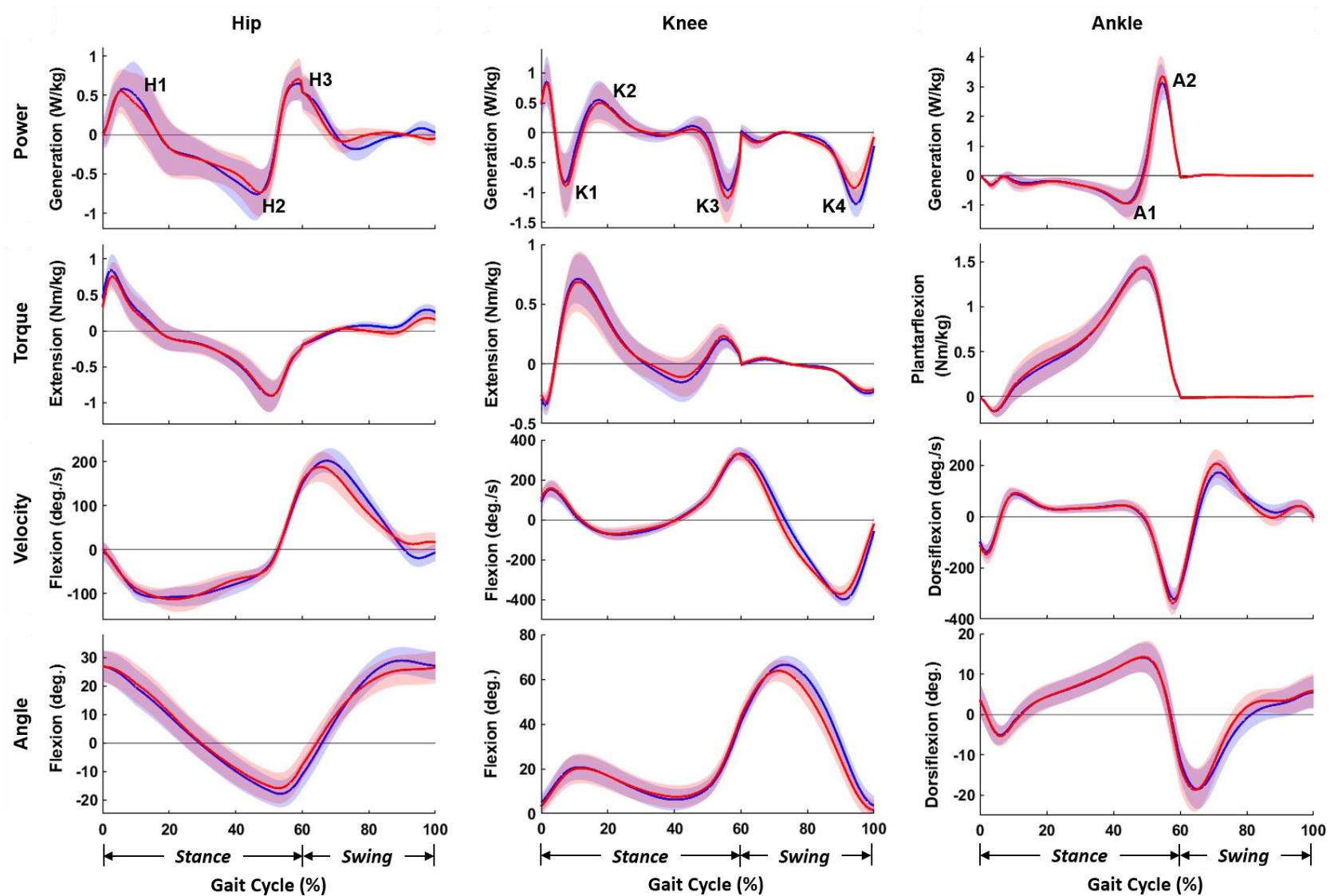


Figure C-1. Mean (solid-line) and standard deviation (shaded-area) of angles, angular velocities, torques, and powers for the hip, knee, ankle in the sagittal plane in NoExo (blue) and EXO_{HIGH} (red) conditions during a gait cycle. Illustrated data for each EXO condition were obtained from five repeats of walking trials of 20 participants.

Chapter 5. Effects of back-support exoskeleton use on single step balance recovery from a forward loss of balance

Abstract

While back-support exoskeletons (BSEs) are a promising ergonomic intervention for reducing the physical demands on the low-back, little is known whether BSE use alters reactive balance following external perturbations. Hence, we investigated the effects of wearing a BSE on single step balance recovery following a forward loss of balance. Sixteen (8M, 8F) young, healthy participants were released from forward-leaning positions and attempted to recover their balance with a single step, while wearing a BSE (backX™) with three different levels of support torque (i.e., no torque, low, and high) and in a control condition (no-exoskeleton). Lean angle was increased until they failed to recover their balance with a single step. Overall, the maximum lean angle from which individuals could successfully recover, following a simulated forward fall, was not significantly altered when wearing the BSE. However, wearing the backX™ under all torque conditions increased reaction times, and impeded hip flexion (i.e., decrease in both peak hip flexion velocity and hip range of motion), resulting in decreased peak knee flexion velocity, knee range of motion, and step length. The margin of stability measures decreased significantly in only the high-torque condition. Overall, our results suggest that use of a back-support exoskeleton, that provides hip extension torque impairs balance recovery. Future work extending kinetic analyses to recovery responses, as well as a study of recovery when responding to slips and trips while walking, would offer a more complete picture of how a BSE may impact dynamic balance recovery following a loss of balance, and help in the formulation of design and safe-use guidelines for industrial exoskeleton adoption.

Keywords: occupational exoskeleton, reactive stepping, balance recovery response, tether-release

5. 1. Introduction

Low back pain (LBP) remains a major workplace health problem (Hoy et al., 2010). Earlier studies have shown that major risk factors for LBP include high and/or repetitive forces on lumbar tissues (Griffith et

al., 2012; Waters et al., 2011). Further support comes from the fact that the LBP is particularly common among workers who are often exposed to heavy lifting (Coenen et al., 2013). Recently, wearable devices, such as occupational back-support exoskeletons (BSEs), have shown promise to control and prevent overexertion related to low back disorders (Nussbaum et al., 2019; Zhu et al., 2021). BSEs are wearable mechanical systems designed to support, augment, and/or assist back extension, by passively or actively producing restorative torques. At present, passive BSEs are relatively more mature for occupational use due to their lower cost, design maturity, and ease of implementation in the workplace (Nussbaum et al., 2019).

BSEs include rigid structures and physical attachment, surrounding or supporting the chest, waist, and/or thigh, to transfer restorative torques to the wearer. Among marketed BSEs, the mass of passive BSEs ranges from 2.8 to 4.5 kg, and the mass of active BSEs ranges from 3.1 to 11 kg (Exoskeleton Report LLC, 2021; Voilqué et al., 2019). Previous investigations have shown that passive BSEs can produce ~35 Nm of extension torque for each hip (Koopman et al., 2019a; Näf et al., 2018). Our earlier work has also shown that as the hip angle gradually varied between 0 and 120 degrees, a passive BSE produced ~28 Nm of extension torque for each hip (Chapter 3). These BSE characteristics, such as their rigid structure, added mass, and the external hip extension torque are expected to place increased demands on the postural control system. Accordingly, some unintended, negative effects associated with BSE use have been reported, such as increased discomfort in the chest and shoulders (Giustetto et al., 2021; Qu et al., 2021), increased quadriceps muscle activity (von Glinski et al., 2019), trunk rigidity (Koopman et al., 2019b), and decreased postural balance in bipedal stance and detriments in walking performance (Park et al., 2021; Park et al., 2020). Based on this prior evidence, we expect that BSE use may also potentially increase the risk of falls.

Occupational falls are a serious problem in terms of injuries, lost work time, and medical/industrial costs (Courtney et al., 2001; Webster, 2000). Data from the Bureau of Labor Statistics show that, in 2019, job-related falls resulted in 800 fatalities and 244,000 nonfatal injuries in the United States (Bureau of Labor Statistics, 2020). Most falls occur while walking due to external perturbations (e.g., trips and slips) (Berg

et al., 1997; Heijnen and Rietdyk, 2016), because they contribute to a loss of balance. Hence, the ability to respond effectively to loss of balance is a critical factor in preventing falls. While postural balance can be recovered by a reactive ankle strategy (i.e., distal to proximal muscle activation) or hip strategy (i.e., proximal to distal muscle activation) when the perturbations are relatively small (Horak and Nashner, 1986; Nashner and McCollum, 1985), a reactive stepping response is required to expand and re-establish base-of-support (BoS) following a large postural perturbation (e.g., trips, slips, pushes, acceleration of the support surface) (Aftab et al., 2016).

Earlier investigations indicate that restricted body movement when wearing a BSE may be associated with unsuccessful balance recovery following postural perturbations. For example, increased constraints on hip and trunk by wearing corsets increased the likelihood of a loss of balance when the upright stance was perturbed (Grüneberg et al., 2004). Another study also used a corset to artificially increase constraints on body movement, and reported that increased stiffness at the hip and trunk negatively influences reactive stepping following a forward loss of balance (Pretty et al., 2019). Moreover, studies on age-related differences in balance recovery have shown that reduced recovery ability in the old is associated with decreased weight transfer time and step velocity (Thelen et al., 1997; Wojcik et al., 1999) and decreased lower limb joint torque (Wojcik et al., 2001).

Based on prior work, balance recovery can be examined using a tether-release protocol in a laboratory-based setting by releasing an individual from a static forward leaning position and having them attempt to recover their balance with a single step (Hsiao-Weckler, 2008). Balance recovery responses are typically quantified using spatiotemporal characteristics of stepping, joint kinematics and kinetics, and ground reaction forces (e.g., Carty et al., 2011; Ringhof et al., 2019). As perturbations are induced unexpectedly in such a tether-release protocol, the direction, amplitude, and speed of compensatory reactions need to be modulated in real time to account for the unpredictable body motion (Maki et al., 2003). Hence, slower and shorter steps are associated with unsuccessful recovery due to an insufficient increase in BoS (Lee et al., 2014; Wojcik et al., 1999). Previous studies have also suggested that greater trunk flexion velocities and

smaller peak knee extensor torques can lead to failed recoveries (Madigan and Lloyd, 2005a; Owings et al., 2001). Furthermore, a pendulum-spring model has similarly indicated that balance recovery can be impaired by deteriorated lower limb flexibility, reaction time, and strength (Hsiao-Wecksler, 2008). Based on this body of literature, a BSE's external torque may affect the ability to successfully flex the hip to generate a quick step-response to recover balance, in reaction to a perturbation. Hence, it is critical to quantify how wearing a BSE affects reactive stepping, trunk movement, and lower limb joint torques during balance recovery following a loss of balance, in order to understand whether BSE use increases the risk of falls.

Thus, the objective of this study was to investigate the effects of wearing a BSE on balance recovery following a forward loss of balance. A tether-release protocol was utilized to simulate a forward fall, and single step was used to recover balance. Based on the characteristics of a BSE (i.e., restricted body movement, added weight, and external torque), we hypothesized that wearing a BSE will negatively affect reactive movements to restore balance following an external perturbation. In addition, we also expected to see a relationship between device torque and recovery responses: the greater the external hip extension torque generated by the BSE, the greater the deterioration expected in step recovery movements.

5. 2. Methods

5. 2. 1. Participants

A convenience sample of 16 young and healthy participants (8 males and 8 females) was recruited from the local university and community. Respective means (SD) of age, body mass, and stature were 28.0 (1.4) years, 72.3 (5.6) kg, and 170.7 (5.7) cm for the males; and 27.5 (5.4) years, 56.1 (2.6) kg, and 164.0 (5.4) cm for the females. All reported being physically active and with no current or recent musculoskeletal disorders or injuries. This research complied with the tenets of the Declaration of Helsinki and was approved by the Institutional Review Board at Virginia Tech. Written informed consent was obtained from each participant prior to data collection.

5. 2. 2. Exoskeleton selection

The backXTM AC (US Bionics Inc., USA), designed to reduce physical demands on the back during forward bending, was used in this study. This exoskeleton has mass of 4.5 kg and consists of a waist strap, chest support, and an external torque generator about each hip that is coupled with the waist strap and the chest support. The four exoskeleton (*EXO*) conditions included in this study were: NoExo (no exoskeleton, control condition), EXO_{OFF} (backXTM with no supportive torque), EXO_{LOW} (backXTM with low supportive torque) and EXO_{HIGH} (backXTM with high supportive torque).

5. 2. 3. Experimental design and procedures

A repeated-measures design was used to assess the effects of the four *EXO* conditions on balance recovery after a forward loss of balance. Each participant completed a single experimental session (~3 hours), during which they performed multiple balance recovery trials in each of the four different *EXO* conditions. The presentation order of *EXO* conditions was counterbalanced across participants using 4×4 Latin Squares. Participants wore athletic shorts/a shirt/shoes (provided by the researchers). In each *EXO* condition, the backXTM was fitted and adjusted to the participant following manufacturer guidelines and a familiarization period (~3-minute for walking and squatting) was provided.

Forward loss of balance was induced by releasing participants from a forward leaning position (initial lean angle = 20°). Participants were asked to recover their balance using a single step of the dominant foot (i.e., the foot used to kick a ball) immediately following release. Successful recoveries were followed by repeated trials at increasing lean angles, in step increments of 2.5°. Failed recoveries were followed by a second trial at the same lean angle. This process was repeated until participants failed to recover their balance for two consecutive trials at the same lean angle (Madigan and Lloyd, 2005b; Thelen et al., 1997). To minimize learning effects, the experimental protocol was begun after a practice session of 10-15 trials of single step recovery in the no-exoskeleton control condition. During the actual experimental session, the order of all conditions was counterbalanced across participants. Five-minute breaks were provided between the different experimental *EXO* conditions.

Participants first stood with feet about shoulder width apart, each foot on a force platform (AMTI, Watertown, MA, USA), with their arms along the sides of the body. They were then asked to lean forward far enough so that their body became straight except for dorsiflexion at the ankles (Figure 5-1 (a)): while one end of a lean support rope was attached to the back of a belt worn by the participant, the other end was held in a releasable clasp affixed to a stable structure. Straight-line posture was visually confirmed using a virtual line drawn between two reflective markers, placed respectively on the right acromion and right lateral malleolus, against a ruler. Verbal instructions to adjust trunk inclination were provided to help bring participants to a straight posture. Tension in the rope was measured using an in-line loadcell (Interface Advanced Force Measurement, Scottsdale, AZ, USA). Lean angle was controlled by adjusting the length of the rope, and the resultant lean angle was directly measured using an inertial measurement unit sensor (Mtw Awinda, Xsens Technologies B.V., the Netherlands) attached on the L1 spinous process of the participant. This was displayed in real-time using a proprietary software program (MT Manager, Xsens Technologies B.V., the Netherlands). Once participants were in position at the correct lean angle, they were released by the experimenter at a random time interval (2-10 s) by manually opening the clasp. To prevent falling to the floor in the event of an unsuccessful balance recovery, a full body safety harness was worn with a lanyard that attached to a sliding track along the ceiling (Figure 5-1). Another loadcell (Cooper Instruments and Systems, Warrenton, VA, USA) mounted in-line with the lanyard was used to measure the force applied to the harness. Three criteria were used to define failed recovery (Madigan and Lloyd, 2005b; Thelen et al., 1997): 1) when more than one step was taken by the dominant foot, 2) when more than 30% body weight (BW) force was applied to the harness at any point during recovery, and 3) when the trailing foot took a step longer than 30% of the participant's body height. Another force platform (Bertec Corporation, Columbus, OH, USA) was placed directly in front such that in the event of a successful recovery, the leading foot would land on it, as shown in Figure 5-1 (b).

5. 2. 4. Instrumentation and outcome measures

Whole body kinematics, ground reaction forces, and loadcell forces were synchronized and collected during balance recovery trials. Reflective markers (n=33) and four rigid marker clusters (Figure B-1) were sampled at 120 Hz using a 12-camera optical motion capture system (Qualisys, Inc., Gothenburg, Sweden), then low-pass filtered (6 Hz cutoff; 4th-order Butterworth; bidirectional) (Nevisipour and Honeycutt, 2020).

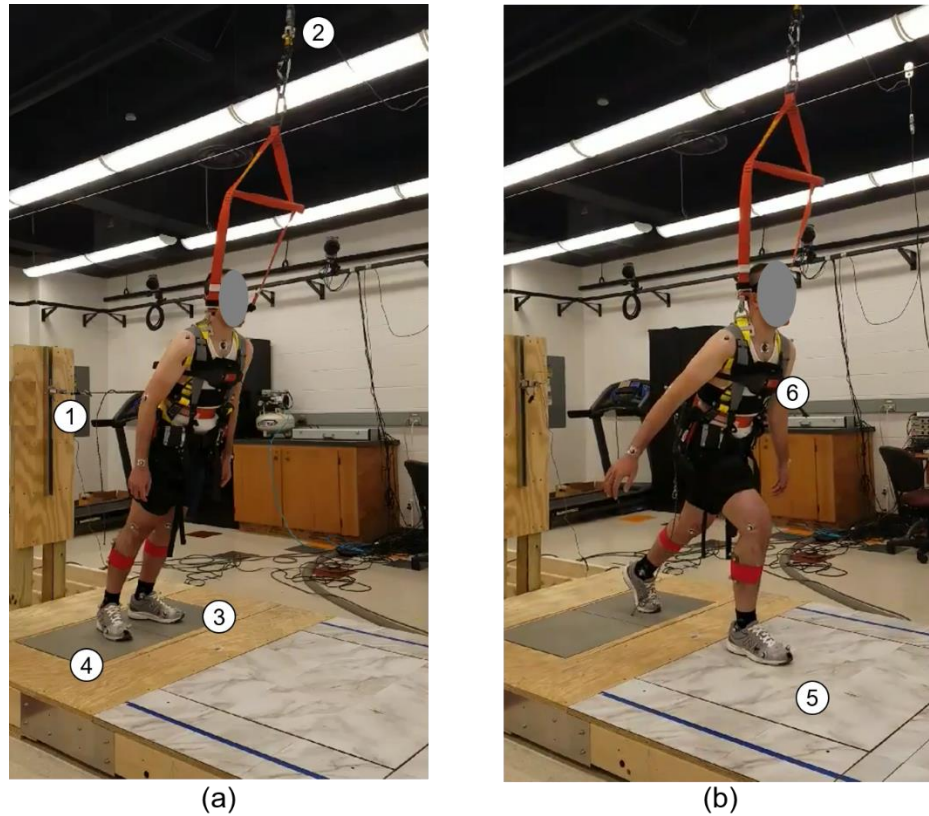


Figure 5-1. Illustration of the experimental setup: (a) tethered lean and (b) balance recovery after release (① in-line loadcell, ② harness loadcell, ③ left force platform, ④ right force platform, ⑤ front force platform, and ⑥ backXTM exoskeleton).

Triaxial ground reaction forces from three force platforms and uniaxial force from two loadcells were sampled at 1200 Hz, subsequently down-sampled to 120 Hz, and low-pass filtered (10 Hz cutoff; 4th-order Butterworth; bi-directional) (Madigan and Lloyd, 2005a). Body segment center-of-masses (CoMs) were calculated using scaling factors (i.e., proportions of segment lengths) provided in Dumas et al. (2007) and used for representing the linear motions of each segment. Three-dimensional rotations of each segment

were defined using the Cardan $Y - x' - z''$ (tilt—obliquity—rotation) convention relative to the laboratory reference frame. Linear and angular velocities of each body segment were calculated using a standard three-point difference equation. Whole body CoM was calculated using a 13-segment model (bilateral foot, shank, thigh, upper arm, and forearm, as well as the pelvis, trunk, and head).

Outcome measures assessed balance recovery ability and responses. Balance recovery ability was defined as the maximum lean angle from which participants could recover their balance with a single step after a loss of balance (Hsiao-Wecksler, 2008). Other balance recovery responses were quantified during single step balance recovery and included spatiotemporal measures, lower limb joint ranges of motion (RoMs) and peak angular velocities, and margin of stability (MoS) measures. Hence, while balance recovery ability was determined for each *EXO* condition, balance recovery responses characterizing the recovery were obtained from each successful balance recovery trial within each *EXO* condition.

Spatiotemporal measures. Temporal events were first extracted during a single step recovery from the in-line loadcell, force platforms, and whole body CoM data (King et al., 2005; Wojcik et al., 1999), as described in Figure 5-2 (a). Specifically, release (perturbation) onset was defined as the instant when the rope tension dropped to zero using the in-line loadcell data. As all participants used the right foot as the dominant (leading) foot for stepping, vertical force from the right force platform was used to detect pushoff and liftoff. Pushoff was defined as the instant when the vertical force exerted by the stepping foot on the right force platform decreased to a local minimum prior to increasing in preparation for weight transfer. Liftoff was defined as the instant when the vertical force exerted by the stepping foot on the right force platform dropped below a threshold of 15 N. Landing was defined as the instant when the vertical force exerted by the stepping foot rose above 15 N on the front force platform. Finally, balance recovery event was defined as the instant when the whole body CoM reached its maximum anterior position.

Temporal measures included reaction time (time interval between release onset and pushoff) and step time (time interval between liftoff and landing). Spatial measures included step length (horizontal distance in

the AP direction of the calcaneus marker on the stepping foot from liftoff and landing) and step angle (Figure 5-2 (b)) (Hsiao-Wecksler and Robinovitch, 2007).

Joint RoM and peak angular velocities. Lower limb joint RoMs and peak angular velocities were obtained from sagittal plane kinematics of the stepping leg during the time interval between release onset and balance recovery. Outcome measures included RoM at the hip, knee, and ankle, and peak velocities in flexion and extension at the hip, knee, and ankle.

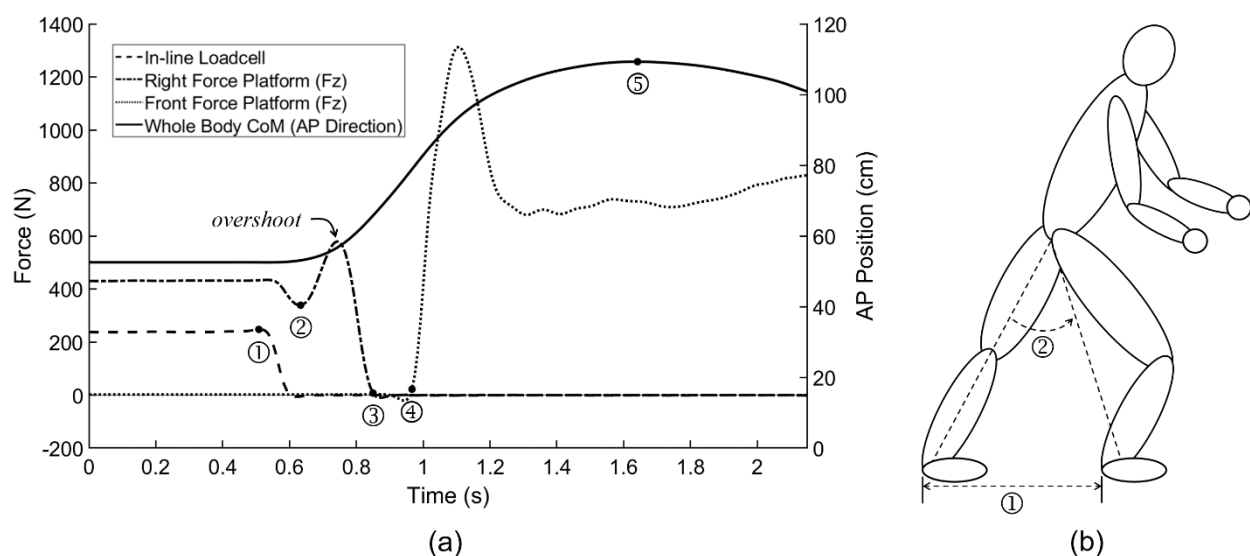


Figure 5-2. (a) Sample time history of forces (in-line loadcell, right force platform [F_z], and front force platform [F_z]) and whole body CoM displacement in anteroposterior (AP) direction during a single step recovery. Numbers indicate ①release onset, ②pushoff, ③liftoff, ④landing, and ⑤balance recovery; (b) Illustration of participant's body configuration at landing. Numbers indicate ①step length and ②step angle.

Margin of Stability (MoS). Extrapolated whole body CoM (XCoM) was obtained using the equation described in Hof et al. (2005). MoS is defined as the minimum distance between XCoM and base-of-support (BoS) borders (Hof et al., 2005). The BoS was determined by reflective markers attached on the 2nd distal phalanx, calcaneus, 1st metatarsal head, and 5th metatarsal head of the feet. MoS measures used in this study included MoS_{LANDING} (i.e., MoS at the instant of landing) and MoS_{RECOVERY} (i.e., MoS at the instant of balance recovery) in anteroposterior (AP) direction. Positive (negative) MoS values indicate that XCoM is

inside (outside) of the BoS boundary, implying that the higher MoS value indicates the more stable (Hof et al., 2005).

5. 2. 5. Statistical analysis

A two-way, repeated measures analysis of variance (ANOVA) was performed on maximum lean angle, with *EXO* as within-subject factor and gender as between-subjects factor. Furthermore, separate three-way, repeated measures ANOVA was performed on each balance recovery response outcome, with *EXO* and lean angle as within-subject factors and gender as between-subjects factor. Lean angle included four levels (Lev1: 20°, Lev2: 22.5°, Lev3: 25°, Lev4: maximum lean angle). While balance recovery responses were measured at multiple lean angles, we started at the minimum lean angle of 20° and successively included outcomes at all angles at which more than 90% of total trials (across participants and *EXO* conditions) were successes. Since all participants completed trials at a maximum lean angle, that was also included in the analysis, although the maximum angle at which recovery was successful varied across participants and *EXO* conditions. Model parametric model assumptions were evaluated. Significant effects were followed by *post hoc* pairwise comparisons using Tukey's HSD tests, and significant interaction effects were further examined using simple-effects analyses. Given the study goals, the subsequent presentation of results and the discussion emphasizes the main and interaction effects of *EXO*. In the case of any significant interaction effects of *EXO* and *GEN*, they were also visually examined to understand whether they were ordinal, and important. In the case of interactions being ordinal, (i.e., no cross-overs or opposing trends as a function of *GEN*), main effects of *EXO* were presented; in case of non-ordinal interactions, results of *EXO* effects were presented separately for each gender. All statistical analyses were completed using JMP® Pro 15.0 (SAS Institute Inc., Cary, NC), using the restricted maximum likelihood (REML) method, with statistical significance concluded when $p < 0.05$.

5. 3. Results

Mean (SD) of maximum lean angle in NoExo, *EXO*_{OFF}, *EXO*_{LOW}, and *EXO*_{HIGH} conditions were 30.6° (3.6°), 31.1° (5.2°), 30.9° (5.0°), and 30.8° (5.8°), respectively. Maximum lean angle was not significantly

different with *EXO* ($p=0.979$) or gender ($p=0.826$), or their interaction ($p=0.146$). However, most balance recovery responses showed significant main effects of *EXO*. Table 5-1 summarizes ANOVA results, regarding *EXO*, lean angle, and gender effects; and these results are presented in more detail for each group of outcomes. Table D-1 and D-2 in the appendix include descriptive summaries of each balance recovery response and summary of *post hoc* comparisons for the significant main effects of *EXO*, respectively.

Table 5-1. Summary (p -values) of ANOVA results for the main and interaction effects (E : *EXO* condition, L : lean angle, G : gender) on each outcome measure for balance recovery responses. Significant differences are highlighted in bold ($p<0.05$).

Outcome measures	Main and interaction effects						
	E	L	G	$E \times L$	$E \times G$	$L \times G$	$E \times L \times G$
<i>Spatiotemporal measures</i>							
Reaction time	<0.001	0.035	0.177	0.761	0.869	0.037	0.746
Step time	0.003	<0.001	0.506	0.221	0.568	0.603	0.726
Step length	<0.001	<0.001	0.333	0.953	0.426	0.655	0.741
Step angle	<0.001	<0.001	0.196	0.961	0.272	0.490	0.657
<i>RoM</i>							
Hip	<0.001	<0.001	0.078	0.984	0.288	0.687	0.886
Knee	<0.001	<0.001	0.700	0.956	0.023	0.795	0.701
Ankle	0.479	0.069	0.384	0.243	0.285	0.225	0.388
<i>Peak angular velocity</i>							
Hip flexion	<0.001	<0.001	0.038	0.828	0.212	0.500	0.761
Hip extension	0.031	0.393	0.291	0.023	0.602	0.748	0.874
Knee flexion	<0.001	<0.001	0.030	0.871	0.029	0.360	0.692
Knee extension	0.028	<0.001	1.000	0.538	0.601	0.146	0.584
Ankle dorsiflexion	0.324	0.331	0.246	0.317	0.319	0.304	0.339
Ankle plantarflexion	0.332	0.337	0.328	0.320	0.352	0.296	0.338
<i>MoS</i>							
MoS _{LANDING}	0.023	<0.001	0.003	0.481	0.171	0.832	0.694
MoS _{RECOVERY}	0.018	<0.001	0.140	0.255	0.258	0.860	0.459

5. 3. 1. Spatiotemporal measures

Significant main effects of *EXO* were observed on all spatiotemporal measures (Table 5-1). *Post hoc* analysis showed that reaction time was higher in all exoskeleton conditions, compared to the no-exoskeleton control condition (Figure 5-3). Step time was lower in *EXO_{OFF}* and *EXO_{LOW}* conditions than in NoExo condition (Figure 5-3). Furthermore, both step length and step angle were lower in all exoskeleton conditions than in NoExo condition (Figure 5-3).

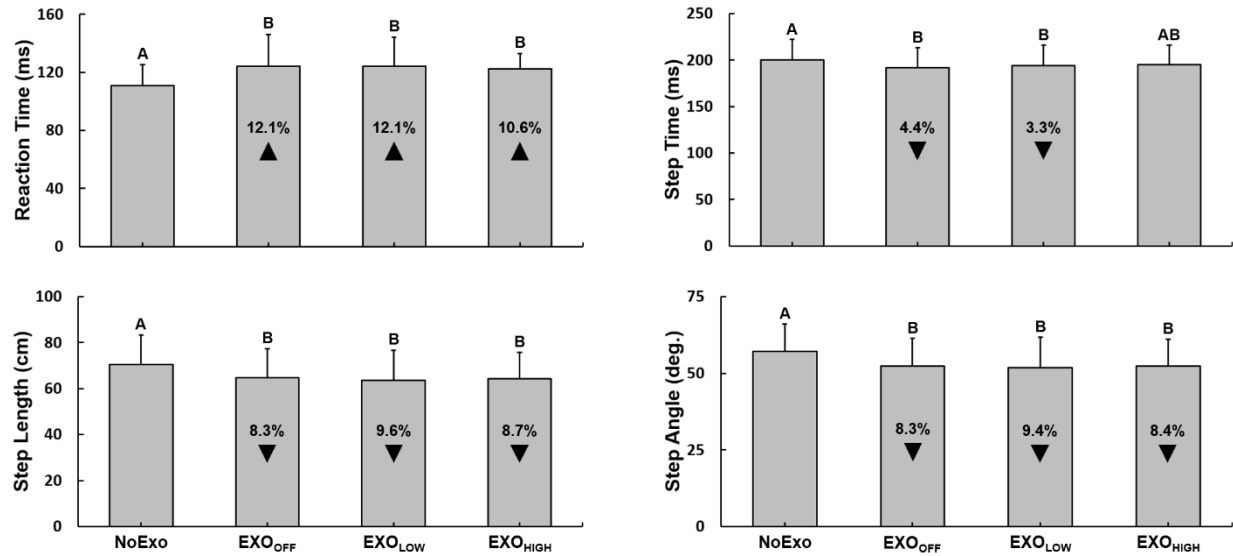


Figure 5-3. Spatiotemporal measures in four *EXO* conditions during a single step recovery following a loss of balance. Data are represented as group means, with error bars signifying standard deviations. Upper case letters specify groupings obtained from pairwise comparisons between *EXO* conditions. The triangle (or inverted triangle) symbols and accompanying numbers indicate a significant increase (or decrease) from the control no-exoskeleton condition, and the corresponding percentage change in dependent measure.

5. 3. 2. Joint RoMs and peak angular velocities

Figure 5-4 presents sample joint angles and angular velocities during a single step recovery. While ankle RoM did not present any significant main or interaction effects of *EXO*, both hip and knee RoMs were lower in all exoskeleton conditions compared to the NoExo condition (Table 5-1 and Figure 5-5).

A significant main effect of *EXO* was also seen in peak velocities in flexion and extension at the hip and knee (Table 5-1). While peak hip flexion velocity was lower in all exoskeleton conditions compared to the NoExo condition, peak hip extension velocity was lower in only EXO_{HIGH}, compared to the NoExo condition (Figure 5-6). Simple effects analysis for the significant interaction effect of *EXO* and lean angle on peak hip extension velocity showed that peak hip extension velocity was lower in EXO_{HIGH} compared to other conditions only in 20° lean angle. Peak knee flexion velocity was lower in EXO_{LOW} and EXO_{HIGH} conditions compared to NoExo, was also lower in EXO_{HIGH} than in EXO_{OFF} (Figure 5-6). Peak knee extension velocity was lower only in EXO_{OFF} condition compared to NoExo (Figure 5-6). Ankle joint velocities did not show any significant main or interaction effects of *EXO* (Table 5-1).

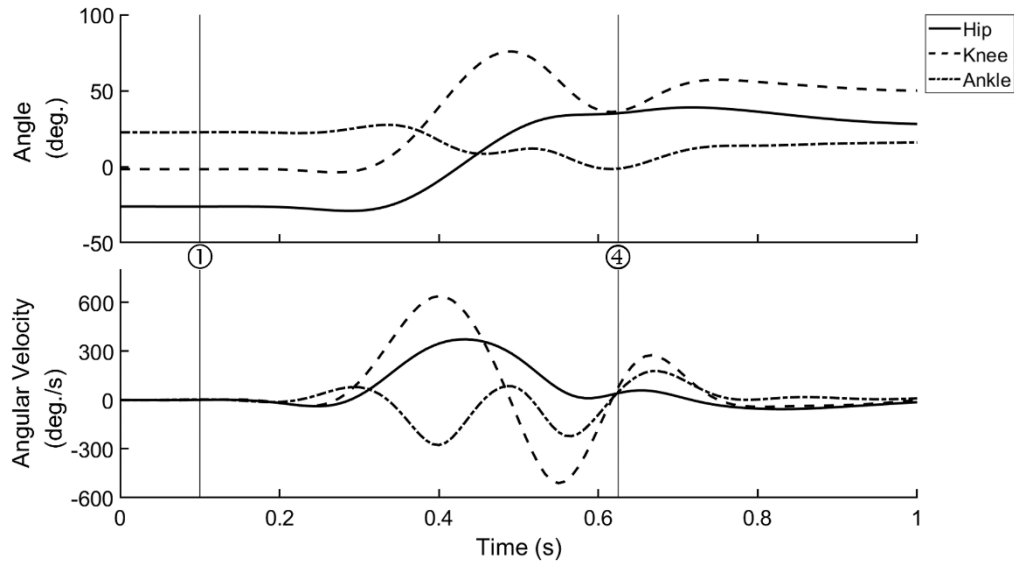


Figure 5-4. Sample joint angles and angular velocities during a single step recovery. Positive values indicate hip flexion, knee flexion, and ankle dorsiflexion. Numbers indicate ①release onset and ④landing.

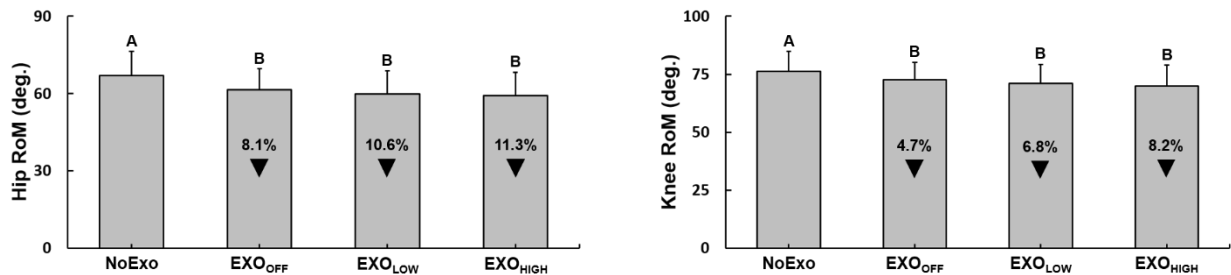


Figure 5-5. Joint RoMs in four *EXO* conditions during a single step recovery following a loss of balance. Data are represented as group means, with error bars signifying standard deviations. Upper case letters specify groupings obtained from pairwise comparisons between *EXO* conditions. The inverted triangle symbols and accompanying numbers indicate a significant decrease from the control no-exoskeleton condition, and the corresponding percentage change in dependent measure.

5. 3. 3. Margin of stability measures

Significant main effects of *EXO* were seen in both $MoS_{LANDING}$ and $MoS_{RECOVERY}$ (Table 5-1). *Post hoc* analysis revealed that $MoS_{LANDING}$ was lower in EXO_{HIGH} compared to NoExo and EXO_{OFF} conditions (Figure 5-7). $MoS_{RECOVERY}$ was lower in the EXO_{HIGH} than in NoExo condition (Figure 5-7).

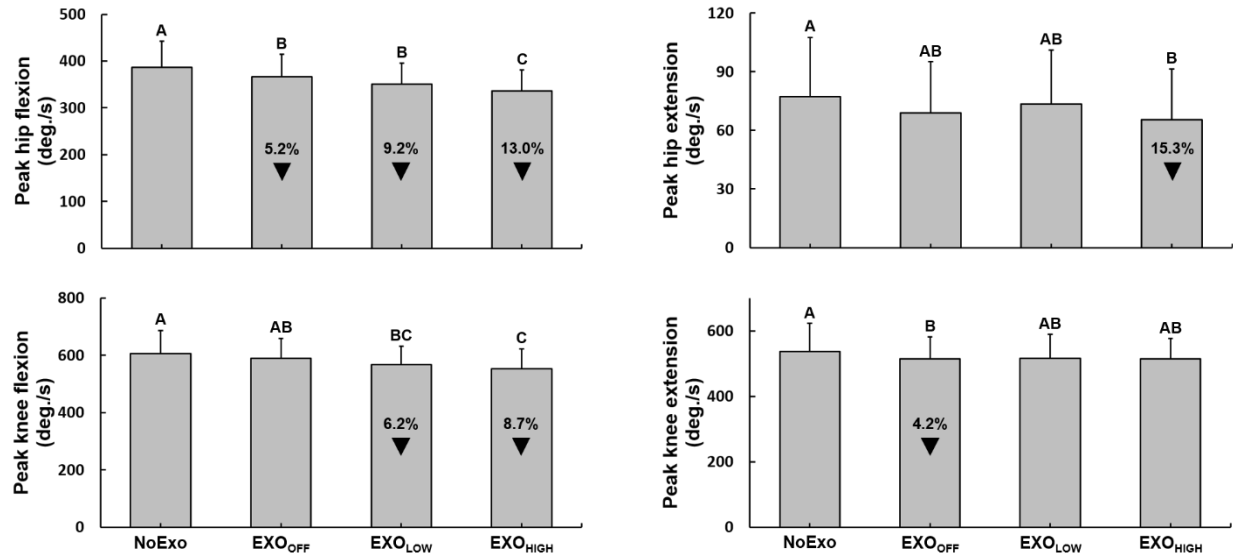


Figure 5-6. Peak joint angular velocities in four *EXO* conditions during a single step recovery following a loss of balance. Data are represented as group means, with error bars signifying standard deviations. Upper case letters specify groupings obtained from pairwise comparisons between *EXO* conditions. The inverted triangle symbols and accompanying numbers indicate a significant decrease from the control no-exoskeleton condition, and the corresponding percentage change in dependent measure.

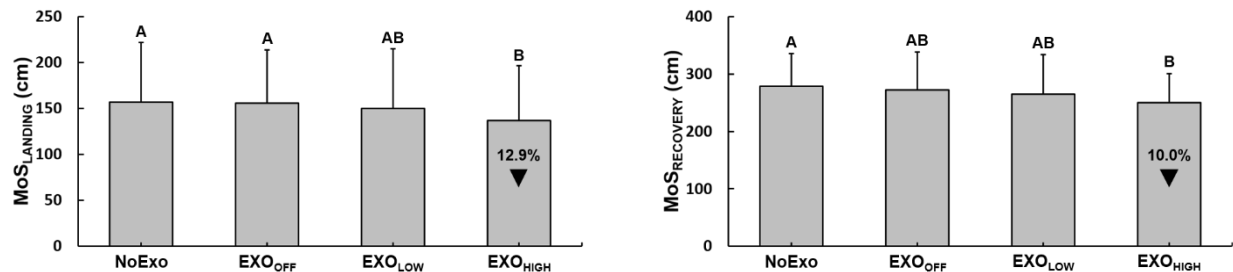


Figure 5-7. MoS measures in four *EXO* conditions during a single step recovery following a loss of balance. Data are represented as group means, with error bars signifying standard deviations. Upper case letters specify groupings obtained from pairwise comparisons between *EXO* conditions. The inverted triangle symbols and accompanying numbers indicate a significant decrease from the control no-exoskeleton condition, and the corresponding percentage change in dependent measure.

5. 4. Discussion

5. 4. 1. Balance recovery ability

Our results on maximum lean angle showed that balance recovery ability did not differ between *EXO* conditions. Experimental order of *EXO* conditions was also not significant ($p=0.112$), suggesting that there were no major learning or fatigue effects on maximum lean angle over time. We further examined the

effects of *EXO* on the magnitude (% increase from the baseline) of overshoot (Figure 5-2 (a)), measured by the right force platform, immediately following pushoff. However, no significant main effect of *EXO* was seen ($p=0.091$), implying that participants did also not exert more effort to recover their balance in exoskeleton conditions. Although maximum lean angle has previously shown significant differences with age and/or gender (Madigan and Lloyd, 2005b; Wojcik et al., 1999), we speculate that as all our participants were young and healthy, it may be not a sensitive enough measure to differentiate the various *EXO* conditions involved in this study. Previous studies have also reported that only measures of balance recovery responses have shown fair-moderate reliability (e.g., Ringhof et al., 2019; Werth et al., 2021), and hence, the reliability of maximum lean angle also needs to be investigated in future work.

5. 4. 2. Balance recovery responses

Wearing the backXTM decreased peak hip flexion velocity by 5-13% (Figure 5-6): changes in EXO_{OFF} condition may be due to constrained movement and/or added mass when wearing the backXTM with no supportive torque, whereas further decrease in EXO_{HIGH} condition may be due to the external hip extension torque provided when wearing the backXTM with high supportive torque. Wearing the backXTM also led to 8-11% decrease in hip RoM (Figure 5-5), by decreasing peak hip flexion angle (6-10%; $p\text{-values}<0.005$). Hence, decrease in both peak hip flexion velocity and hip RoM indicate that wearing the backXTM impeded hip flexion during reactive stepping.

From Figure 5-4, it can be observed that, between rope release and landing (i.e., stepping), both the hip and the knee flexed. Therefore, changes in hip kinematics may have caused synchronized changes in the knee, through a decrease in peak knee flexion velocity and knee RoM (Figure 5-5 and 5-6). The decreased peak hip flexion velocity may also be the primary reason that resulted in decreased step length (8-10%), step time (3-4%), and step angle (8-9%) when wearing the backXTM (Figure 5-3). Note that spatiotemporal measures did not present differences between the three exoskeleton conditions, suggesting that while wearing the BSE affected them, the level of external hip extension torque had minimal effects on these spatiotemporal measures.

Previous studies on age-related changes have shown that stepping responses are associated with balance recovery performance. For example, Carty et al. (2012) noted that weakness in the hip flexors negatively influenced balance recovery. Madigan and Lloyd (2005b) also reported that older adults showed slower peak hip (26% decrease) and knee (30% decrease) flexion velocities. Furthermore, large RoM at hip and knee was found to be the critical factor for successful balance recovery (Wojcik et al., 1999, 2001). Hence, our results from spatiotemporal measures, joint RoMs, and peak joint velocities indicate that wearing the backXTM impaired balance recovery.

Reaction time increased by 10-12% in all exoskeleton conditions (Figure 5-3). Although restricted body movement, added mass, and external hip extension torque may not affect reaction time, there are some explanations to consider. First, wearing the backXTM may have interfered with joint proprioception (Stirling et al., 2020). Second, given that postural anxiety can lead to reduced spinal reflex excitability (Sibley et al., 2007), increased postural anxiety when wearing the backXTM may also have resulted in increased reaction time. Third, Do et al. (1999) noted that, in reactive stepping following a loss of balance, the control of short swing phase duration can reduce reaction time. Hence, it can be speculated that participants may have programmed a longer swing phase duration prior to release, to increase BoS when wearing the backXTM, and this in turn increased reaction time. Whatever the reason may be, successful recovery from unexpected balance loss in real life, and/or from slips and trips that occur during walking, may not allow individuals the affordance to react more slowly.

Both MoS_{LANDING} and MoS_{RECOVERY} decreased (10-13%) in the EXO_{HIGH} condition (Figure 5-7), indicating that wearing the backXTM with high supportive torque impaired dynamic stability during balance recovery following a loss of balance (Hof et al., 2005; Ringhof et al., 2019). From Figure 5-8, it can be observed that XCoM is located inside of the BoS immediately after landing, explaining why MoS values are always positive at both landing and balance recovery in this study. Figure 5-8 also shows that XCoM gradually moves towards the anterior direction after rope release, and moves back to the posterior after landing (i.e., stepping). Hence, it appears that both decreased step length and increased step time are associated with

decrements in MoS, as the former one can lead to a decrease in BoS and the latter diverts the direction of XCoM movement more slowly (Hsiao-Weckslar and Robinovitch, 2007). These associations further explain why MoS values decreased in EXO_{HIGH} condition only, given that, compared to the control condition, step length decreased in all exoskeleton conditions and step time decreased in EXO_{OFF} and EXO_{LOW} conditions only (Figure 5-3).

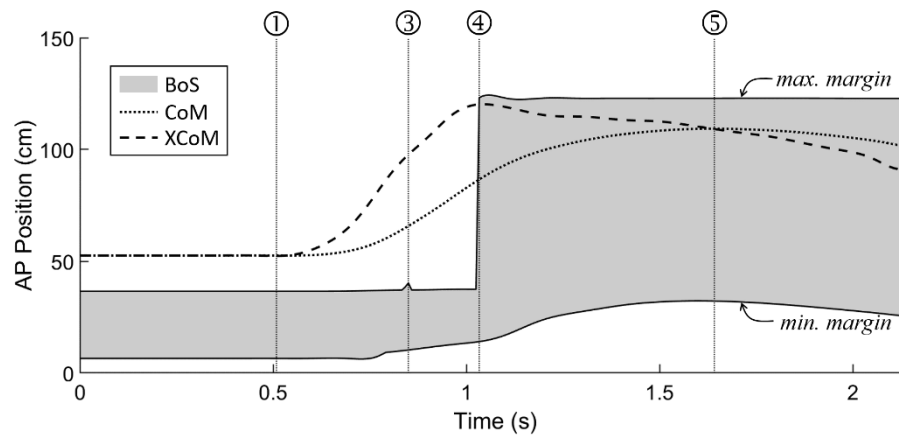


Figure 5-8. An example showing how base of support (BoS), whole body center-of-mass (CoM), and extrapolated whole body CoM (XCoM) change in anteroposterior (AP) direction. Numbers indicate ①release onset, ③liftoff, ④landing, and ⑤balance recovery

5. 4. 3. Limitations

This study has some limitations common with most laboratory-based research conducted using exoskeletons such as individual differences in BSE fit/adjustments that may have influenced our findings, focus only on the short-term effects of BSE use, and inclusion of only rather healthy and young individuals. Further, although we minimized any anticipatory movements by confirming vertical force exerted by each foot to be equal before releasing the participants, unlike falls outside of laboratory, participants were expecting a fall during the experimental protocol. Lastly, lower limb muscle activities were not measured in this study. Hence, while our results from balance recovery responses delivered valuable findings, changes in muscle coordination strategies or co-contractions could not be investigated.

5. 4. 4. Conclusions

This study comparatively evaluated the effects of wearing a BSE with various levels of hip-extension torque settings on balance recovery following a loss of balance. Overall, the maximum lean angle from which individuals could successfully recover, following a simulated forward fall, was not altered when wearing the BSE. However, results from recovery responses showed that wearing the BSE negatively affected balance recovery at all lean angles. Specifically, wearing the backXTM increased reaction times, and impeded hip flexion (i.e., decrease in both peak hip flexion velocity and hip RoM) during reactive stepping, and this in turn decreased peak knee flexion velocity, knee RoM, and step length, all suggesting impaired balance recovery. While constrained movement and/or added mass of the exoskeleton may be the primary factor that decreases balance recovery performance, results from the margin of stability measures suggest that external hip extension torque of the BSE can further impair balance recovery. Future work extending kinetic analyses to recovery responses, as well as a study of recovery when responding to slips and trips while walking, would offer a more complete picture of how a BSE may impact dynamic balance recovery following a loss of balance.

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Appendix D

Table D-1. Mean (SD) of each outcome measure for balance recovery responses in 20° (i.e., minimum lean angle) and maximum lean angle.

Outcome measures	Lean angle = 20°				Lean angle = maximum			
	NoExo	EXO _{OFF}	EXO _{LOW}	EXO _{HIGH}	NoExo	EXO _{OFF}	EXO _{LOW}	EXO _{HIGH}
<i>Spatiotemporal measures</i>								
Reaction time	104.69	126.56	125.52	117.19	118.23	125.52	127.08	127.08
(ms)	(13.25)	(40.85)	(36.45)	(10.74)	(16.73)	(9.36)	(9.86)	(10.76)
Step time	203.13	190.63	184.90	188.02	208.33	200.00	205.73	205.21
(ms)	(27.53)	(22.54)	(24.00)	(23.37)	(18.76)	(19.72)	(15.43)	(16.63)
Step length	62.03	55.42	52.97	54.10	82.88	77.54	77.00	75.23
(cm)	(9.47)	(9.10)	(10.86)	(10.07)	(9.76)	(9.43)	(7.07)	(4.16)
Step angle	50.63	45.64	43.49	44.97	67.19	62.17	61.62	60.91
(deg.)	(5.62)	(6.08)	(8.79)	(7.78)	(5.42)	(6.58)	(5.33)	(4.95)
<i>RoM</i>								
Hip	60.12	55.79	52.56	52.02	75.87	69.76	68.73	67.21
(deg.)	(7.16)	(5.78)	(7.34)	(9.02)	(7.36)	(5.38)	(5.43)	(5.41)
Knee	69.39	68.38	65.22	64.24	83.56	79.38	77.87	76.56
(deg.)	(5.42)	(5.32)	(8.40)	(10.38)	(6.97)	(6.53)	(5.89)	(7.76)
Ankle	28.46	25.91	26.75	25.8	36.32	35.37	34.96	35.25
(deg.)	(5.27)	(4.89)	(5.58)	(5.89)	(3.45)	(3.57)	(4.38)	(4.72)
<i>Peak angular velocity</i>								
Hip flexion	346.30	333.31	328.69	303.86	413.92	393.51	373.56	356.16
(deg./s)	(59.16)	(48.58)	(53.17)	(52.92)	(50.41)	(35.17)	(38.89)	(38.56)
Hip extension	78.09	71.27	77.99	48.88	80.15	72.54	73.02	76.49
(deg./s)	(34.23)	(26.08)	(28.48)	(14.43)	(23.87)	(32.41)	(23.20)	(27.19)
Knee flexion	545.62	541.97	530.57	508.86	638.87	628.43	598.10	584.36
(deg./s)	(68.38)	(72.91)	(72.26)	(78.17)	(78.31)	(63.46)	(59.97)	(69.60)
Knee extension	487.16	466.28	470.94	464.90	581.27	564.48	551.64	538.27
(deg./s)	(75.26)	(68.21)	(76.89)	(62.05)	(89.56)	(57.44)	(66.98)	(46.08)
Ankle dorsiflexion	161.89	149.20	153.16	149.41	159.00	174.98	177.37	182.55
(deg./s)	(36.66)	(33.04)	(35.74)	(37.61)	(31.00)	(36.99)	(42.41)	(38.31)
Ankle plantarflexion	269.61	262.48	272.47	265.14	309.14	321.50	332.00	335.02
(deg./s)	(50.31)	(53.54)	(66.24)	(66.22)	(47.16)	(44.67)	(43.41)	(40.95)
<i>MoS</i>								
MoS _{LANDING}	222.93	203.40	194.48	176.20	95.35	111.15	91.97	71.59
(cm)	(41.00)	(49.33)	(67.44)	(37.52)	(47.20)	(42.83)	(55.98)	(59.84)
MoS _{RECOVERY}	310.09	293.97	283.69	256.12	254.13	257.73	236.08	212.50
(cm)	(53.01)	(53.30)	(73.08)	(51.25)	(50.98)	(56.96)	(79.40)	(44.04)

Table D-2. Summary of *post hoc* pairwise comparisons (Tukey's HSD tests) for the significant main effects of exoskeleton on each outcome measure for balance recovery responses. Table entries are *p* values, and significant differences are highlighted in bold ($p < 0.05$).

Outcome measures	EXO condition					
	NoExo vs. EXO _{OFF}	NoExo vs. EXO _{LOW}	NoExo vs. EXO _{HIGH}	EXO _{OFF} vs. EXO _{LOW}	EXO _{OFF} vs. EXO _{HIGH}	EXO _{LOW} vs. EXO _{HIGH}
<i>Spatiotemporal measures</i>						
Reaction time	< 0.001	< 0.001	< 0.001	0.999	0.970	0.938
Step time	0.002	0.028	0.070	0.853	0.639	0.982
Step length	< 0.001	< 0.001	< 0.001	0.916	0.983	0.992
Step angle	< 0.001	< 0.001	< 0.001	0.956	0.999	0.981
<i>RoM</i>						
Hip	< 0.001	< 0.001	< 0.001	0.506	0.234	0.962
Knee	0.003	< 0.001	< 0.001	0.559	0.107	0.775
<i>Peak angular velocity</i>						
Hip flexion	0.002	< 0.001	< 0.001	0.068	< 0.001	0.036
Hip extension	0.183	0.833	0.031	0.648	0.884	0.228
Knee flexion	0.131	< 0.001	< 0.001	0.094	< 0.001	0.225
Knee extension	0.046	0.097	0.065	0.991	0.999	0.999
<i>MoS</i>						
MoS _{LANDING}	1.000	0.849	0.035	0.872	0.041	0.233
MoS _{RECOVERY}	0.914	0.560	0.014	0.915	0.085	0.315

Chapter 6. Conclusion

The primary goal of this work was to investigate the effects of wearing a back-support exoskeleton (BSE) on balance and postural stability, in a variety of situations that can be encountered in the workplace when wearing a BSE. A comprehensive and systematic evaluation of passive BSE(s) was performed under controlled laboratory conditions, focusing on three classes of human activity: 1) the maintenance of a specified posture, 2) voluntary movement, and 3) the reaction to an external perturbation. Overall, our results suggest that wearing a BSE can have complex effects on the postural control system, and can compromise balance and/or postural stability in static standing, voluntary walking, and reactive stepping following a forward loss of balance. Results from this work are expected to further support our long-term goal of developing safety and ergonomics guidelines to minimize BSE-related fall risk in the workplace. Furthermore, delineation of exoskeleton effects into effects of changes in human+exoskeleton center of mass, those caused by restrictions in joint motion due to the rigidity of the exoskeleton, and effects of external device-generated torque; and quantification of a variety of postural responses as a function of exoskeleton support torque levels, and are also expected to guide future exoskeleton design.

6. 1. Summary of major results

In Chapter 2, we report new evidence regarding whether BSE use affects standing balance and functional limits of stability. Specifically, results from this study demonstrated that BSE use increased center-of-pressure (COP) median frequency and mean velocity during bipedal stance. In unipedal stance, wearing a BSE with high supportive torque led to smaller COP displacement and sway area, and a longer time to contact the stability boundary, especially among males, indicating slightly improved postural balance. However, BSE use had minimal effects on functional limits of stability. While both alterations in center-of-mass (COM) and supportive torque affected bipedal standing, restricted motion effects were relatively minimal, and the effects of supportive torque were predominant in unipedal standing.

Chapter 3 presents the first examination of the effects of wearing a BSE with multiple supportive torque levels on gait performance and stability during level walking. Results from this study showed that wearing a BSE with high supportive torque was associated with decrease in required coefficient-of-friction (7%) and maximum Lyapunov exponent of trunk kinematics (3%), indicating decreased slip risk and slightly improved trunk stability, respectively. However, BSE use with high supportive torque also led to increased step width (10%) and gait variability (8-19%), and decreased margin of stability (26%) in the mediolateral direction during the swing phase. Overall, changes in step width and increased gait variability were the most substantial effects found from the BSE. Given that increased gait variability is associated with increased fall risk among older adults and increased step width is associated with higher metabolic cost, our results may suggest that using such an exoskeleton with high external torque could adversely affect fall risk and walking energetics.

In Chapter 4, we quantified the effects of using a BSE on lower limb joint kinematics and kinetics during level walking. The torque profile of the BSE was measured using a dynamometer, and used for assessing hip torque generated by the participant only. Overall, wearing a BSE impeded hip flexion by decreasing hip range of motion and hip flexion velocity. When hip extension torques were required for walking (i.e., the initial 0-10% and final 75-100% of the gait cycle), the BSE generated substantial external hip extension torque, and participants showed adapted gait patterns (i.e., decrease in hip extension torque and extensor power), and this in turn may have led to decrease in knee extension velocity, flexion torque, and extensor power. However, participants increased hip extensor power and ankle dorsiflexor power, while the BSE did not provide high external torque, to compensate for the added mass and inertial effects of the device during the stance phase. When hip flexion torques were required for walking (i.e., the middle 15-70% of the gait cycle), although the BSE with high supportive torque did not generate substantial hip extension torque, participants showed changes in lower limb kinematics (i.e., increase in ankle plantarflexion and dorsiflexion velocities) and kinetics (i.e., increase in hip flexor, knee extensor, and ankle plantarflexor powers) to increase mechanical energy generation for propelling the leg into the swing phase. Hence, results from this

study further support the findings from Chapter 3, that wearing a BSE may increase metabolic costs and decrease postural stability during level walking.

Finally, Chapter 5 describes the first ever study to assess the effects of wearing a BSE on single step balance recovery from a forward loss of balance. Overall, the maximum lean angle from which individuals could successfully recover, following a simulated forward fall, was not significantly altered when wearing a BSE under any torque setting. However, wearing the BSE under all torque conditions increased reaction times, and impeded hip flexion (i.e., decrease in both peak hip flexion velocity and hip range of motion), resulting in decreased peak knee flexion velocity, knee range of motion, and step length. Margin of stability measures decreased significantly in the high-torque condition. Results from this study suggest that constrained movement and/or added mass of the exoskeleton may be the primary factors that decrease balance recovery performance, and that external hip extension torque of the BSE can further impair balance recovery.

6. 2. Overall limitations

There are some potential limitations common for the four studies presented here. First, participants were all relatively young. Hence, the generalizability of our findings to diverse populations is unclear. Second, the BSE was fitted and adjusted following manufacturers' instructions. Hence, individual differences in BSE fit/adjustments and relative motions between the BSE and wearer may have caused some misalignment between the exoskeleton and human hip-joint axes, thereby affecting the effectiveness of torque transfers from the device to the humans. Third, muscle activities were not measured in this work. Hence, changes in muscle coordination strategies or co-contractions that may have been associated with our findings could not be investigated. Lastly, although learning/familiarization trials were built into all protocols, we did not investigate the long-term effects of BSE use on postural balance. As such, our results may not be generalizable to long-term users whose postural control could evolve with repeated and regular use of a BSE.

6. 3. Overall applications

Our work shows that exoskeleton use can deteriorate balance and/or postural stability in situations of static standing, voluntary walking, and reacting to an external perturbation. Hence, there are several points that should be considered for practical industrial adoption of BSEs.

Results from bipedal stance indicated that wearing a BSE can decrease standing balance, potentially due to a vertically elevated COM of the human+exoskeleton system. Among the two devices tested, a more substantial change in postural balance was observed when wearing the Laevo™ exoskeleton, as compared to the SuitX™ device. Since both devices can produce similar levels of torque, this difference may have stemmed from the differences in device anchor points and torque transfer mechanisms between the two BSEs investigated. However, while a number of dependent measures were included in this study, the most substantial change (50% difference from no-exoskeleton) condition was observed only in the median frequency of the center of pressure in the anterior-posterior direction. Hence, the practical implications of BSE use for static standing tasks are not currently clear, based on these findings. Future investigations of tasks involving prolonged standing in a variety of postures characteristic of manual material handling are recommended. It is also of note that the heaviest BSE investigated in this study weighted only 4.5 kg – if the device were heavier, standing balance may be more severely impacted. Although our results from unipedal stance present that wearing a BSE can improve postural balance, this may be not relevant in practice, as unipedal standing may not be common in occupational activities.

Results from level walking suggest that BSE use, especially with high supportive torque, decreases walking stability, and requires more mechanical energy to move forward, thereby increasing metabolic costs. Hence, workers need to consider turning off device assistance prior to walking. This is in line with the manufacturer's recommendation for the specific BSE investigated in our later studies, the backX™, that users should turn off device assistance prior to walking. Despite that, the effects of high-torque condition were quantified in this study, to understand the worst-case safety challenges associated with not turning off device assistance (as it may not be practical in several cases, for a worker using the exoskeleton to lift a

heavy load with both hands, to turn off the device, prior to subsequently carrying that load to transfer it to a different location). Our investigation of high-torque assistance is also useful from a design standpoint, for future devices, and also other device models where there are no explicit instructions to turn off the device prior to locomotion. Finally, unlike simulated level walking in this work, walking in the workplace may also include turning and stair climbing as well as carrying tasks. Given that such tasks are more challenging in terms of postural control, and that some of them require greater degrees of hip flexion (in which the exoskeleton's extension torque may be more strongly opposing user motion), BSE use may have larger effects on balance and postural stability, compared to level walking performed in a controlled laboratory environment. Although we only considered short-term effects in this study and they did not show any modifications to distributions of joint torques, these effects and any alterations in motor strategies would also need to be carefully considered, in the formulation of future guidelines.

Results from reactive stepping show that BSE use may impair balance recovery, regardless of external torque level, and that the high supportive torque condition was associated with further decrease in dynamic stability. Since there is potential for any of the diverse activities that workers typically engage in during manual material handling (e.g., forward bending, lifting, short-distance walking at high speeds) to cause loss of balance, several subsets of such activities may pose practical postural challenges for successful execution of balance-recovery when wearing a BSE. Also, any risk factors that can cause postural perturbations (e.g., slippery surfaces, obstacles on a walkway) may further exacerbate such risks. These risks need to be further quantified in future studies.

As mentioned above, findings from this work are limited to short-term effects for young and healthy individuals. Hence, future work considering segments of the population with impaired strength, increased reaction time, and mobility or balance decrements, such as older workers, is strongly recommended, before widespread adoption of BSEs in safety-critical environments in industry.

Also, from a future design standpoint, BSE devices that automatically recognize/classify human activity and adapt in those cases when the external hip extension torque is undesirable (e.g., by being sensitive to gait phase), have the potential to compensate for the adverse effects of hip-extension torque on postural control and stability.

In summary, given the growing interest in exoskeletons as a workplace intervention to control and prevent work-related musculoskeletal disorders, this work highlights potential balance and postural stability issues associated with BSE use in a variety of situations. We consider that the potential benefits of this emerging technology can be maximized by ergonomic guidelines/practices and improved BSE design to minimize fall risk in the workplace.