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Full title:

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

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Abstract

Background: 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.

Research question: Does using a BSE adversely affect gait performance and stability, and are such effects dependent on specific BSE external torque characteristics?

Methods: Twenty participants (10M, 10F) completed five level over-ground walking trials and a five-minute treadmill walking trial while wearing a BSE (backX^{TM}) 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 (MLE) of trunk kinematics and the margin-of-stability (MoS).

Results: Using the backX^{TM} with high supportive torque decreased slip risk (7% decrease in RCoF) and slightly improved trunk stability (3% decrease in MLE). 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.

Significance: 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.

Key words: occupational exoskeleton; walking stability; workplace fall.

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., [1, 2]). 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 [3] 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) [2, 3]. Hip flexion torque generated during the swing phase of gait is a major contributor to the propulsive power needed for walking [4]. Specifically, ~12 Nm of hip flexion torque is required during the swing phase [5]. However, previous investigations have shown that passive BSEs can produce ~35 Nm of extension torque at each hip [6, 7]. A stable gait pattern requires appropriate magnitudes of lower limb joint torques generated at specific times for each gait event during the stance and swing phases [5]. Hence, external hip-extension torque generated by a BSE could compromise gait performance and stability, and thereby increase the risk of falling.

To assess gait performance, spatiotemporal gait parameters (e.g., step length, stance phase duration) were traditionally utilized. However, stride-to-stride variability in gait kinematic patterns has been increasingly utilized as an alternative measure of locomotion ability, as such variable is thought to reflect the ability of the central neuromuscular control system to regulate gait and maintain a steady walking pattern. In fact, gait variability is associated with instability and fall risk [8, 9]. Additionally, the required coefficient-of-friction (i.e., RCoF, the friction needed during walking) and the minimum foot clearance (MFC) are indicative of slip- and trip-induced fall risks, respectively. Previous investigations have shown that both RCoF and MFC are influenced by walking speed, joint kinematics, and/or leg joint torques, especially about the knee and the hip [10, 11].

To assess gait stability, the maximum Lyapunov exponent (MLE) and margin of stability (MoS) have been quantified. MLE indicates the ability of the motor system to attenuate small perturbations [12], and is a measure that is sensitive to changes in gait stability resulting from aging, neuro-atypical conditions, and various experimental conditions (e.g., [13, 14]). MoS is defined as the distance between the extrapolated center-of-mass (CoM) and the boundaries of the base-of-support [15]. MoS is also sensitive to differences in walking speed, walking environment, age, and pathological conditions (e.g., [16, 17]).

Exoskeletons for gait rehabilitation have been extensively examined in the literature (e.g., [18]). 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). Using an arm-support exoskeleton was found to increase the velocity of the center-of-pressure during quiet standing [19], possibly due to elevation of the CoM of the body + exoskeleton system. Although not directly comparable, a recent study of a powered whole-body exoskeleton [20] reported that wearing the exoskeleton was associated with slower gait and greater role of the hip joint in regulating gait, as evidenced by greater hip range-of-motion and contributions of hip joint motion to gait principal components. Using a passive BSE led to a slower preferred gait speed and shorter stride length [2]. We recently investigated the effects of two different passive BSEs on postural balance during quiet upright stance and found a decrease in postural balance [21].

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. 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 MLE and decrease in MoS). We also hypothesized that such changes will be more pronounced as the external BSE-generated torque increases.

2. Methods

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. The sample size was determined based on a *priori* power analysis using postural stability measures reported in previous investigations [22, 23], for a significance level $\alpha = 0.05$ and power $(1 - \beta) = 0.8$. 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.

2. 2. Exoskeleton selection

The backX™ AC (US Bionics Inc., Berkeley, CA), designed to reduce physical demands on the back during forward bending [24], 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 1). The external torque generator includes gas springs that store energy during forward bending (i.e., trunk/hip flexion), and this energy is released and thereby provides trunk/hip extension torque and assistance, for example during lifting tasks [24]. The four exoskeleton (*EXO*) conditions included in this study were: NoExo (no exoskeleton, control condition), EXO_{OFF} (backX™ with no supportive torque), EXO_{LOW} (backX™ with low supportive torque) and EXO_{HIGH} (backX™ with high supportive torque).

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 (e.g., RCoF), whereas the treadmill protocol allowed us to collect enough strides to compute other dependent measures (e.g., MLE).

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 [25]. 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 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 a procedure described earlier [26]. Briefly, participants were asked to stand at a starting position that was selected to allow each of their feet to land naturally on each force platform, and to walk across the track at the predetermined PWS. Although we confirmed foot placement, participants were not aware of where the force plates were located, so as to not disturb their natural walking. Participants completed five acceptable walking trials (i.e., foot placement on the force platform and at PWS) in each *EXO* condition. Mean (SD) of gait speed in NoExo, EXO_{OFF}, EXO_{LOW}, and EXO_{HIGH} conditions were 1.39 (0.15), 1.39 (0.16), 1.40 (0.15), and 1.40 (0.15), respectively. Gait speed did not significantly differ between *EXO* conditions (p=0.211) during over-ground walking. In the treadmill session, participants walked for five minutes on the treadmill at their PWS in each *EXO* condition (Figure 1 (b)). This walking duration was used to provide sufficient strides for subsequent analyses of gait kinematic variability and MLE [27, 28].

[Figure 1 here]

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 A1) 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 MLE. Gait events in each walking trial (i.e., heel-strike and toe-off of each foot) were detected using a coordinate-based algorithm [29]. Body segment CoMs were calculated using body segment scaling factors [30]. Three-dimensional rotations of each segment were obtained based on segmental coordination systems [30] and were expressed following 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 $r = 0.2$ [31]. Maximum RCoF was computed according to the method described in [11]. 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 MLE, trunk linear and angular velocities [28] were first calculated using 3-point differences and then scaled to body size [32]. 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 [14, 33]. 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 MLE [12, 33]. 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 [15]. MoS is defined as the minimum distance between XCoM and base-of-support borders [15]. 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 2). While a positive MoS value indicated that the XCoM was inside of the base-of-support boundary, negative MoS values indicated that the XCoM was outside of the boundary.

[Figure 2 here]

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. 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 and were qualitatively interpreted as 0.01 = small, 0.06 = medium, 0.14 = large [34].

3. Results

Descriptive summaries of the outcomes are included in the Appendix (Table A1). Table 1 summarizes ANOVA results, regarding *EXO* and *GEN* effects; these results are present in more detail for each group of outcomes. Table A2 presents summary of *post hoc* comparisons for the significant main effects of *EXO* on each outcome measure.

[Table 1 here]

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

EXO had significant main and/or interactive effects on all spatiotemporal gait measures (Table 1 and Figure 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 time 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*×*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).

RCoF and MFC differed significantly across *EXO* conditions (Table 1 and Figure 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 1). RCoF was 15% higher in males than in females.

[Figure 3 here]

3. 2. Gait variability

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

[Figure 4 here]

3. 3. Maximum Lyapunov exponent (MLE)

MLE of trunk kinematics during level walking differed significantly between *EXO* conditions (Table 1 and Figure 5). Specifically, MLE 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}.

3. 4. Margin of stability (MoS)

All MoS measures showed a significant main effect of *EXO* (Table 1 and Figure 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 *EXO* main effect and the *EXO*×*GEN* interaction effect were significant for MoS_{ML} at heel strike (Table 1 and Figure 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 1), and these measures were 27% and 22% higher in males than in females, respectively.

[Figure 5 here]

4. Discussion

4. 1. Gait performance

Walking with the backX™ 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). 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. Although changes such as reduced step length and increased swing time are generally detrimental, and indicate that individuals may have a harder time swinging their legs forward [35, 36], it is unclear whether these changes, which were rather small in our study (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). 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 [37, 38]. In fact, hip range-of-motion in the coronal plane significantly increased (8-9%) when wearing the backX™, compared to the control condition. While increased step width may lead to increased energetic costs in longer-term walking [2, 39], whether this strategy may actually be optimal when using a BSE needs to be investigated in future studies.

The EXO_{HIGH} condition caused a 6-7% decrease in RCoF (i.e., decreased slip risk) compared to the other conditions (Figure 3). When the maximum RCoF occurred (i.e., the instant between heel-strike and midstance [11]), the EXO_{HIGH} condition led to significant changes in foot angle (i.e., the angle between the foot segment and the ground in the sagittal plane; 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 backX™ substantially increased gait variability (7-19%), and the change in gait variability increased with the magnitude of external device torque (Figure 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 [40].

It may also be the case that participants were simply not adequately adapted to the exoskeleton, or additional familiarization/practice is needed. Similar magnitudes in stride-to-stride gait variability have been associated with an increased risk of falling among older adults [8, 9]. Our findings among young adults may not be directly comparable to results from older adults investigated in previous studies: while older adults may show changes in gait variability, they also show other changes such as reductions in force generation and sensory capacity that may affect their fall risk [41]. As younger individuals do not have such age-related impairments in motor or sensory capacity affecting postural control, it is unclear whether similar magnitudes of changes in gait variability will increase fall risk among younger groups as well.

4. 2. Gait stability

The EXO_{HIGH} condition significantly lowered the MLE values of the trunk compared to both the control and no-torque conditions (Figure 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 [42], 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 [28] 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 decreased MoS_{AP} compared to the control condition (Figure 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 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 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.

One important thing to note is that individual PWS was determined in the no-exoskeleton condition, and this speed was then maintained in all *EXO* conditions. However, individuals may naturally choose to walk slower when wearing a BSE, to help reduce gait variability and/or increase dynamic stability. In fact, results from our preliminary study, which investigated the effects wearing BSEs on PWS, showed that PWS decreased slightly in *EXO* conditions (mean difference between *EXO* conditions = 0.087 m/s, which is roughly a ~7% change in PWS). Hence, some of the changes found here may be explained by differences in natural PWS, and furthermore, allowing individuals to walk at their PWS could potentially mitigate some of the changes in postural stability measures assessed here.

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. However, for all outcome measures that showed both statistical significance and substantial differences with *EXO* use, most individuals (>85%) exhibited a consistent direction of the overall effects. 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.

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 may adversely affect walking energetics and fall risk.

Conflict of Interest Statement

The authors have no conflicts of interest to report.

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Tables

Table 1. Summary of ANOVA results for the main and interaction effects of *EXO*, *GEN*, and gait speed 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			
	EXO	*GEN*	*EXO × GEN*	Gait Speed
Spatiotemporal parameters				
Step length	**< 0.001 (0.063)**	0.115 (0.312)	**0.032 (0.024)**	**< 0.001 (0.496)**
Step width	**< 0.001 (0.132)**	0.842 (0.009)	0.218 (0.012)	0.137 (0.004)
Gait cycle time	**< 0.001 (0.105)**	0.058 (0.544)	0.087 (0.018)	**< 0.001 (0.549)**
Swing time	**< 0.001 (0.077)**	0.505 (0.040)	0.143 (0.014)	**< 0.001 (0.246)**
Double support time	**< 0.001 (0.116)**	0.756 (0.014)	**0.015 (0.027)**	**< 0.001 (0.323)**
Slip risk				
RCoF	**< 0.001 (0.049)**	**0.005 (0.149)**	0.957 (0.001)	0.593 (0.000)
Trip risk				
MFC	**< 0.001 (0.052)**	0.847 (0.005)	**0.013 (0.029)**	0.323 (0.002)
SD				
Step length	**< 0.001 (0.298)**	0.169 (0.000)	0.386 (0.054)	**0.004 (0.000)**
Step width	**< 0.001 (0.395)**	0.075 (0.000)	0.062 (0.126)	0.571 (0.000)
Gait cycle time	**0.012 (0.183)**	0.054 (0.000)	0.506 (0.042)	**< 0.001 (0.000)**
Swing time	**< 0.001 (0.422)**	0.756 (0.000)	0.634 (0.031)	**0.001 (0.000)**
Double support time	**< 0.001 (0.413)**	0.739 (0.000)	0.903 (0.010)	**0.006 (0.000)**
SaEn				
Step length	0.155 (0.092)	0.642 (0.000)	0.418 (0.051)	0.793 (0.000)
Step width	0.298 (0.065)	0.780 (0.000)	0.637 (0.031)	0.474 (0.000)
Gait cycle time	**0.005 (0.210)**	0.066 (0.000)	0.140 (0.096)	**< 0.001 (0.000)**
Swing time	**0.010 (0.189)**	0.062 (0.000)	0.614 (0.033)	**< 0.001 (0.000)**
Double support time	0.586 (0.035)	0.763 (0.000)	0.466 (0.046)	0.563 (0.000)
MLE				
Trunk kinematics	**< 0.001 (0.296)**	0.127 (0.000)	0.160 (0.087)	0.054 (0.000)
MoS				
MoS_{AP} during swing	**0.011 (0.030)**	0.398 (0.151)	0.142 (0.014)	**< 0.001 (0.850)**
MoS_{ML} during swing	**< 0.001 (0.052)**	0.096 (0.109)	0.539 (0.006)	0.704 (0.000)
MoS_{AP} at heel strike	**< 0.001 (0.287)**	**0.003 (0.690)**	0.098 (0.017)	**< 0.001 (0.727)**
MoS_{ML} at heel strike	**< 0.001 (0.097)**	**< 0.001 (0.606)**	**0.038 (0.023)**	0.570 (0.002)

Figure captions

Figure 1. (a) the backX™ (www.suitx.com) used in the study; arrows indicate external trunk/hip extension torque provided by the exoskeleton, and (b) illustration of experimental trials during over-ground walking.

Figure 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.

Figure 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*. Levels not connected by the same letters are significantly different. 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. 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. Levels not connected by the same letters are significantly different. 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 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. Lower case letters show results from pairwise comparisons between *EXO* conditions within *GEN*. Levels not connected by the same letters are significantly different. 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.

Figures

Figure 1

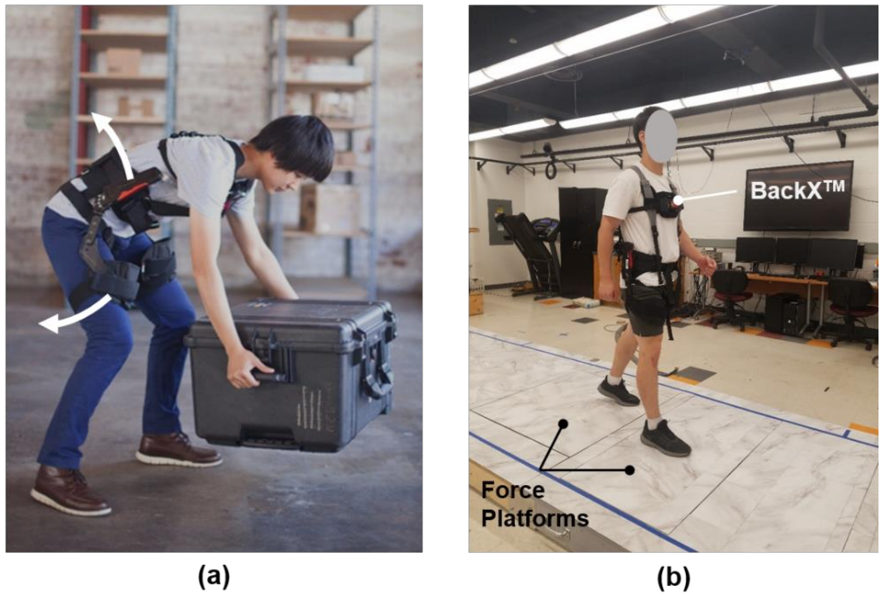


Figure 2

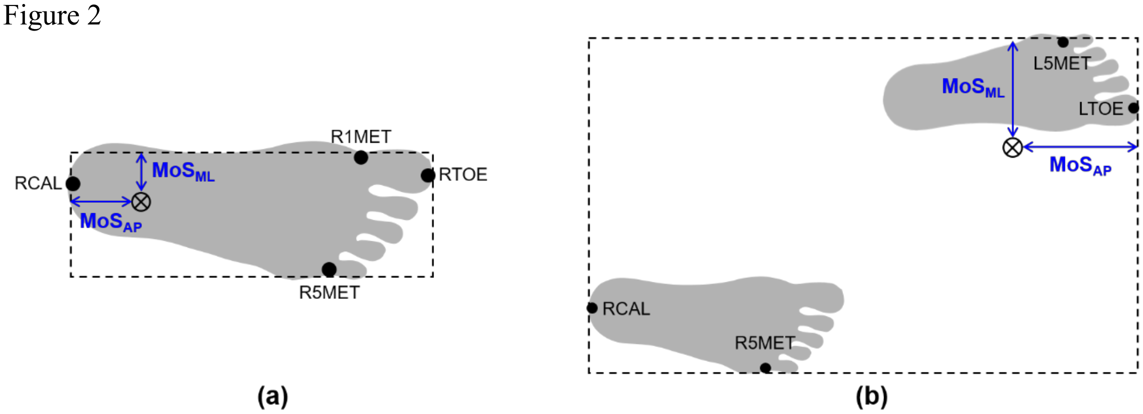
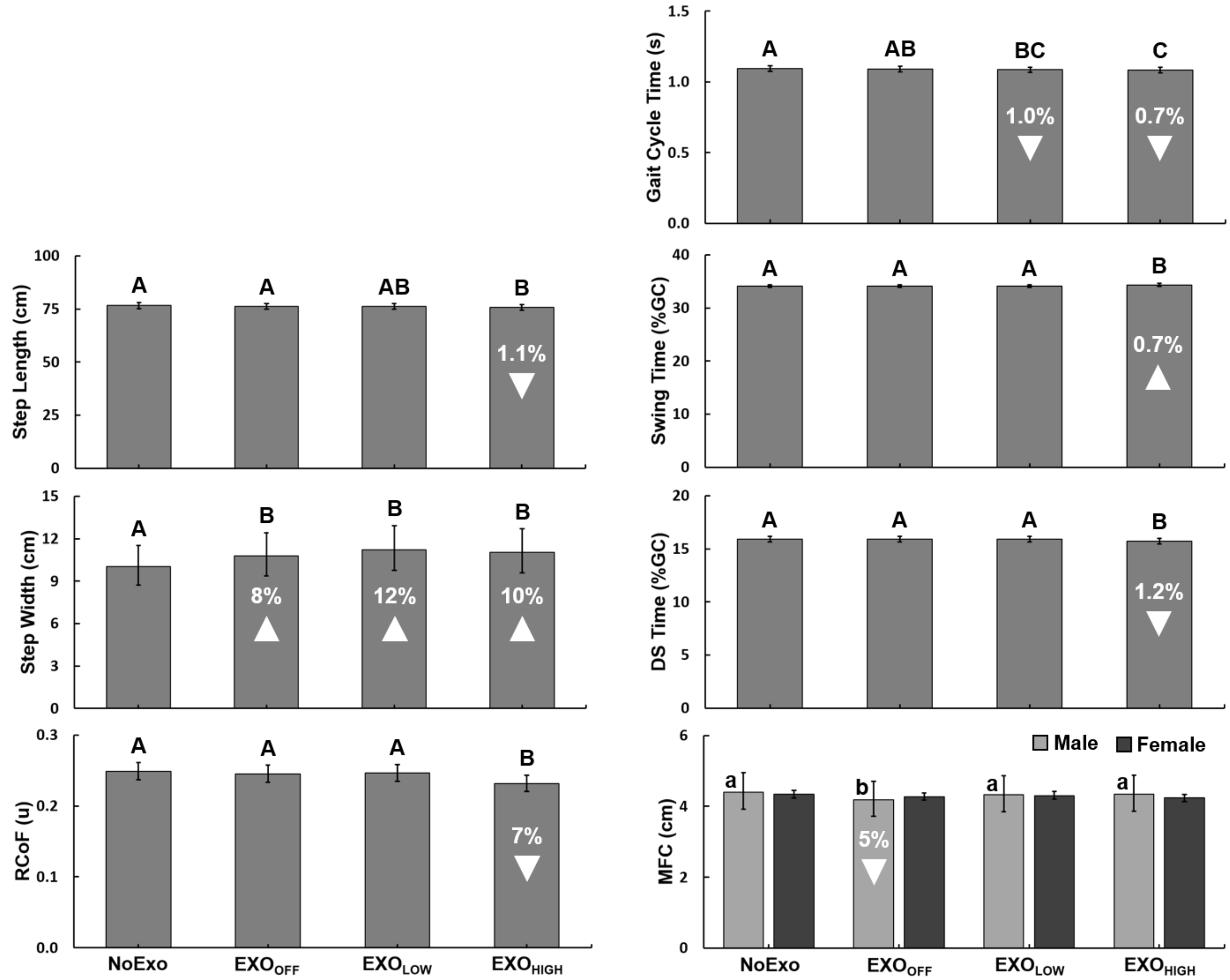
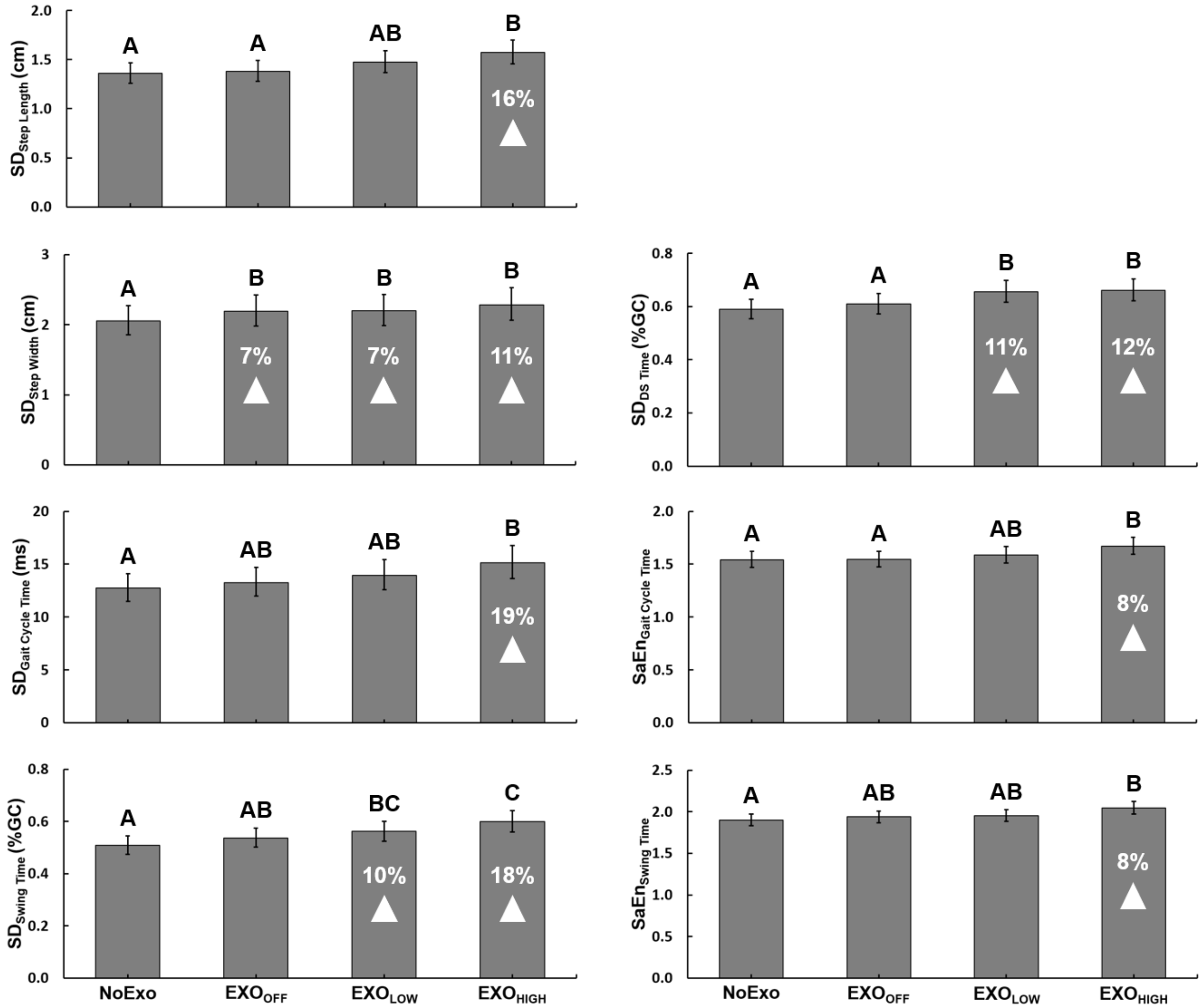


Figure 3



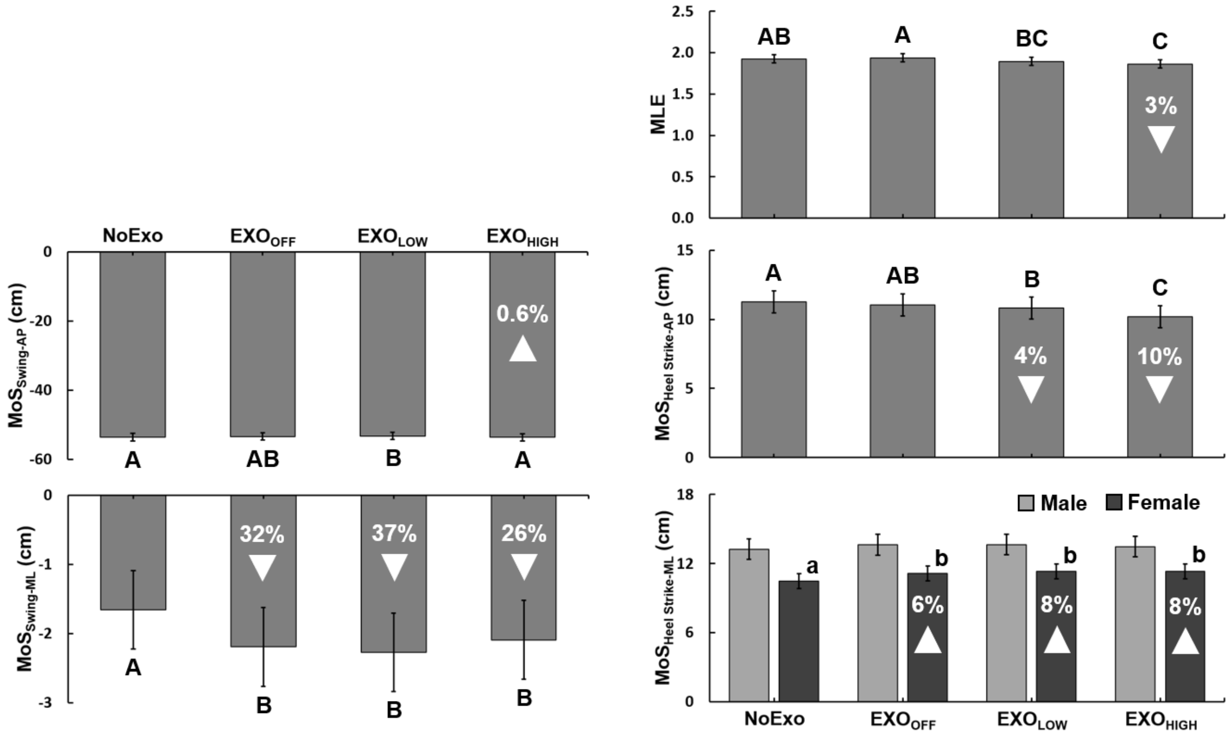
Gait Cycle Time (s)
A
AB
BC
C
1.0%
0.7%
Step Length (cm)
A
A
AB
B
1.1%
Swing Time (%GC)
A
A
A
B
0.7%
Step Width (cm)
A
B
B
B
8%
12%
10%
DS Time (%GC)
A
A
A
B
1.2%
RCoF (u)
A
A
A
B
7%
MFC (cm)
Male
Female
a
b
a
a
5%
NoExo
EXO_{OFF}
EXO_{LOW}
EXO_{HIGH}

Figure 4



SD_Step Length (cm)
A
A
AB
B
16%
SD_Step Width (cm)
A
B
B
B
7%
7%
11%
SD_DS Time (%GC)
A
A
B
B
11%
12%
SD_Gait Cycle Time (ms)
A
AB
AB
B
19%
SaEn_Gait Cycle Time
A
A
AB
B
8%
SD_Swing Time (%GC)
A
AB
BC
C
10%
18%
SaEn_Swing Time
A
AB
AB
B
8%
NoExo
EXO_OFF
EXO_LOW
EXO_HIGH
NoExo
EXO_OFF
EXO_LOW
EXO_HIGH

Figure 5



NoExo
EXO_OFF
EXO_LOW
EXO_HIGH
MoS_Swing-AP (cm)
0.6%
A
AB
B
A
MoS_Swing-ML (cm)
32%
37%
26%
A
B
B
B
MLE
AB
A
BC
C
3%
MoS_Heel Strike-AP (cm)
A
AB
B
C
4%
10%
Male
Female
MoS_Heel Strike-ML (cm)
a
b
b
b
6%
8%
8%

Appendices

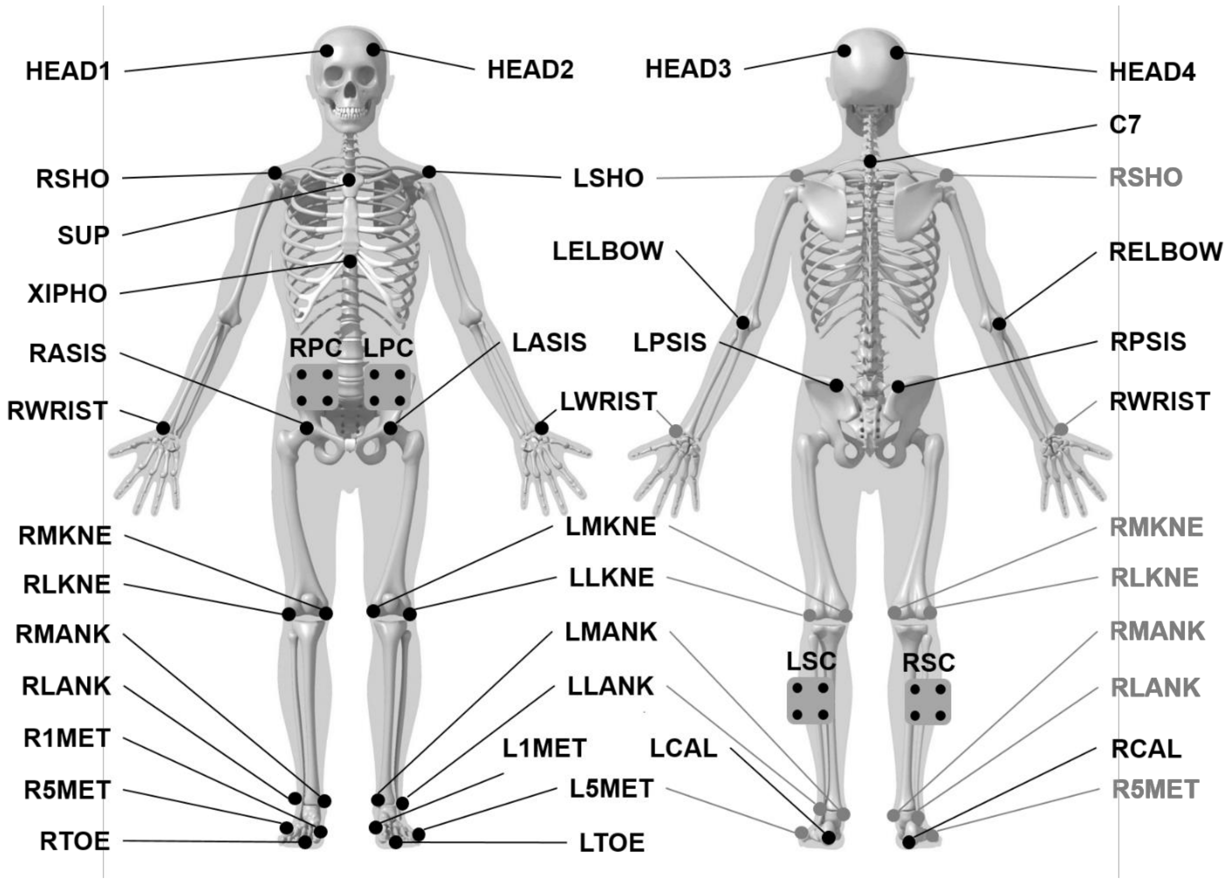


Figure A1. Anterior (left) and posterior (right) views of the marker set for gait kinematics (TOE = 2nd distal phalanx; CAL = calcaneus; 1MET = 1st metatarsal head; 5MET = 5th metatarsal head; ANK = malleoli; KNE = femoral epicondyle; ASIS = anterior superior iliac spine; PSIS = posterior superior iliac 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 backX^{TM} covered anatomical landmarks on pelvis (i.e., LASIS, RASIS, LPSIS, and RPSIS). Hence, pelvis clusters (i.e., LPC and PRC) attached on the pelvis belt of the backX^{TM} 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 Professional^{TM} (C-Motion Inc., Germantown, MD).

Table A1. Least squares means (95% confidence intervals) of all outcome measures.

Outcome measures	*EXO* condition			
	NoExo	EXO_{OFF}	EXO_{LOW}	EXO_{HIGH}
Spatiotemporal gait parameters				
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 time (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)
Slip risk				
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)
Trip risk				
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)
SD				
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 time (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)
SaEn				
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 time	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)
MLE				
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)
MoS				
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 A2. 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	*EXO* 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$_{OFF}$
Spatiotemporal gait parameters						
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 time	< **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
Slip risk						
RCoF	< **0.001**	**0.009**	**0.005**	0.940	0.995	0.846
Trip risk						
MFC	0.055	0.234	0.795	0.371	**0.028**	< **0.001**
SD						
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 time	**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
SaEn						
Gait cycle time	**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
MLE						
Trunk kinematics	**0.004**	< **0.001**	0.286	0.255	**0.049**	0.856
MoS						
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**

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

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Conflict of Interest: We declare that we have no commercial relationships that may lead to any conflict of interests.

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Abstract

Background: 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.

Research question: Does using a BSE adversely affect gait performance and stability, and are such effects dependent on specific BSE external torque characteristics?

Methods: Twenty participants (10M, 10F) completed five level over-ground walking trials and a five-minute treadmill walking trial while wearing a BSE (backX^{TM}) 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 (MLE) of trunk kinematics and the margin-of-stability (MoS).

Results: Using the backX^{TM} with high supportive torque decreased slip risk (7% decrease in RCoF) and slightly improved trunk stability (3% decrease in MLE). 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.

Significance: 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.

Key words: occupational exoskeleton; walking stability; workplace fall.

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., [1, 2]). 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 [3] 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) [2, 3]. Hip flexion torque generated during the swing phase of gait is a major contributor to the propulsive power needed for walking [4]. Specifically, ~12 Nm of hip flexion torque is required during the swing phase [5]. However, previous investigations have shown that passive BSEs can produce ~35 Nm of extension torque at each hip [6, 7]. A stable gait pattern requires appropriate magnitudes of lower limb joint torques generated at specific times for each gait event during the stance and swing phases [5]. Hence, external hip-extension torque generated by a BSE could compromise gait performance and stability, and thereby increase the risk of falling.

To assess gait performance, spatiotemporal gait parameters (e.g., step length, stance phase duration) were traditionally utilized. However, stride-to-stride variability in gait kinematic patterns has been increasingly utilized as an alternative measure of locomotion ability, as such variable is thought to reflect the ability of the central neuromuscular control system to regulate gait and maintain a steady walking pattern. In fact, gait variability is associated with instability and fall risk [8, 9]. Additionally, the required coefficient-of-friction (i.e., RCoF, the friction needed during walking) and the minimum foot clearance (MFC) are indicative of slip- and trip-induced fall risks, respectively. Previous investigations have shown that both RCoF and MFC are influenced by walking speed, joint kinematics, and/or leg joint torques, especially about the knee and the hip [10, 11].

To assess gait stability, the maximum Lyapunov exponent (MLE) and margin of stability (MoS) have been quantified. MLE indicates the ability of the motor system to attenuate small perturbations [12], and is a measure that is sensitive to changes in gait stability resulting from aging, neuro-atypical conditions, and various experimental conditions (e.g., [13, 14]). MoS is defined as the distance between the extrapolated center-of-mass (CoM) and the boundaries of the base-of-support [15]. MoS is also sensitive to differences in walking speed, walking environment, age, and pathological conditions (e.g., [16, 17]).

Exoskeletons for gait rehabilitation have been extensively examined in the literature (e.g., [18]). 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). Using an arm-support exoskeleton was found to increase the velocity of the center-of-pressure during quiet standing [19], possibly due to elevation of the CoM of the body + exoskeleton system. Although not directly comparable, a recent study of a powered whole-body exoskeleton [20] reported that wearing the exoskeleton was associated with slower gait and greater role of the hip joint in regulating gait, as evidenced by greater hip range-of-motion and contributions of hip joint motion to gait principal components. Using a passive BSE led to a slower preferred gait speed and shorter stride length [2]. We recently investigated the effects of two different passive BSEs on postural balance during quiet upright stance and found a decrease in postural balance [21].

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. 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 MLE and decrease in MoS). We also hypothesized that such changes will be more pronounced as the external BSE-generated torque increases.

2. Methods

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. The sample size was determined based on a *priori* power analysis using postural stability measures reported in previous investigations [22, 23], for a significance level $\alpha = 0.05$ and power $(1 - \beta) = 0.8$. 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.

2. 2. Exoskeleton selection

The backX^{TM} AC (US Bionics Inc., Berkeley, CA), designed to reduce physical demands on the back during forward bending [24], 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 1). The external torque generator includes gas springs that store energy during forward bending (i.e., trunk/hip flexion), and this energy is released and thereby provides trunk/hip extension torque and assistance, for example during lifting tasks [24]. The four exoskeleton (*EXO*) conditions included in this study were: NoExo (no exoskeleton, control condition), EXO_{OFF} (backX^{TM} with no supportive torque), EXO_{LOW} (backX^{TM} with low supportive torque) and EXO_{HIGH} (backX^{TM} with high supportive torque).

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 (e.g., RCoF), whereas the treadmill protocol allowed us to collect enough strides to compute other dependent measures (e.g., MLE).

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 [25]. 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 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 a procedure described earlier [26]. Briefly, participants were asked to stand at a starting position that was selected to allow each of their feet to land naturally on each force platform, and to walk across the track at the predetermined PWS. Although we confirmed foot placement, participants were not aware of where the force plates were located, so as to not disturb their natural walking. Participants completed five acceptable walking trials (i.e., foot placement on the force platform and at PWS) in each *EXO* condition. Mean (SD) of gait speed in NoExo, EXO_{OFF}, EXO_{LOW}, and EXO_{HIGH} conditions were 1.39 (0.15), 1.39 (0.16), 1.40 (0.15), and 1.40 (0.15), respectively. Gait speed did not significantly differ between *EXO* conditions (p=0.211) during over-ground walking. In the treadmill session, participants walked for five minutes on the treadmill at their PWS in each *EXO* condition (Figure 1 (b)). This walking duration was used to provide sufficient strides for subsequent analyses of gait kinematic variability and MLE [27, 28].

[Figure 1 here]

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 A1) were sampled at 120 Hz using a

12-camera optical motion capture system (Qualisys, Inc., Gothenburg, Sweden), then low-pass filtered (6 Hz cutoff; 4^{th}-order Butterworth; bidirectional). Ground reaction forces from the two force platforms were sampled at 1200 Hz, low-pass filtered (36 Hz cutoff; 2^{nd}-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; 4^{th}-order Butterworth; bidirectional) and used to compute gait variability and MLE. Gait events in each walking trial (i.e., heel-strike and toe-off of each foot) were detected using a coordinate-based algorithm [29]. Body segment CoMs were calculated using body segment scaling factors [30]. Three-dimensional rotations of each segment were obtained based on segmental coordination systems [30] and were expressed following 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 $r = 0.2$ [31]. Maximum RCoF was computed according to the method described in [11]. 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 MLE, trunk linear and angular velocities [28] were first calculated using 3-point differences and then scaled to body size [32]. 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 [14, 33]. 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 MLE [12, 33]. 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 [15]. MoS is defined as the minimum distance between XCoM and base-of-support borders [15]. 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 2). While a positive MoS value indicated that the XCoM was inside of the base-of-support boundary, negative MoS values indicated that the XCoM was outside of the boundary.

[Figure 2 here]

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. 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 and were qualitatively interpreted as 0.01 = small, 0.06 = medium, 0.14 = large [34].

3. Results

Descriptive summaries of the outcomes are included in the Appendix (Table A1). Table 1 summarizes ANOVA results, regarding *EXO* and *GEN* effects; these results are present in more detail for each group of outcomes. Table A2 presents summary of *post hoc* comparisons for the significant main effects of *EXO* on each outcome measure.

[Table 1 here]

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

EXO had significant main and/or interactive effects on all spatiotemporal gait measures (Table 1 and Figure 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 time 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*×*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).

RCoF and MFC differed significantly across *EXO* conditions (Table 1 and Figure 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 1). RCoF was 15% higher in males than in females.

[Figure 3 here]

3. 2. Gait variability

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

[Figure 4 here]

3. 3. Maximum Lyapunov exponent (MLE)

MLE of trunk kinematics during level walking differed significantly between *EXO* conditions (Table 1 and Figure 5). Specifically, MLE 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}.

3. 4. Margin of stability (MoS)

All MoS measures showed a significant main effect of *EXO* (Table 1 and Figure 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 *EXO* main effect and the $EXO \times GEN$ interaction effect were significant for MoS_{ML} at heel strike (Table 1 and Figure 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 1), and these measures were 27% and 22% higher in males than in females, respectively.

[Figure 5 here]

4. Discussion

4. 1. Gait performance

Walking with the backX™ 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). 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. Although changes such as reduced step length and increased swing time are generally detrimental, and indicate that individuals may have a harder time swinging their legs forward [35, 36], it is unclear whether these changes, which were rather small in our study (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). 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 [37, 38]. In fact, hip range-of-motion in the coronal plane significantly increased (8-9%) when wearing the backX™, compared to the control condition. While increased step width may lead to increased energetic costs in longer-term walking [2, 39], whether this strategy may actually be optimal when using a BSE needs to be investigated in future studies.

The EXO_{HIGH} condition caused a 6-7% decrease in RCoF (i.e., decreased slip risk) compared to the other conditions (Figure 3). When the maximum RCoF occurred (i.e., the instant between heel-strike and midstance [11]), the EXO_{HIGH} condition led to significant changes in foot angle (i.e., the angle between the foot segment and the ground in the sagittal plane; 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 backX™ substantially increased gait variability (7-19%), and the change in gait variability increased with the magnitude of external device torque (Figure 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 [40].

It may also be the case that participants were simply not adequately adapted to the exoskeleton, or additional familiarization/practice is needed. Similar magnitudes in stride-to-stride gait variability have been associated with an increased risk of falling among older adults [8, 9]. Our findings among young adults may not be directly comparable to results from older adults investigated in previous studies: while older adults may show changes in gait variability, they also show other changes such as reductions in force generation and sensory capacity that may affect their fall risk [41]. As younger individuals do not have such age-related impairments in motor or sensory capacity affecting postural control, it is unclear whether similar magnitudes of changes in gait variability will increase fall risk among younger groups as well.

4. 2. Gait stability

The EXO_{HIGH} condition significantly lowered the MLE values of the trunk compared to both the control and no-torque conditions (Figure 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 [42], 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 [28] 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 decreased MoS_{AP} compared to the control condition (Figure 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 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 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.

One important thing to note is that individual PWS was determined in the no-exoskeleton condition, and this speed was then maintained in all *EXO* conditions. However, individuals may naturally choose to walk slower when wearing a BSE, to help reduce gait variability and/or increase dynamic stability. In fact, results from our preliminary study, which investigated the effects wearing BSEs on PWS, showed that PWS decreased slightly in *EXO* conditions (mean difference between *EXO* conditions = 0.087 m/s, which is roughly a ~7% change in PWS). Hence, some of the changes found here may be explained by differences in natural PWS, and furthermore, allowing individuals to walk at their PWS could potentially mitigate some of the changes in postural stability measures assessed here.

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. However, for all outcome measures that showed both statistical significance and substantial differences with *EXO* use, most individuals (>85%) exhibited a consistent direction of the overall effects. 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.

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 may adversely affect walking energetics and fall risk.

Conflict of Interest Statement

The authors have no conflicts of interest to report.

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Tables

Table 1. Summary of ANOVA results for the main and interaction effects of *EXO*, *GEN*, and gait speed 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			
	EXO	*GEN*	*EXO × GEN*	Gait Speed
Spatiotemporal parameters				
Step length	**< 0.001 (0.063)**	0.115 (0.312)	**0.032 (0.024)**	**< 0.001 (0.496)**
Step width	**< 0.001 (0.132)**	0.842 (0.009)	0.218 (0.012)	0.137 (0.004)
Gait cycle time	**< 0.001 (0.105)**	0.058 (0.544)	0.087 (0.018)	**< 0.001 (0.549)**
Swing time	**< 0.001 (0.077)**	0.505 (0.040)	0.143 (0.014)	**< 0.001 (0.246)**
Double support time	**< 0.001 (0.116)**	0.756 (0.014)	**0.015 (0.027)**	**< 0.001 (0.323)**
Slip risk				
RCoF	**< 0.001 (0.049)**	**0.005 (0.149)**	0.957 (0.001)	0.593 (0.000)
Trip risk				
MFC	**< 0.001 (0.052)**	0.847 (0.005)	**0.013 (0.029)**	0.323 (0.002)
SD				
Step length	**< 0.001 (0.298)**	0.169 (0.000)	0.386 (0.054)	**0.004 (0.000)**
Step width	**< 0.001 (0.395)**	0.075 (0.000)	0.062 (0.126)	0.571 (0.000)
Gait cycle time	**0.012 (0.183)**	0.054 (0.000)	0.506 (0.042)	**< 0.001 (0.000)**
Swing time	**< 0.001 (0.422)**	0.756 (0.000)	0.634 (0.031)	**0.001 (0.000)**
Double support time	**< 0.001 (0.413)**	0.739 (0.000)	0.903 (0.010)	**0.006 (0.000)**
SaEn				
Step length	0.155 (0.092)	0.642 (0.000)	0.418 (0.051)	0.793 (0.000)
Step width	0.298 (0.065)	0.780 (0.000)	0.637 (0.031)	0.474 (0.000)
Gait cycle time	**0.005 (0.210)**	0.066 (0.000)	0.140 (0.096)	**< 0.001 (0.000)**
Swing time	**0.010 (0.189)**	0.062 (0.000)	0.614 (0.033)	**< 0.001 (0.000)**
Double support time	0.586 (0.035)	0.763 (0.000)	0.466 (0.046)	0.563 (0.000)
MLE				
Trunk kinematics	**< 0.001 (0.296)**	0.127 (0.000)	0.160 (0.087)	0.054 (0.000)
MoS				
MoS_{AP} during swing	**0.011 (0.030)**	0.398 (0.151)	0.142 (0.014)	**< 0.001 (0.850)**
MoS_{ML} during swing	**< 0.001 (0.052)**	0.096 (0.109)	0.539 (0.006)	0.704 (0.000)
MoS_{AP} at heel strike	**< 0.001 (0.287)**	**0.003 (0.690)**	0.098 (0.017)	**< 0.001 (0.727)**
MoS_{ML} at heel strike	**< 0.001 (0.097)**	**< 0.001 (0.606)**	**0.038 (0.023)**	0.570 (0.002)

Figure captions

Figure 1. (a) the backX^{TM} (www.suitx.com) used in the study; arrows indicate external trunk/hip extension torque provided by the exoskeleton, and (b) illustration of experimental trials during over-ground walking.

Figure 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.

Figure 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*. Levels not connected by the same letters are significantly different. 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. 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. Levels not connected by the same letters are significantly different. 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 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. Lower case letters show results from pairwise comparisons between *EXO* conditions within *GEN*. Levels not connected by the same letters are significantly different. 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.

Figures

Figure 1

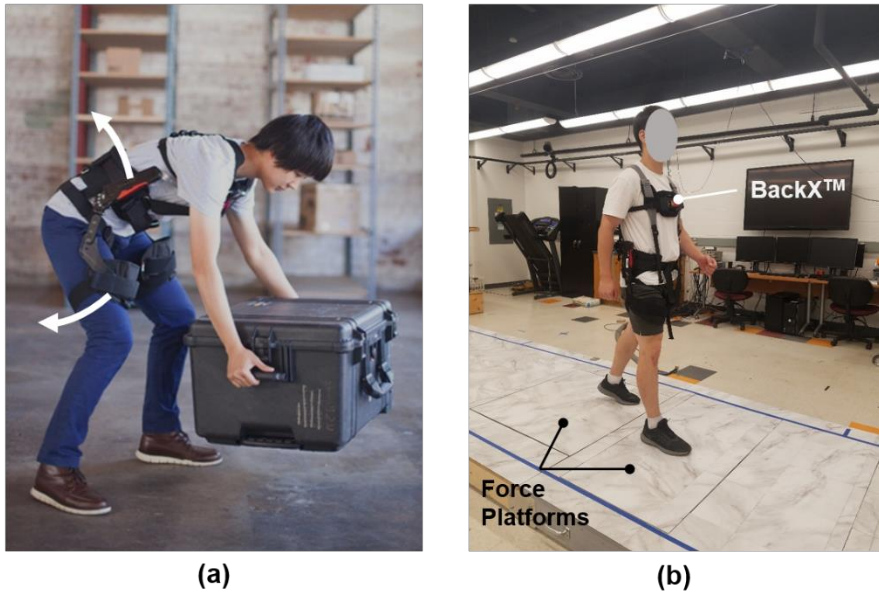


Figure 2

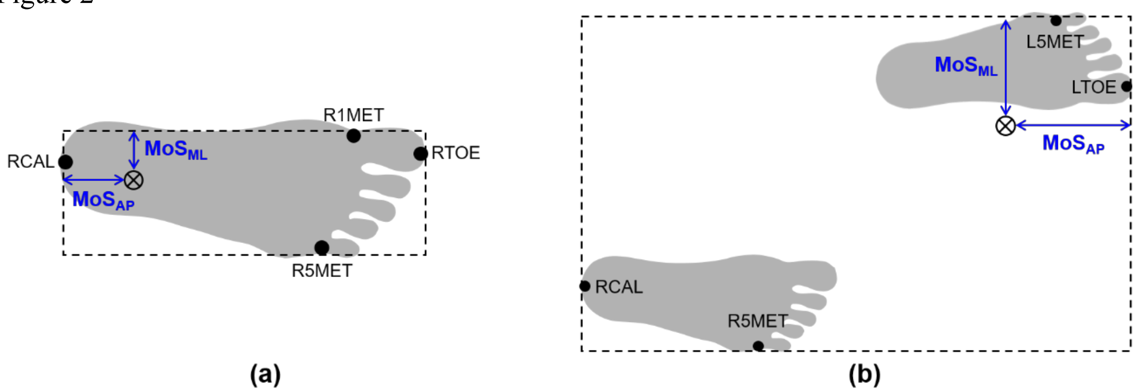
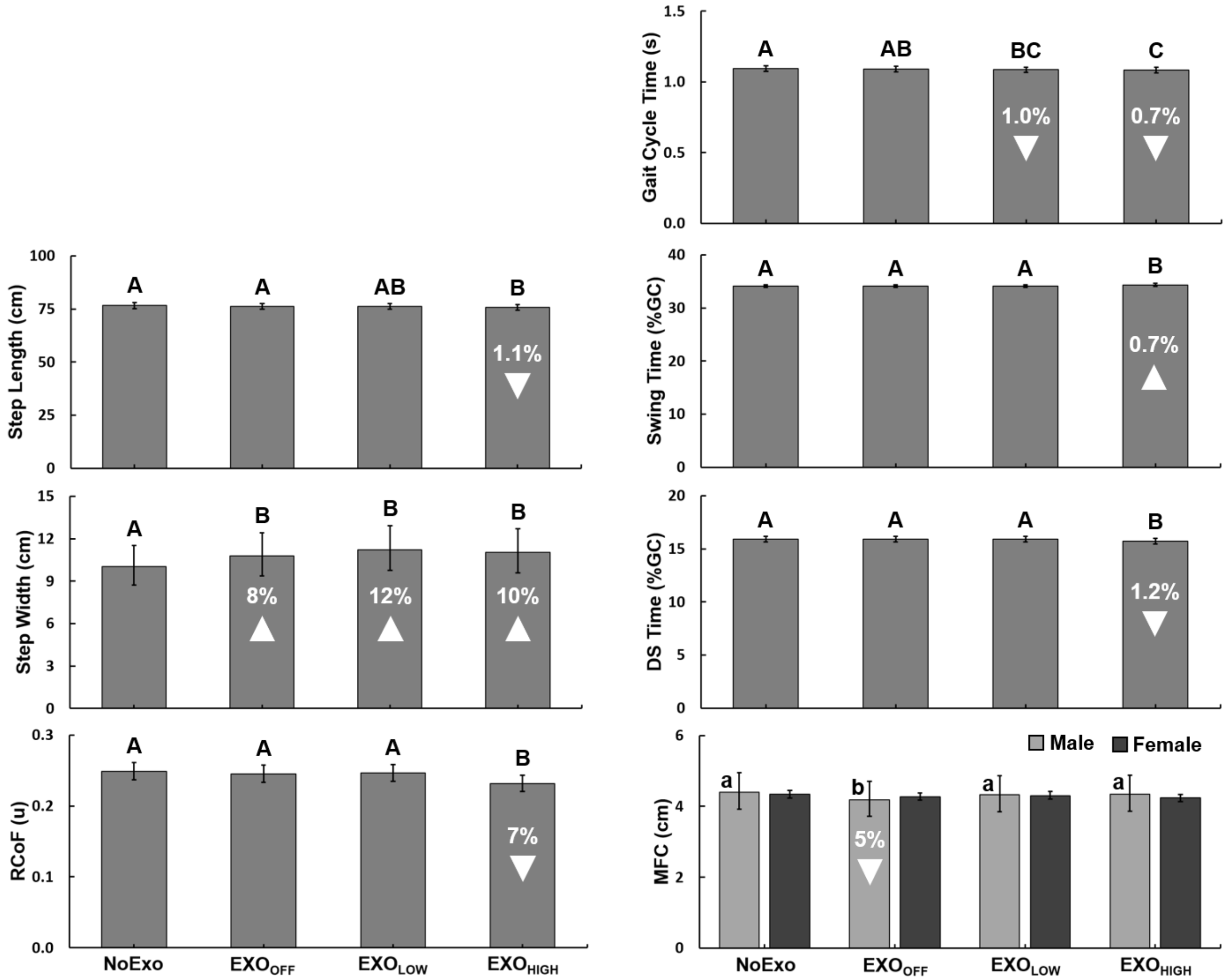
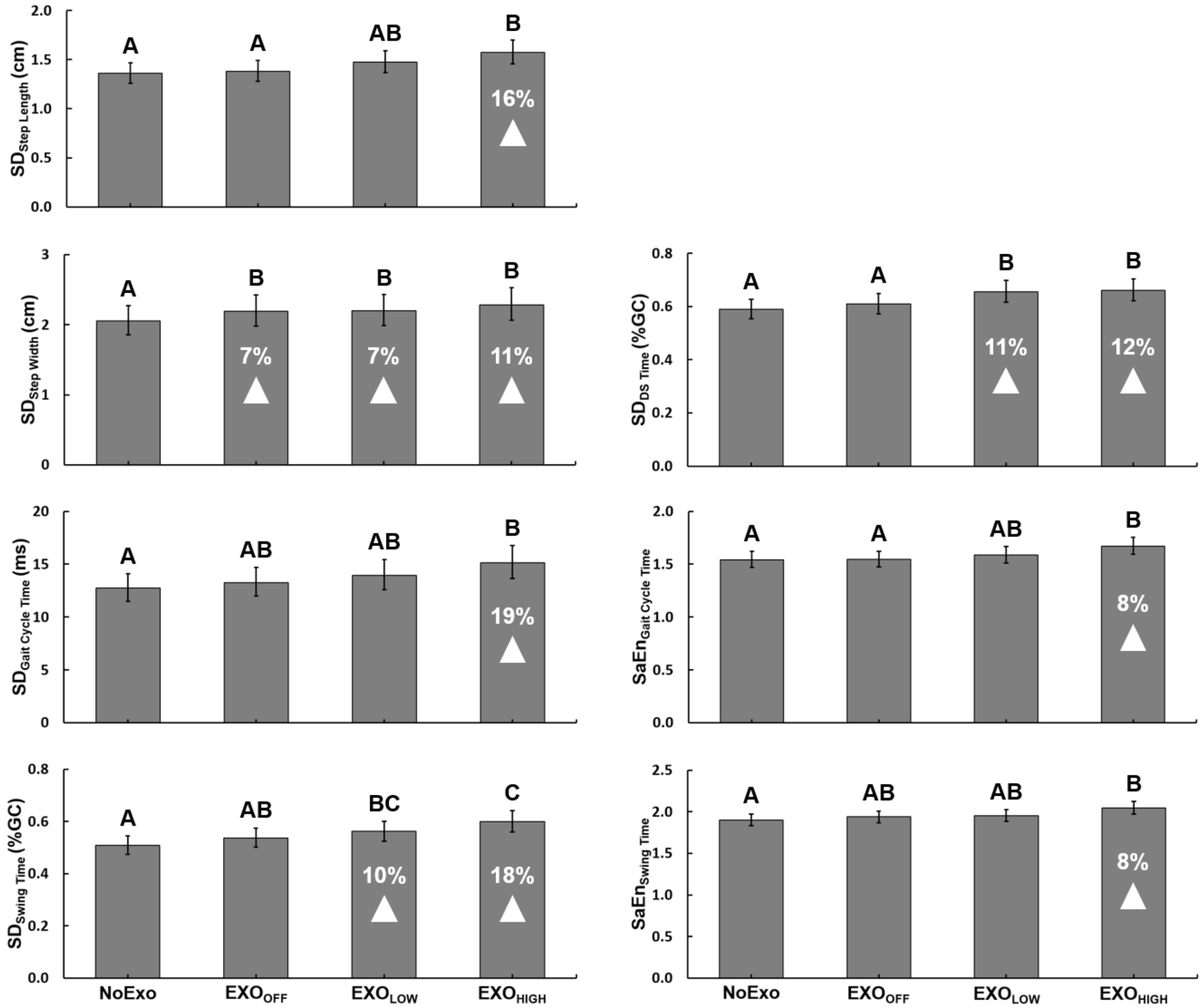


Figure 3



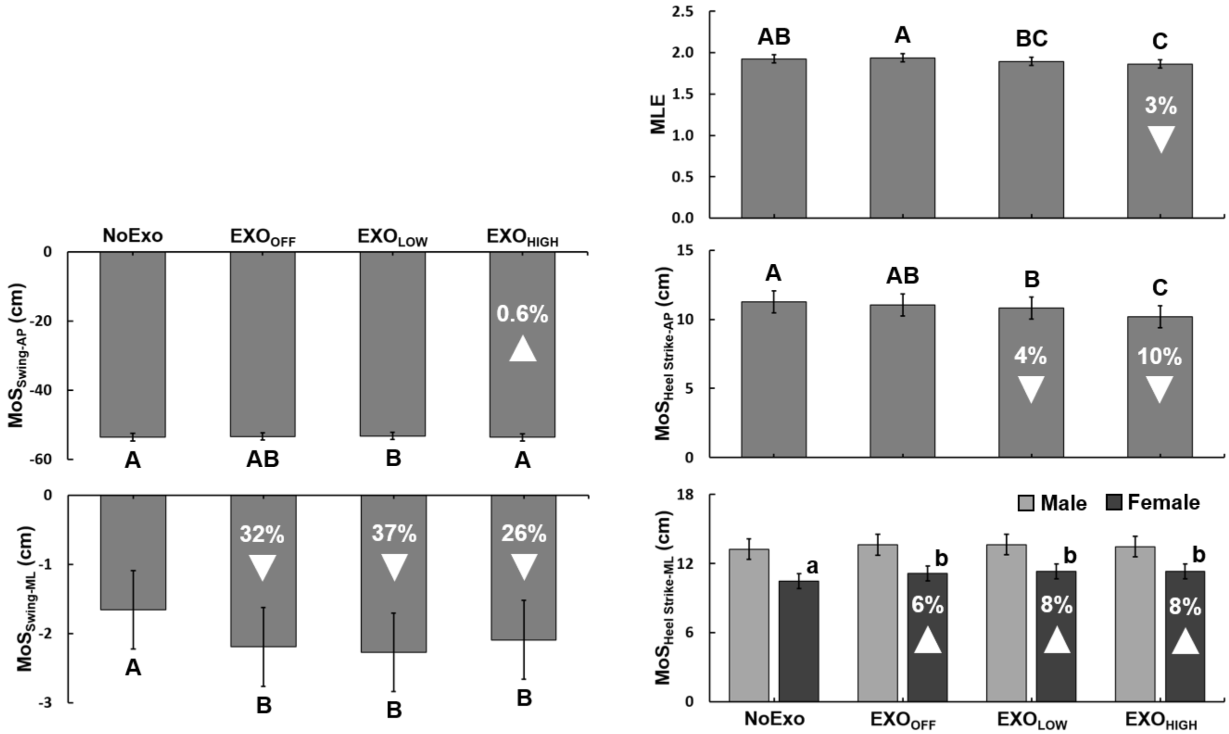
Gait Cycle Time (s)
1.5
1.0
0.5
0.0
A
AB
BC
C
1.0%
0.7%
Step Length (cm)
100
75
50
25
0
A
A
AB
B
1.1%
Swing Time (%GC)
40
30
20
10
0
A
A
A
B
0.7%
Step Width (cm)
15
12
9
6
3
0
A
B
B
B
8%
12%
10%
DS Time (%GC)
20
15
10
5
0
A
A
A
B
1.2%
RCoF (u)
0.3
0.2
0.1
0.0
A
A
A
B
7%
MFC (cm)
6
4
2
0
Male
Female
a
b
a
a
5%
NoExo
EXO_{OFF}
EXO_{LOW}
EXO_{HIGH}

Figure 4



$SD_{Step\ Length}$ (cm)
2.0
1.5
1.0
0.5
0.0
A
A
AB
B
16%
$SD_{Step\ Width}$ (cm)
3
2
1
0
A
B
B
B
7%
7%
11%
$SD_{DS\ Time}$ (%GC)
0.8
0.6
0.4
0.2
0.0
A
A
B
B
11%
12%
$SD_{Gait\ Cycle\ Time}$ (ms)
20
15
10
5
0
A
AB
AB
B
19%
$SaEn_{Gait\ Cycle\ Time}$
2.0
1.5
1.0
0.5
0.0
A
A
AB
B
8%
$SD_{Swing\ Time}$ (%GC)
0.8
0.6
0.4
0.2
0.0
A
AB
BC
C
10%
18%
$SaEn_{Swing\ Time}$
2.5
2.0
1.5
1.0
0.5
0.0
A
AB
AB
B
8%
NoExo
EXO_{OFF}
EXO_{LOW}
EXO_{HIGH}

Figure 5



NoExo
EXO_{OFF}
EXO_{LOW}
EXO_{HIGH}
$MoS_{Swing-AP}$ (cm)
0
-20
-40
-60
0.6%
A
AB
B
A
$MoS_{Swing-ML}$ (cm)
0
-1
-2
-3
32%
37%
26%
A
B
B
B
MLE
2.5
2.0
1.5
1.0
0.5
0.0
AB
A
BC
C
3%
$MoS_{Heel\ Strike-AP}$ (cm)
15
10
5
0
A
AB
B
C
4%
10%
$MoS_{Heel\ Strike-ML}$ (cm)
18
12
6
0
Male
Female
a
b
b
b
6%
8%
8%
NoExo
EXO_{OFF}
EXO_{LOW}
EXO_{HIGH}

Appendices

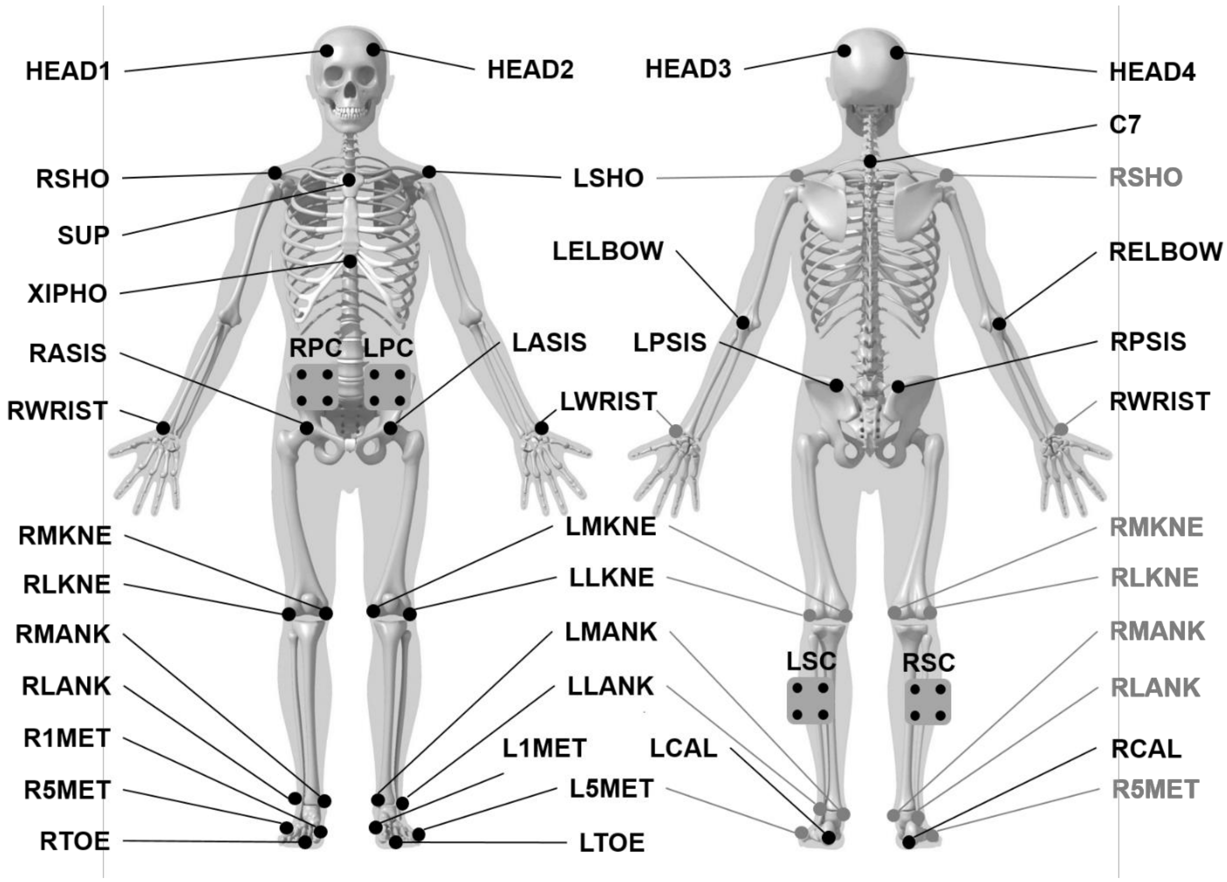


Figure A1. Anterior (left) and posterior (right) views of the marker set for gait kinematics (TOE = 2nd distal phalanx; CAL = calcaneus; 1MET = 1st metatarsal head; 5MET = 5th metatarsal head; ANK = malleoli; KNE = femoral epicondyle; ASIS = anterior superior iliac spine; PSIS = posterior superior iliac 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 backX^{TM} covered anatomical landmarks on pelvis (i.e., LASIS, RASIS, LPSIS, and RPSIS). Hence, pelvis clusters (i.e., LPC and PRC) attached on the pelvis belt of the backX^{TM} 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 Professional^{TM} (C-Motion Inc., Germantown, MD).

Table A1. Least squares means (95% confidence intervals) of all outcome measures.

Outcome measures	*EXO* condition			
	NoExo	EXO_{OFF}	EXO_{LOW}	EXO_{HIGH}
Spatiotemporal gait parameters				
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 time (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)
Slip risk				
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)
Trip risk				
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)
SD				
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 time (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)
SaEn				
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 time	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)
MLE				
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)
MoS				
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 A2. 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	*EXO* 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}
Spatiotemporal gait parameters						
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 time	**< 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
Slip risk						
RCoF	**< 0.001**	**0.009**	**0.005**	0.940	0.995	0.846
Trip risk						
MFC	0.055	0.234	0.795	0.371	**0.028**	**< 0.001**
SD						
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 time	**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
SaEn						
Gait cycle time	**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
MLE						
Trunk kinematics	**0.004**	**< 0.001**	0.286	0.255	**0.049**	0.856
MoS						
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**