TITLE PAGE

Full title:

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

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Abstract

We assessed the effects of using a passive back-support exoskeleton (BSE) on lower limb joint kinematics and kinetics during level walking. Twenty young, healthy participants completed level walking trials while wearing a BSE (backX^{TM}) with three different levels of hip-extension support torque (i.e., no torque, low, and high) and in a control condition (no-BSE). When hip extension torques were required for gait – initial 0-10% and final 75-100% of the gait cycle – the BSE with high supportive torque provided ~10 Nm of external hip extension torque at each hip, resulting in beneficial changes in participants' gait patterns. Specifically, there was a ~10% reduction in muscle-generated hip extension torque and ~15-20% reduction in extensor power. During the stance-swing transition, however, BSE use produced undesirable changes in lower limb kinematics (e.g., 5-20% increase in ankle joint velocity) and kinetics (e.g., ~10% increase in hip flexor, knee extensor, and ankle plantarflexor powers). These latter changes likely stemmed from the need to increase mechanical energy for propelling the leg into the swing phase. BSE use may thus increase the metabolic cost of walking. Whether such use also leads to muscle fatigue and/or postural instability in long-distance walking needs to be confirmed in future work.

Keywords: gait biomechanics, assistive device, hip extension torque, joint power

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.[13, 30] As work-related musculoskeletal disorders (WMSDs) of the low back continue to be a serious health problem,[9] 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.*[20] and Motmans *et al.*[23]). However, there also are concerns regarding unexpected or unintended challenges associated with BSE use, such as increased discomfort[10] and decreased agility[13] when wearing a BSE.

Given that BSEs are designed primarily to assist during physical tasks in manual material handling requiring lifting/lowering and/or stooped postures,[13, 30] understanding whether BSE use has any adverse effects when performing relevant ambulatory tasks such as walking and carrying is important. Recent studies have shown that walking while wearing a BSE was perceived as more difficult[3] and that both preferred speed and stride length decreased.[2] These effects may be due to the external torque (in hip extension) provided by the BSE when the included angle between the thigh and the trunk decreases. Indeed, Park *et al*.[25] found that wearing a BSE altered gait performance during level walking (e.g., increase in both step width and gait variability), and that such changes were more pronounced with higher external torque settings in the BSE.

While previous studies on BSE use have examined gait performance, there is limited understanding of the specific changes in lower limb joint kinematics and kinetics that lead to changes in gait spatiotemporal characteristics. Lower limb kinematics and kinetics are critical to understanding gait mechanics and energetics, and furthermore, abnormal joint kinematics

and/or kinetics are also associated with gait instability. For example, when compared with young adults, healthy older adults during walking exhibit distinct joint kinematics (e.g., increase in hip flexion angle during stance phase, decrease in foot angle at heel-strike, and decrease in late-stance ankle plantarflexion angle) and kinetics (e.g., increase in early-stance hip extensor power generation, late-stance hip flexor power generation, and late-stance knee power absorption).[6, 22, 32] Similarly, the design characteristics of ankle-foot orthoses have been studied based on the resulting changes in not just ankle, but also hip and knee kinetics and kinematics.[1, 16]

Although studies in other contexts (e.g., rehabilitation devices or gait assistive devices) suggest that the external torque provided by an exoskeleton at the hip may not substantially alter gait kinematics or kinetics during walking, these results are not directly applicable to occupational BSEs. For example, Lewis and Ferris[17] demonstrated that healthy individuals had 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 between approximately 1/3rd to 1/2nd of the gait cycle. Unlike rehabilitation devices, however, passive occupational BSEs are not designed explicitly to follow the human gait cycle. Hence, given the differences in design and/or supportive torque magnitudes between occupational BSEs and rehabilitation devices, and that little is known regarding the effects of occupational BSEs on lower limb joint kinematics and kinetics, these effects warrant further study.

Thus, we investigated the effects of different external torque levels of a passive BSE on lower limb joint (i.e., hip, knee, and ankle) kinematics and kinetics during level walking. Joint kinematics were quantified using range of motion (RoM) and angular velocities, while joint kinetics were investigated using torque and power. We hypothesized that wearing a BSE would decrease hip RoM and increase hip flexion torque, due to the external hip extension

torque generated by the device. Although the external torque constraints of a BSE are associated primarily with the hip joint, we also hypothesized that these constraints would cause secondary changes in other lower limb joint kinematics and kinetics, such as knee and ankle.[6, 18]

MATERIALS AND METHODS

Participants

A convenience sample of 20 gender-balanced, young, and healthy participants was recruited from the university and local community. Participants included 10 men 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 women 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 any data collection. Data reported here are from a larger study that included both treadmill and over-ground walking trials: results on gait performance and stability measures, primarily computed from treadmill walking trials, were published previously,[25] and this manuscript reports on lower limb joint kinematics and kinetics from over-ground walking trials.

Exoskeleton

The backX^{TM} AC (US Bionics Inc., Berkeley, CA), designed to reduce physical demands on the back during forward bending, was used in this study (Fig. 1 (a)). This exoskeleton has mass of 4.5 kg and consists of a waist strap, chest support, two thigh straps, and an external torque generator about each hip (i.e., bilateral torque-generating mechanism) that is coupled with the waist strap and the chest support. The four exoskeleton (*EXO*) conditions included in this study were: 1) NoExo (no exoskeleton, control condition); 2) EXO_{OFF} (backX^{TM} with no supportive torque); 3) EXO_{LOW} (backX^{TM} with low supportive torque); and 4) EXO_{HIGH} (backX^{TM} with high supportive torque). The EXO_{OFF} condition was included to isolate the effects of the additional mass and any movement restrictions 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 latter two conditions (EXO_{LOW} and EXO_{HIGH}). The torques applied by the exoskeleton device in the different settings were measured in Madinei,[21] that involved an isokinetic hip flexion/extension test protocol in a dynamometer, using the exoskeleton mounted on a mannequin. Fig. 1 (b) shows backX^{TM} torque profiles in EXO_{OFF}, EXO_{LOW}, and EXO_{HIGH} during isokinetic hip flexion and extension at 40 deg./s.

Experimental design and procedures

We used a repeated-measures design to assess the effects of different BSE torque levels on lower limb joint kinematics and kinetics during level walking. Level walking was performed on a linear track (1.5 m wide × 15.5 m long), in the middle of which were embedded two force platforms (AMTI, Boston, MA; and Bertec Corporation, Columbus, OH) that were level with the rest of the track. The presentation order of *EXO* conditions was

counterbalanced using 4×4 Balanced Latin Squares. All conditions were tested over a single session (~2 hours). They were first fitted with the backX™ following the manufacturer's recommendations and then were familiarized with the device.

Participants were asked to walk across the track at their predetermined preferred walking speed (PWS). To determine PWS, participants were asked to walk on a treadmill (T101, Horizon Fitness Inc., Cottage Grove, WI) without wearing the BSE, following the procedure reported by Jordan *et al*.[11] Briefly, participants walked on the treadmill at a randomly set, relatively slow speed, and then we increased the speed in 0.045 m/s increments until participants reported that the current speed was their PWS. Afterwards, we increased the speed by 0.045 m/s and then lowered it by 0.045 m/s to re-establish PWS. This procedure was repeated three times, and the speeds determined for each of the three trials were averaged to obtain the PWS. 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. Following this familiarization period, they then completed five acceptable walking trials (i.e., proper foot placement on the force platforms and at the PWS), during which experimental data were collected. Participants were not aware of where the force platforms were located, to not disturb their natural walking.

Instrumentation and outcome measures

Kinematics of the bilateral foot, shank, and thigh, as well as the pelvis and trunk, and ground reaction forces, were measured during the walking trials. Reflective markers (n=33) and four rigid marker clusters (Fig. 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). Triaxial ground reaction forces and torques from the force platforms were sampled at 1200 Hz, subsequently down-sampled to 120 Hz, and low-pass filtered (10 Hz cutoff; 2nd-order Butterworth; bi-directional). Subsequent data processing was performed using custom code in MATLAB (R2021b, The MathWorks Inc., Natick, MA).

Gait events in each walking trial (i.e., heel-strike and toe-off of each foot) were detected using a coordinate-based algorithm.[33] During each acceptable walking trial, one stride was extracted for further analysis, using a window starting from heel-strike on the first force platform. Three-dimensional rotations of each segment were defined using the Cardan $Y - x' - z''$ (tilt—obliquity—rotation) convention relative to the laboratory reference frame. Joint angles were calculated as Cardan angles between adjacent local segments with an order of rotation of flexion-extension, abduction-adduction, and internal-external rotation. Linear and angular accelerations of each body segment and joint angular velocities were calculated using a three-point difference equation.

Sagittal plane hip, knee, and ankle kinetics were assessed from the limb that contacted the first force platform only, assuming left/right symmetry. Body segment centers-of-mass were calculated using scaling factors provided in Dumas *et al.*[7] and used for representing the linear motions of each segment. Body segment inertial parameters were estimated using equations based on body segment mass and length.[7] Joint torques were calculated based on the ground reaction forces/torques and body segment kinematics, using a bottom-up 3D inverse dynamics model.[15] Joint torques were presented with respect to proximal segment coordinate systems. The calculated hip torques ($T_{\text{HIP-TOTAL}}$) included torque generated by participant plus the backX™. Hence, hip torque generated by the participant only ($T_{\text{HIP-HUMAN}}$)

was calculated by subtracting the torque generated by the device ($T_{HIP\text{-}BSE}$) from $T_{HIP\text{-}TOTAL}$. $T_{HIP\text{-}BSE}$ values for any given hip angle and angular velocity were estimated using a two-dimensional interpolation technique, applied on the backX™ torque profiles reported in Madinei.[21] 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[31]: 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). Joint torque and power quantities were normalized to body mass.

Outcome measures for joint kinematics included RoM at the hip, knee, and ankle, and peak velocities in flexion and extension at the hip, knee, and ankle. Hip angle was defined as the angle between the thigh and the trunk (i.e., trunk-based hip angle[14]), because external torques of the backX™ depend on the angle between the thigh and the trunk (Fig. 1 (b)). 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. Peak hip extension and flexion torques generated by participants plus BSE were also quantified. Outcome measures for joint power included peak values of H1 ($H1_{GENERATION}$), H2 ($H2_{ABSORPTION}$), H3 ($H3_{GENERATION}$), K1 ($K1_{ABSORPTION}$), K2 ($K2_{GENERATION}$), K3 ($K3_{ABSORPTION}$), K4 ($K4_{ABSORPTION}$), A1 ($A1_{ABSORPTION}$), and A2 ($A2_{GENERATION}$).

Statistical analysis

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 (Table 1). Effects of presentation order of the *EXO* conditions was explored initially. While this effect was significant for some measures, statistical significance of the main and interaction effects of *EXO* and *GEN* were unchanged. Hence, to keep the analysis consistent for all dependent measures, the presentation order of the *EXO* conditions was excluded in the final ANCOVA models. Significant effects were followed by *post hoc* pairwise comparisons using Bonferroni corrected *t*-tests, and significant interaction effects were further examined using simple effects analyses. Significant interaction effects were also visually examined to understand whether they were ordinal and important. If interactions were ordinal, (i.e., no cross-overs or opposing trends as a function of *GEN*), main *EXO* effects are presented; in cases of non-ordinal interactions, results of *EXO* effects are presented separately for each gender.[19] 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.[5] Summary results were back-transformed and presented in the original units below, 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 and with statistical significance concluded when $p < 0.05$.

RESULTS

Summary results across all walking trials and participants are shown in Fig. 2, including sagittal plane angles, angular velocities, torques, and powers for the hip, knee, ankle in the NoExo and EXO_{HIGH} conditions. Fig. A2 shows hip extension torques generated during EXO_{OFF}, EXO_{LOW}, and EXO_{HIGH} conditions over a gait cycle. Table 1 summarizes ANCOVA results of *EXO* and *GEN* effects; these results are presented in more detail for each group of outcomes below. Tables A1 and A2 in the appendix include descriptive summaries of the outcomes and a summary of *post hoc* comparisons for the significant main effects of *EXO* on each outcome measure, respectively.

Joint RoM

There were significant main effects of *EXO* on 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} (Fig. 3). Simple effects analysis of the *EXO*×*GEN* interaction showed that, while all pairwise differences were significant for men, among women hip RoM was lower in EXO_{HIGH} than in EXO_{LOW} and was lower in EXO_{LOW} than in NoExo and EXO_{OFF} conditions. Knee RoM was lower in EXO_{HIGH} when compared with NoExo, and ankle RoM was lower in EXO_{HIGH} than in EXO_{OFF} and EXO_{LOW} conditions (Fig. 3).

Joint angular velocity

EXO had 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 (Fig. 4). In contrast, peak hip flexion velocity

was higher in the NoExo condition when compared with other conditions and was higher in EXO_{OFF} and EXO_{LOW} than in EXO_{HIGH} (Fig. 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 extension velocity was higher in NoExo when compared with EXO_{OFF} and EXO_{HIGH} conditions and was higher in EXO_{OFF} and EXO_{LOW} when compared with EXO_{HIGH} condition (Fig. 4).

Both the main effect of *EXO* and the interaction effect of *EXO*×*GEN* were significant for peak ankle plantarflexion and dorsiflexion velocities (Table 1). Specifically, peak ankle plantarflexion velocity increased in all exoskeleton conditions when compared with the NoExo condition (Fig. 4). Simple effects analysis showed that such changes were significant for both men and women, but women had a relatively larger increase (6-7%) than men (3-4%). Peak ankle dorsiflexion velocity was least in NoExo, higher in EXO_{OFF}, still higher in EXO_{LOW}, and highest in EXO_{HIGH} (Fig. 4). Peak ankle dorsiflexion velocity was lower in NoExo than in both EXO_{LOW} and EXO_{HIGH}, and was lower in EXO_{OFF} than in EXO_{HIGH} among men. However, peak ankle dorsiflexion velocity was lower in the NoExo condition when compared with other conditions, and was lower in EXO_{OFF} and EXO_{LOW} than in EXO_{HIGH} among women. There were significant main effects of *GEN* on peak hip flexion, peak ankle plantarflexion, and peak ankle dorsiflexion velocities (Table 1), all were higher among women.

Joint torque

Significant main effects of *EXO* were found on both peak hip extension and flexion torques (Table 1). Specifically, peak hip extension torque was lower in EXO_{HIGH} condition when compared with other conditions, while peak hip flexion torque was lower in EXO_{LOW} when

compared with no exoskeleton control condition (Fig. 5). While peak hip extension torque generated by participants plus BSE did not differ between *EXO* conditions (p=0.26), results from peak hip flexion torque generated by participants plus BSE were identical with peak hip flexion torque generated by participants only.

There were significant mains effect of *EXO* on 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} (Fig. 5). However, peak knee flexion torque was higher in the control condition when compared with other conditions and was higher in EXO_{OFF} than in EXO_{LOW} and EXO_{HIGH} (Fig. 5).

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 men did not show any significant differences between *EXO* conditions, among women peak ankle plantarflexion torque was lower in the control condition than in EXO_{LOW} and was lower in EXO_{HIGH} when compared with EXO_{LOW} (Fig. 5). There were significant main effects of *GEN* on peak knee flexion and peak ankle plantarflexion torques (Table 1), and both were higher among men.

Joint power

EXO had a significant main effect on both $H1_{GENERATION}$ and $H2_{ABSORPTION}$ peak powers (Table 1). Specifically, $H1_{GENERATION}$ peak power was higher in EXO_{OFF} when compared with other conditions, and was lower in EXO_{HIGH} than in other conditions (Fig. 6). $H2_{ABSORPTION}$ peak power was higher in NoExo than in EXO_{OFF} and EXO_{LOW}, and was higher in EXO_{HIGH} when

compared with EXO_{OFF} (Fig. 6). Both the main effect of *EXO* and the *EXO×GEN* interaction effect were significant for $H3_{GENERATION}$ peak power. Although $H3_{GENERATION}$ 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 women, whereas $H3_{GENERATION}$ peak power in men was lower in EXO_{OFF} when compared with NoExo and EXO_{HIGH} conditions (Fig. 6).

While $K1_{GENERATION}$ peak power did not show any significant main or interaction effects, significant main effects of *EXO* were found for both $K2_{GENERATION}$ and $K4_{ABSORPTION}$ peak powers (Table 1). *Post hoc* analysis showed that $K2_{GENERATION}$ peak power was lower in EXO_{HIGH} when compared with the other conditions (Fig. 7). $K4_{ABSORPTION}$ peak power was higher in the control condition when compared with other conditions, was higher in EXO_{OFF} than in EXO_{LOW} and EXO_{HIGH}, and was higher in EXO_{LOW} than in EXO_{HIGH} (Fig. 7). Both the main effect of *EXO* and the *EXO×GEN* interaction effect were significant for $K3_{ABSORPTION}$ peak power (Table 1). Simple effects analysis revealed that $K3_{ABSORPTION}$ peak power was lower in the control condition than in EXO_{OFF} only among men, whereas $K3_{ABSORPTION}$ peak power among women was lower in the control condition when compared with the other conditions, and was lower in EXO_{OFF} than in EXO_{HIGH} (Fig. 7).

Significant main effects of *EXO* were found for 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} (Fig. 7). $A2_{GENERATION}$ peak power was lower in the control condition when compared with other conditions among both men and women (Fig. 7), but the difference in $A2_{GENERATION}$ peak power between NoExo and EXO_{OFF} was not significant for men. There were significant main effects of *GEN* on $K3_{ABSORPTION}$ and $A1_{ABSORPTION}$ peak powers (Table 1), and both were higher among men.

DISCUSSION

We evaluated the effects of different BSE torque-assistance levels on lower limb joint kinematics and kinetics during level walking, among a young and healthy participant group. We hypothesized that wearing a BSE would decrease hip RoM and increase hip flexion torque. We also hypothesized that wearing a BSE would cause secondary changes in other lower limb joint kinematics and kinetics, such as knee and ankle. Our results showed that wearing a BSE decreased hip RoM but had minimal effect on hip flexion torque. However, wearing a BSE altered hip extension torque and led to additional changes in knee and ankle kinematics and kinetics. Hence, our second hypothesis was confirmed. Overall, our results suggest that wearing a BSE had beneficial effects during the initial and final of the gait cycle. However, wearing a BSE led to more mechanical energy generation (by the hip and ankle) and absorption (by the knee) from ~60% gait cycle to swing forward, and helps to explain why an earlier study reported that wearing a BSE increased metabolic costs during walking.[2]

Specifically, wearing the BSE with high supportive torque (EXO_{HIGH}) was slightly beneficial to hip *extension* motion during gait, in that peak hip extension torque was lower (9%) in EXO_{HIGH} when compared with the NoExo condition (Fig. 5). From Fig. 2, it can be seen 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 BSE generated ~10 Nm of external hip extension torque (Figs. 1 (b) and A1), 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. Consequently, there was also a 12-22% decrease in $H1_{GENERATION}$ peak power in the EXO_{HIGH} condition (Fig. 6).

There was, however, a 11-19% increase in $H1_{GENERATION}$ peak power in the EXO_{OFF} condition, when compared with the no-exoskeleton control condition (Fig. 6). We believe that participants may have generated more H1 power in the EXO_{OFF} condition to compensate for the added mass and/or inertial effects of the device during the stance phase.

Peak hip flexion torque did not differ between NoExo and EXO_{HIGH} conditions (Fig. 5). During the times when hip flexion torque contributes to gait (i.e., during 15-70% of the gait cycle; Fig. 2), the hip was flexed less than ~10° or mostly extended. At this low degree of flexion, the BSE generated less than ~1.6 Nm of external hip extension torque (Figs. 1 (b) and A1). Hence, wearing the BSE had minimal effect on flexion torque at the hip during walking. It is important to note, however, that the current 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 substantially more flexed than while walking. Although hip flexion torque did not increase with exoskeleton use, among women, $H3_{GENERATION}$ peak power increased in both the EXO_{LOW} and EXO_{HIGH} (7-11%) conditions, when compared with NoExo and EXO_{OFF} conditions (Fig. 6). Hip flexor power (H3) is generated primarily to propel the leg into the swing phase by adding mechanical energy to the limb.[8, 26] Thus, women may have made more effort to increase $H3_{GENERATION}$ power in both EXO_{LOW} and EXO_{HIGH} conditions against the external hip extension torque provided by the BSE at ~65% of the gait cycle (i.e., when hip is flexed; Fig. 2).

There was an overall decrease in Hip RoM by 8% in EXO_{HIGH} when compared with the control condition across all participants (Fig. 3). EXO_{HIGH} also led to a 7% decrease in peak hip flexion velocity (Fig. 4). Both changes may indicate that wearing the BSE with high supportive torque impeded hip flexion during walking.

EXO_{HIGH} led to a slight (6%) decrease in peak knee extension torque (Fig. 5); this change may reflect the co-dependence between the hip and knee as they extend during the initial phase of gait. Since peak knee extension torque occurred at about 10% of the gait cycle (Fig. 2), the decrease in peak knee extension torque may have led to decreased $K2_{GENERATION}$ peak power (15-23%) in the EXO_{HIGH} condition (Fig. 7). Previous work indicates that $K2_{GENERATION}$ power is mainly produced by concentric action of the quadriceps to extend the knee prior to heel-off,[26, 27] which agrees 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} when compared with NoExo (p<0.001) and EXO_{OFF} (p<0.001).

$K3_{ABSORPTION}$ peak power increased by 9-13% in all exoskeleton conditions only among women (Fig. 7). The role of $K3_{ABSORPTION}$ power is to prevent knee collapse in late stance and early swing through eccentric rectus femoris contraction.[8] Hence, an increase in $K3_{ABSORPTION}$ peak power indicates that wearing the BSE led to greater energy transfer from the knee to the hip near 60% of the gait cycle, resulting in increased hip flexor power to propel the leg into swing.[24] Our results also show that $H3_{GENERATION}$ peak power increased in the EXO_{LOW} and EXO_{HIGH} conditions when compared with the control condition among women (Fig. 6).

When compared with the control condition, wearing the BSE decreased peak knee flexion torque by 5-11%. We believe that the modified walking patterns (i.e., decrease in peak hip extension torque), occurring when the BSE generated hip extension torque, resulted in synchronized changes in the knee, through a decrease in peak knee flexion torque. This speculation is further supported by the fact that the decreased activation in biceps femoris (a biarticular muscle) to decrease hip extension torque can subsequently lead to a decrease in knee flexion torque during walking.[12] Knee flexion torque, especially during terminal swing, has a role in reducing knee extension velocity, resulting in energy absorption.[28] Our results

showed corresponding changes in that peak knee extension velocity decreased in EXO_{OFF} and EXO_{HIGH} when compared with the NoExo condition (Fig. 4).

At the ankle joint, the EXO_{LOW} condition slightly increased peak ankle plantarflexion torque by 3% when compared with the control condition, but only among women, whereas men did not present such differences between *EXO* conditions (Fig. 5). Ankle plantarflexion torque is the most important energy generation source during the gait cycle.[4] However, whether this 3% increase among women 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 an 8-14% increase in $A1_{ABSORPTION}$ peak power when compared with both NoExo and EXO_{HIGH} (Fig. 7). As discussed in the hip joint section, this increase may have occurred because participants needed to absorb more A1 power when the BSE was worn in conditions where the device generated no or low torque (EXO_{OFF} and EXO_{LOW} conditions), to compensate for the added mass and/or 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.[29]

All exoskeleton conditions led to increases in both peak ankle plantarflexion (4-5%) and dorsiflexion (7-19%) velocity when compared with the control condition (Fig. 4). Such increases may have resulted from an increase in $A2_{GENERATION}$ peak power (6-8%) in all exoskeleton conditions (Fig. 7). $A2_{GENERATION}$ 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. Hence, we believe that increased $A2_{GENERATION}$ peak power indicates synchronized changes with increased $K3_{ABSORPTION}$ and $H3_{GENERATION}$ peak powers when wearing the BSE.

Gender differences

A significant and non-ordinal *EXO×GEN* interaction effect was evident for peak ankle plantarflexion torque, $H3_{GENERATION}$, and $K3_{ABSORPTION}$ peak power. Overall, the effects of the external torque level of the backX™ on these measures were larger among women (Figs. 5-7), indicating that women generated and absorbed relatively more mechanical energy during level walking while wearing a BSE. Interestingly, there were no significant gender differences in anthropometric measures (i.e., stature, mass, chest circumference, waist-hip ratio, and thigh circumference) or walking speed in this study. While gender differences in strength (e.g., maximum hip flexion torque) may have influenced our findings, this needs to be verified in follow-up studies, and suggests future studies to explore the relationship between hip strength and the metabolic energy expenditure associated with BSE use.

Limitations and future work

This study has some limitations common with most laboratory-based research conducted on exoskeleton use among healthy individuals. These limitations include: individual differences in BSE fit/adjustments may have influenced our findings, our focus was only on the short-term effects of BSE use, and only healthy and young individuals were included in our study. Level walking was studied in only straight-walking conditions. In addition, the backX™ torque profiles, reported by Madinei,[21] was obtained using a dynamometer and a mannequin, over 0-120 deg. of hip flexion/extension, covering the range of hip excursions that are typical during gait, but only at discrete speeds in steps of 20 deg./s.

Thus, there may have been some errors in our torque estimates at the interpolated velocities. Also, although we estimated the torque being generated by the device, relative motion and misalignment between the device and human anatomy may have affected the effective torques transferred to the humans. Finally, lower limb muscle activities were not measured here, and hence any changes in muscle coordination strategies or co-contractions could not be investigated. Future work extending kinetic analyses to other tasks, such as turning, obstacle crossing, stairclimbing, or recovering from perturbations, would offer more complete evidence of how wearing a BSE might impact postural control during commonly encountered occupational activities. Inclusion of older adults would also help evaluate the effects of BSE use on a more diverse sample, with consideration of differences in strength, joint mobility, and coordination strategies.

Conclusions

When hip extension torques were required for walking (i.e., the initial 0-10% and final 75-100% of the gait cycle), the backX™ exoskeleton, with high supportive torque, generated ~10 Nm of external hip extension torque. Participants generally showed positive adaptations in gait patterns during these phases, by decreasing hip extension torque and extensor power. Such adaptations, in turn, might have led to additional changes including decreases in knee extension velocity, flexion torque, and extensor power. However, participants increased hip extensor power and ankle dorsiflexor power during the phases when the exoskeleton did not provide high external torque, probably to compensate for the added mass and/or inertial effects of the device during the stance phase. During the stance-swing transition, participants showed undesirable 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), which likely occurred to increase mechanical energy generation for propelling the leg into the swing phase. Furthermore, wearing the backX™ impeded hip flexion, as indicated by decreased hip RoM and peak hip flexion velocity. Wearing a BSE thus could increase metabolic costs during walking. Whether this also leads to muscle fatigue and/or postural instability in long-distance walking needs to be confirmed in future work.

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CONFLICT OF INTEREST

The authors declare that they have no conflicts of interest.

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