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Full title:

Wearing a back-support exoskeleton alters lower-limb joint kinetics during single-step recovery following a forward loss of balance

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Abstract

We assessed the effects of a passive, back-support exoskeleton (BSE) on lower-limb joint kinetics during the initiation and swing phases of recovery from a forward loss of balance. Sixteen (8M, 8F) young, healthy participants were released from static forward-leaning postures and attempted to recover their balance with a single-step while wearing a BSE (backX^{TM}) with different levels of support torque and in a control condition. The BSE provided ~15–20 Nm of external hip extension torque on the stepping leg at the end of initiation and beginning of swing phases. Participants were unable to generate sufficient hip flexion torque, power, and work to counteract this external torque, although they sustained hip flexion torque for a more prolonged period, resulting in slightly increased hip contribution to positive leg work (compared to control). However, net positive leg work, and the net contribution of hip joint (human + BSE) to total leg work decreased with BSE use. While all participants had changes in hip joint kinetics, a significant compensatory increase in ankle contribution to positive leg work was observed only among females. Our results suggest that BSE use adversely affects reactive stepping by decreasing the stepping leg kinetic energy for forward propulsion, and that the relative contributions of lower-limb joints to total mechanical work done during balance recovery are altered by BSE use. BSEs may thus need to be implemented with caution for dynamic tasks in occupational settings, as they may impair balance recovery following a forward loss of balance.

Keywords: Occupational exoskeleton, falls, reactive stepping, muscle power

1. Introduction

Occupational back-support exoskeletons (BSEs) are being introduced as a new workplace intervention to reduce physical demands on the low back (Ali et al., 2021; Kermavnar et al., 2021). BSEs are wearable devices designed to support, augment, and/or assist the back and hip muscles, by producing restorative torques (i.e., hip/back extension torques). There have been considerable efforts reported that demonstrate the benefits of BSE use through laboratory-based (e.g., Dos Anjos et al., 2022) as well as field studies (e.g., Siedl and Mara, 2021). However, the need to identify any potential adverse consequences of BSE use (e.g., increased risk of falls) has also been emphasized for the safe adoption of BSEs in the workplace (Nussbaum et al., 2019).

An increase in fall risk resulting from wearing a BSE may be a critical safety concern as the consequence of occupational falls can lead to injuries, reduced workplace productivity, and increased economic burden (Kodithuwakku Arachchige et al., 2021). Falls commonly occur due to some unexpected external perturbations during locomotion such as tripping or slipping while walking (Berg et al., 1997; Heijnen and Rietdyk, 2016). Given that such external perturbations can cause a loss of balance, the ability to respond effectively to a loss of balance is an important factor in preventing falls. While postural balance can be recovered by non-stepping ankle or hip strategies when the perturbations are relatively small (Horak and Nashner, 1986), a stepping response is required to expand and re-establish the base-of-support following larger postural perturbations such as those encountered during slips, trips, pushes, and accelerations of the support surface (Aftab et al., 2016).

Several lower-limb exoskeletons have been developed to assist reactive stepping. For example, Monaco et al. (2017) showed an powered pelvis orthosis, that can detect slippages and provide counteracting torques at the hip, improves postural stability when elderly people and/or transfemoral amputees were exposed to unexpected slip-like perturbations. Zha et al. (2019) also reported that wearing a powered hip exoskeleton helps regain single-step balance from forward or backward disturbance by detecting the balance loss and generating assistive torques at the hip. However, few studies have shown how wearing a BSE affects reactive stepping following large postural perturbations.

In an earlier study (Park et al., 2022b), we investigated the effects of wearing a passive, commercially-available BSE on reactive stepping behaviors after a forward loss of balance (i.e., simulated trips) using a tether-release protocol. We found that BSE use impaired reactive stepping by decreasing hip flexion angle and angular velocity, and that the BSE reduced dynamic stability when a high level of external supportive torque was engaged. These findings suggest that the external hip extension torque generated by a BSE may impede hip flexion of the wearer. However, while kinematic analysis describes the stepping behaviors constituting the recovery response, quantifying joint kinetics helps understand whether biomechanical strategies for balance recovery change with BSE use. Especially compared to level walking, the biomechanical demand of generating and absorbing power during recovery following trip-induced perturbations primarily involves the hip and knee joints, respectively, more than the ankle (Shokouhi et al., 2024). Thus, BSE use may substantially alter joint kinetics during balance recovery. Analyzing changes in lower-limb joint torque, power, and work during reactive stepping associated with BSE use would thereby enhance our understanding of the underlying mechanisms associated with successful recoveries. Specifically, such an analysis could reveal whether BSE use affects the total positive and negative work done by the stepping leg, and how BSE use alters the relative contributions of different lower limb joints to total mechanical work done during balance recovery. Thus, the goal of this study was to understand the impact of BSE use on lower limb joint torques, power, and work, with a focus on the relative contributions of the hip, knee, and ankle to the recovery response.

For this purpose, we performed an inverse dynamics analysis of data from Park et al. (2022b) to investigate the effects of wearing a BSE on lower-limb joint kinetics. In the tether-release protocol, a successful single-step recovery from a forward loss of balance typically involves an initiation phase (release onset to toe-off of the stepping foot), a swing phase (toe-off to heel-strike of the stepping foot), and a landing phase (heel-strike of the stepping foot to stabilization) (Fig. 1). The initiation and swing phases involve concentric contraction of the ankle plantar flexors and the hip flexors. Based on our previous kinematic analysis (Park et al., 2022b) showing that BSE use impeded hip flexion, we focused on the initiation and swing phases in this study. We hypothesized that wearing a BSE would decrease hip flexion torque, power, and positive work. Given that among older adults compensatory changes in joint torques and power were observed in the ankle joint to account for reduced hip flexion strength (Madigan, 2006), we also hypothesized that a BSE-induced reduction in hip flexion kinetics would increase the relative contribution of the ankle to total mechanical work done during balance recovery.

2. Methods

2.1. Participants

A convenience sample of 16 volunteers was recruited from the university and local community. Participants included eight males with mean (SD) age of 28.0 (1.4) years, stature of 170.7 (5.7) cm, and body mass of 72.3 (5.6) kg; corresponding values for the eight females were 27.5 (5.4) years, 164.0 (5.4) cm, and 56.1 (2.6) kg, respectively. Participants reported no current or recent (past 12 months) musculoskeletal injuries or disorders. The research reported herein 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 all participants prior to data collection.

2.2. Overview of experimental design and procedures

Detailed description of experimental design and procedures are reported in Park et al. (2022b). Briefly, a repeated-measures experiment was completed in which each participant performed multiple balance recovery trials in each of four *BSE* conditions: 1) NoBSE (no BSE, control condition); 2) BSE_{OFF} (backX with no supportive torque); 3) BSE_{LOW} (backX with low supportive torque); and 4) BSE_{HIGH} (backX with high supportive torque). The BSE used was the backX^{TM} model AC (US Bionics Inc., USA) – a passive BSE designed to reduce physical demands on the back during activities requiring forward bending of the trunk (Fig. A1 in the Appendix). The presentation order of *BSE* conditions was counterbalanced across participants using multiple 4 × 4 Latin Squares.

Forward loss of balance was induced using a tether-release protocol in which participants were released from a static forward leaning posture with an initial lean angle of 20°. They were asked to recover their balance by taking a single step with their dominant foot immediately following release (Fig. A2 in the appendix). Successful recoveries were followed by subsequent trials at increasing lean angles in increments of 2.5°. In the case of a failed recovery, a second trial was conducted at the same lean angle. This process was repeated until participants failed to recover their balance during two consecutive trials at the same lean angle (Madigan and Lloyd, 2005; Thelen et al., 1997).

During the tether-release protocol, participants first stood with each foot on a separate force platform (AMTI, Watertown, MA, USA) and were asked to lean forward using only dorsiflexion at the ankles. One end of a support rope was attached to the back of a belt worn by the participant, while the other end was held in a releasable clasp affixed to a rigid structure. Tension in the rope was measured using an in-line loadcell (Interface Force Measurement Solutions, Scottsdale, AZ, USA). Participants were released by the experimenter by manually opening the clasp. Successful recoveries involved the stepping foot landing on another force platform (Bertec Corporation, Columbus, OH, USA) placed directly in front of the participant.

2.3. Instrumentation and outcome measures

Kinematics of the body segments (bilateral foot, shank, and thigh as well as the pelvis and trunk) were sampled at 120 Hz using reflective markers (n=33), four rigid marker clusters (Park et al., 2022a), and a 13-camera optical motion capture system (Qualisys, Inc., Gothenburg, Sweden), and then low-pass filtered (6 Hz cutoff; 4^{th}-order Butterworth; bidirectional). Triaxial ground reaction forces from the three force platforms and uniaxial force from the in-line loadcell were sampled at 1200 Hz, subsequently low-pass filtered (10 Hz cutoff; 4^{th}-order Butterworth; bi-directional), and down-sampled to 120 Hz. Data processing was performed using custom code in MATLAB (R2021, The MathWorks Inc., Natick, MA, USA).

Temporal events – specifically tether release onset, toe-off and heel-strike of the stepping foot – were detected from the in-line loadcell and force platforms (King et al., 2005). Further analysis was performed for the initiation (i.e., from release onset to toe-off of the stepping foot) and swing (i.e., from toe-off to heel-strike of the stepping foot) phases (Fig. 1). Three-dimensional rotations of each body 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 estimated using a three-point central difference equation.

[**Fig. 1** Here]

Sagittal plane hip, knee, and ankle kinetics were assessed from the stepping leg. The body was modeled as a three-dimensional system of four rigid segments connected by frictionless pin joints, which included the foot, shank, thigh, and a head/arms/trunk (HAT). The mass and inertial properties of these segments were defined using scaling factors provided by Dumas et al. (2007). Joint torques were calculated using a conventional bottom-up inverse dynamic method to solve the governing Newton-Euler equations as described by Winter (2009). Calculated hip torques ($T_{\text{HIP-TOTAL}}$) included the sum of the torque generated by participant and BSE. Hence, hip torque generated by the participant only ($T_{\text{HIP-HUMAN}}$) was calculated by subtracting the torque generated by the device ($T_{\text{HIP-BSE}}$) from $T_{\text{HIP-TOTAL}}$. $T_{\text{HIP-BSE}}$ values for any given hip angle and angular velocity were estimated using a two-dimensional interpolation technique that was applied to the backX torque profiles reported in Madinei et al. (2022). Joint power was computed as the product of joint torque and joint angular velocity. Joint work was calculated by integrating power with respect to time. The individual contribution of each joint was determined as the ratio of its work to the total leg work (i.e., sum of work at the hip, knee, and ankle). Joint torques, power, and work were normalized to participant body mass.

2.4. Statistical analysis

Separate two-way, mixed-factor analyses of variance (ANOVAs) were performed on peak joint torques, peak joint power, joint work, and joint contribution to leg work obtained during recovery from the maximum lean angle at which participants could recover their balance, with *BSE* as a within-subject factor and sex (*SEX*) as a between-subjects factor. Presentation order of *BSE* (*ORD*) was included as a blocking factor. The effect of maximum lean angle was explored initially but excluded in the final ANOVA models, as this effect was not significant for any outcome measure. Significant effects were followed by *post hoc* pairwise comparisons using Tukey's HSD tests, and significant interaction effects were examined using simple effects analyses. Parametric model assumptions were verified using Shapiro-Wilk tests. Partial eta-squared (η_p^2) was used to quantify effect sizes for main/interaction effects. All statistical analyses were completed using JMP® Pro (v. 15.0, SAS Institute Inc., Cary, NC) with statistical significance concluded when $p<0.05$.

3. Results

Joint torques and powers during the initiation and swing phases of single-step recovery followed relatively consistent patterns in all participants (Fig. 2). Joint torques included five major phases: two at the hip (H_T1–H_T2), two at the knee (K_T1–K_T2), and one at the ankle (A_T1). Joint power consisted of seven major phases: two at the hip (H_P1–H_P2), three at the knee (K_P1–K_P3), and two at the ankle (A_P1–A_P2).

[**Fig. 2** Here]

We observed a significant decrease in peak H_T1 flexion (human + BSE) torque and peak H_P1 generation power (human + BSE) and hip positive work (human + BSE) with BSE use, supporting our first hypothesis. As for our second hypothesis, peak A_T1 plantarflexion torque and peak A_P2 generation power did not significantly increase with BSE use. However, ankle positive work increased with BSE use. For females, the relative contribution of ankle joint to total positive work increased significantly with BSE use. Thus, our second hypothesis, suggesting changes in hip joint torque would cause compensatory changes in ankle joint, was supported, but only for females. Summaries of ANOVA results, mean (SD) of outcome measures, and summary of *post hoc* pairwise comparisons are included in the appendix (Tables A1-9).

3.1. Joint torque, power, and work

Peak H_T1 flexion torque exerted by the participant plus backX (human + BSE) was lower in BSE_{LOW} and BSE_{HIGH} than in the control NoBSE condition (Fig. 3 (a)). Similarly, both peak H_P1 generation (human + BSE) power and hip positive (human + BSE) work were lower in BSE_{OFF}, BSE_{LOW}, and BSE_{HIGH} than in NoBSE (Fig. 4 (a) and Fig. 5 (a)). Interestingly, the peak H_T1 flexion torque exerted by the participant (human only) remained consistent across all *BSE* conditions (Fig. 3 (b)). However, peak H_P1 generation (human only) power and hip positive (human only) work were found to be lower in BSE_{LOW} and BSE_{OFF}, respectively, when compared to control (Fig. 4 (b) and Fig. 5(b)). For extension, peak H_T2 extension (human + BSE and human only) torques were lower in BSE_{LOW} and BSE_{HIGH} than in NoBSE (Fig. 3 (c) and (d)). A similar pattern was observed for hip power and work. When the BSE was worn, peak H_P2 absorption (human + BSE and human only) power as well as hip negative (human + BSE and human only) work were lower compared to NoBSE (Fig. 4 (c) and (d) and Fig. 5 (c) and (d)).

Significant *BSE*×*SEX* interaction effects were found on peak knee flexion torque, power, and positive work. Specifically among males, peak K_T1 flexion torque, K_P1 generation power, and knee positive work was higher in BSE_{OFF} than in BSE_{HIGH}, whereas females did not present significant differences between *BSE* conditions (Fig. 3 (e), Fig. 4 (e), and Fig. 5 (e)). There were no significant interaction effects on knee torque, power, and work associated with knee extension. Peak K_T2 extension torque, K_p2 absorption power, K_P3 generation power, and knee negative work were lower in BSE_{LOW} and BSE_{HIGH} compared to the NoBSE condition (Fig. 3 (f), Fig. 4 (f) and (g), and Fig. 5 (f)). Ankle torque and power showed no significant differences across *BSE* conditions. However, ankle positive work was significantly different across *BSE* conditions, which exhibited an increasing trend with greater BSE assistance (*post hoc* comparisons did not show statistical significance).

[**Fig. 3** Here]

[**Fig. 4** Here]

[**Fig. 5** Here]

3.2. Joint contributions to leg work

Leg positive and negative work were lower in BSE_{OFF}, BSE_{LOW}, and BSE_{HIGH}, compared to NoBSE (Fig. 6 (a) and (b)). The contribution of the hip joint (human + BSE) to leg positive work was lower in BSE_{OFF}, BSE_{LOW}, and BSE_{HIGH} compared to NoBSE (Fig. 6 (a)). Conversely, knee contribution to leg positive work was higher in BSE_{OFF} than NoBSE (Fig. 6 (a)). Ankle contribution to leg positive work was higher in BSE_{LOW} and BSE_{HIGH}, compared to NoBSE (Fig. 6 (a)). These changes, however, were evident only among females; males showed no significant changes in ankle contribution to leg positive work with BSE use. Hip (human + BSE) contribution to leg negative work was lower in BSE_{HIGH} compared to NoBSE (Fig. 6 (b)). Knee and ankle contributions to leg negative work did not differ between *BSE* conditions.

[**Fig. 6** Here]

4. Discussion

Both the BSE_{LOW} and BSE_{HIGH} conditions caused a 19-21% reduction in peak H_T1 flexion (human + BSE) torque, compared to the NoBSE condition (Fig. 3 (a)). The hip joint of the stepping leg was flexed by ~70° during the transition from the second half of the initiation phase to the first half of the swing phase (i.e., the time when hip flexion torques were needed for successful balance recovery; Fig. 2). During this time, the exoskeleton provided ~15 Nm and ~20 Nm of external hip extension torques at the hip of the stepping leg in the BSE_{LOW} and BSE_{HIGH} conditions, respectively (Fig. 7). While the peak H_T1 flexion (human only) torque remained consistent across *BSE* conditions (Fig. 3 (b)), the contribution of the hip joint (human only) to the net positive leg work increased by ~20% from the control to BSE_{HIGH} condition (Fig. 6 (a)). This increased hip flexion work performed by the participants, despite no change in peak torque, seems to have been achieved by maintaining flexion for a longer duration, as indicated in Fig. 2. Nevertheless, there was a net decrease of ~30% in peak H1 generation power and a ~40% decrease in hip positive work in the BSE_{HIGH} condition (when considering human + BSE) (Fig. 4 (a) and Fig. 5 (a)). These findings (i.e., increased hip joint work despite no change in hip joint torque) indicate that participants might have been exerting their maximum torque and were thus unable to sufficiently compensate for the external hip extension torque provided by the BSE. Alternatively, participants might have not adapted to BSE use, necessitating additional time to learn to efficiently integrate the BSE's assistance or resistance into natural movement patterns.

[**Fig. 7** Here]

Interestingly, both the BSE_{LOW} and BSE_{HIGH} conditions led to a decrease in peak H_T2 extension (human + BSE) torque, by 17-20% compared to the NoBSE condition (Fig. 3 (c)). This change occurred even though the BSE provided substantial external hip extension torque. The hip joint of the stepping leg was flexed ~70-85° during the second half of the swing phase (i.e., the time during which hip extension torque was needed for successful balance recovery; Fig. 2). During this time, the exoskeleton provided ~16 Nm and ~22 Nm of external hip extension torques in the BSE_{LOW} and BSE_{HIGH} conditions, respectively (Fig. 7). However, participants reduced their exertion much more when compared to the NoBSE condition; specifically, there was a 35-48% decrease in peak H_T2 extension (human only) torque in the BSE_{LOW} and BSE_{HIGH} conditions (Fig. 3 (d)). Note that participants generated hip flexion torques for a longer duration in the BSE_{HIGH} (vs. NoBSE) condition (Fig. 2, increased from 68 (SD 16) to 99 (SD 23) ms from NoBSE to BSE_{HIGH}). Such changes indicate that participants may have tried to increase hip flexion torque when external hip extension torques were provided, and this in turn may have resulted in decreased hip extension torque (i.e., reciprocal inhibition; Hamill and Knutzen, 2006).Subsequently, a substantial reduction occurred in peak H_P2 absorption (human + BSE and human only) power and hip negative (human + BSE and human only) work during BSE use (Fig. 4 (c) and (d) and Fig. 5 (c) and (d)).

Both the BSE_{LOW} and BSE_{HIGH} conditions decreased peak K_T2 extension torque, by 10-14% compared to the NoBSE condition (Fig. 3 (f)). Note that when peak K_T2 extension torque occurred (i.e., at around the middle of the swing phase), participants typically generated hip extension torques in the NoBSE condition. However, they were still exerting hip flexion torques at this time in the BSE_{HIGH} condition (Fig. 2). While the primary hip extensor is the bicep femoris, generating simultaneous knee extension and hip flexion torque requires activation of the rectus femoris (Mansfield and Neumann, 2018). Hence, generating knee extension torque during the swing phase of balance recovery in the BSE_{LOW} and BSE_{HIGH} conditions may have been particularly challenging, since the rectus femoris is less efficient at producing knee extension torque when hip extension torque also needs to be generated (Miyamoto et al., 2012). Peak K_P2 absorption power and K_P3 generation power occurred immediately prior to and after peak K_T2 extension torque (Fig. 2). Decreased knee extension torques may have been the reason for decreased peak K_P2 absorption power, peak K_P3 generation power, and knee negative work in BSE_{LOW} and BSE_{HIGH} conditions (Fig. 4 (f) and (g) and Fig. 5 (f)).

Given that hip flexors and knee extensors are the predominant sources of kinetic energy for forward progression of the stepping leg during balance recovery (Madigan, 2006), our finding of a decrease in hip and knee joint kinetics indicate that BSE use can have adverse effects on balance recovery, and can further explain the impaired balance recovery kinematics (i.e., decrease in hip/knee ranges of motion and hip/knee angular velocities) observed in our earlier study (Park et al., 2022b). Overall, BSE use led to a 14-18% decrease in leg positive (human + BSE) work (Fig. 6 (a)) and a 24-33% decrease in leg negative (human + BSE) work (Fig. 6 (b)). While the hip contribution from the human to leg positive work increased with BSE use, it was not successful given that the net hip contribution (human + BSE) work still decreased with BSE use. This decreased net hip contribution was associated with an increased ankle contribution in BSE_{LOW} and BSE_{HIGH} among females (Fig. 6 (a)). Similar changes have been observed in older adults. Compared to young adults, older adults exhibit substantial decrease in peak H_T1 flexion torque, H_T2 extension torque, H_P1 generation power, and K_P3 generation power during the initiation and swing phases of single-step recovery (Madigan, 2006; Wojcik et al., 2001). However, they also exhibit compensatory changes such as increased A_P2 generation power, which can help moving the shank superiorly and anteriorly (Madigan, 2006). Given that females generally have lower hip flexion strength compared to males (Harbo et al., 2012), the relative impact of the external hip extension torque provided by the BSE may have been greater for females, potentially leading to compensatory changes (i.e., increased ankle contribution to leg positive work). Further studies are needed to understand whether this observed sex difference in BSE effects on lower limb kinetics is caused primarily by differences in hip flexion strength or if there are other contributing factors.

The current analysis shares the same limitations in our earlier report (Park et al., 2022b), such as individual differences in size and strength contributing to different fit of the BSE and different relative effects of BSE-applied torques on the body joints. Although we estimated the torque generated by the backX, relative motion and misalignment between the device and human anatomy may have affected the effective torques transferred to the humans. Furthermore, we focused only on the short-term effects of one specific BSE device use among a young and healthy group of individuals. The protocol used in our experiment may differ from real trip recovery scenarios, considering the anticipation of a potential fall, regulation of the initial leaning posture, and the restriction to using only a single-step. Finally, given that peak H_T1 flexion (human only) torque, peak K_T1 flexion torque, and leg positive (human + BSE) work were lower in the last *BSE* condition than in the first *BSE* condition (i.e., significant effect of *ORD*), repetitive recovery trials may have caused muscle fatigue in the hip and knee flexors. While participants were provided 5-minute break between different *BSE* conditions, longer and more frequent breaks may have been needed to prevent muscle fatigue.

In summary, peak H_T1 flexion (human only) torque remained unchanged during recovery from maximum lean angle when wearing the BSE (i.e., backX) with supportive torque, and individuals exerted hip flexion torques for longer durations during the initiation and swing phases of single-step recovery following a forward loss of balance. These changes resulted in unchanged power, but a slightly increased hip contribution to net positive leg work. Nonetheless, these adjustments proved insufficient to resist the extension torque applied by the BSE, resulting in a net decrease in total positive leg work and hip joint (human + BSE) contribution to leg work. While both males and females exhibited these changes in hip joint kinetics with BSE use, a significant compensatory increase in ankle contribution was observed among females only. These findings indicate that wearing a BSE had adverse effects during the initiation and swing phases of balance recovery by decreasing the kinetic energy for reactive stepping.

Declaration of Competing Interest

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

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Figure Captions

Fig. 1. Representative body positions during the initiation, swing, and landing phase of a successful single-step recovery.

Fig. 2. Means (line) and standard deviations (shaded area) of sagittal plane torques and powers for the hip (human + BSE), hip (human only), knee, and ankle during the initiation and swing phases of single-step recovery following a forward loss of balance. Data are shown separately for the NoBSE (blue) and BSE_{HIGH} (red) conditions. Illustrated data for each *BSE* condition were obtained during recovery from the maximum lean angle at which participants *(n*=16) could recover their balance. Although not a primary focus here, joint angles and angular velocities are also illustrated to aid in interpretation.

Fig. 3. Peak torques at the hip and knee in the four *BSE* conditions. Data are presented as group means, with error bars signifying standard deviations. Upper case letters specify groupings obtained from pairwise comparisons between *BSE* conditions. Pairs of conditions not sharing the same letters are significantly different (e.g., A is significantly different from BC; AB is not significantly different from B). The downward pointing arrows and accompanying numbers indicate a significant decrease (and the corresponding percentage change in the measure) from the NoBSE condition.

Fig. 4. Peak power at the hip and knee in the four *BSE* conditions. Data are presented as group means, with error bars signifying standard deviations. Upper case letters specify groupings obtained from pairwise comparisons between *BSE* conditions. Lower case letters specify groupings obtained from pairwise comparisons between *BSE* conditions within *GEN*. Means of pairs of conditions not sharing the same letters are significantly different (e.g., A is significantly different from C; b is not significantly different from ab). The downward pointing arrows and accompanying numbers indicate a significant decrease (and the corresponding percentage change in the measure) from the NoBSE condition.

Fig. 5. Work at the hip, knee, and ankle in the four *BSE* conditions. Data are presented as group means, with error bars signifying standard deviations. Upper case letters specify groupings obtained from pairwise comparisons between *BSE* conditions. Lower case letters specify groupings obtained from pairwise comparisons between *BSE* conditions within *GEN*. Means of pairs of conditions not sharing the same letters are significantly different (e.g., A is significantly different from B; a is not significantly different from ab). The downward pointing arrows and accompanying numbers indicate a significant decrease (and the corresponding percentage change in the measure) from the NoBSE condition.

Fig. 6. Relative contribution of hip (human + BSE), knee, and ankle to (a) positive and (b) negative work in the four *BSE* conditions. The area of each pie chart is proportional to total positive or negative work done by the stepping leg (i.e., sum of work at the hip, knee, and ankle). The symbol * indicates a significant change from the NoBSE condition.

Fig. 7. Mean (line) and standard deviations (shaded area) of external hip extension torques generated by BSE_{OFF} (black), BSE_{LOW} (blue), and BSE_{HIGH} (red) during the initiation and swing phases of single-step recovery following a forward loss of balance. Illustrated data for each *BSE* condition were obtained during recovery from maximum lean angle at which participants (*n*=16) could recover their balance.

Fig. 1.

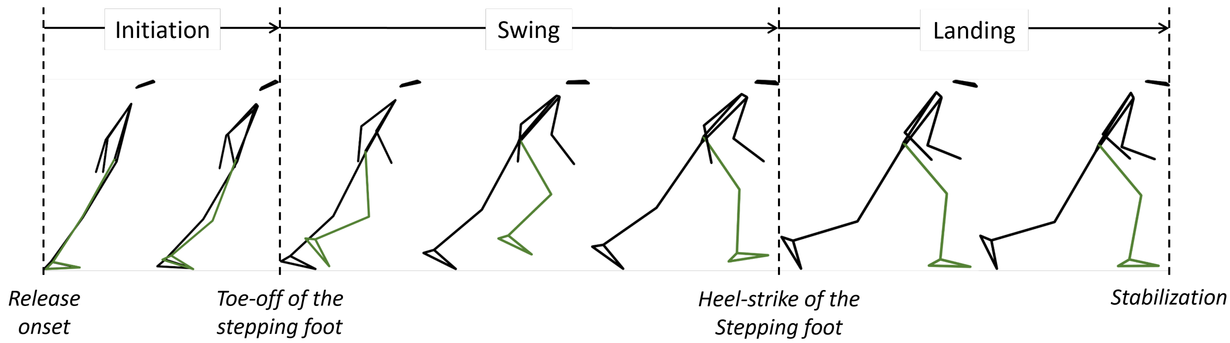

Initiation
Swing
Landing
Release onset
Toe-off of the stepping foot
Heel-strike of the Stepping foot
Stabilization

Fig. 2.

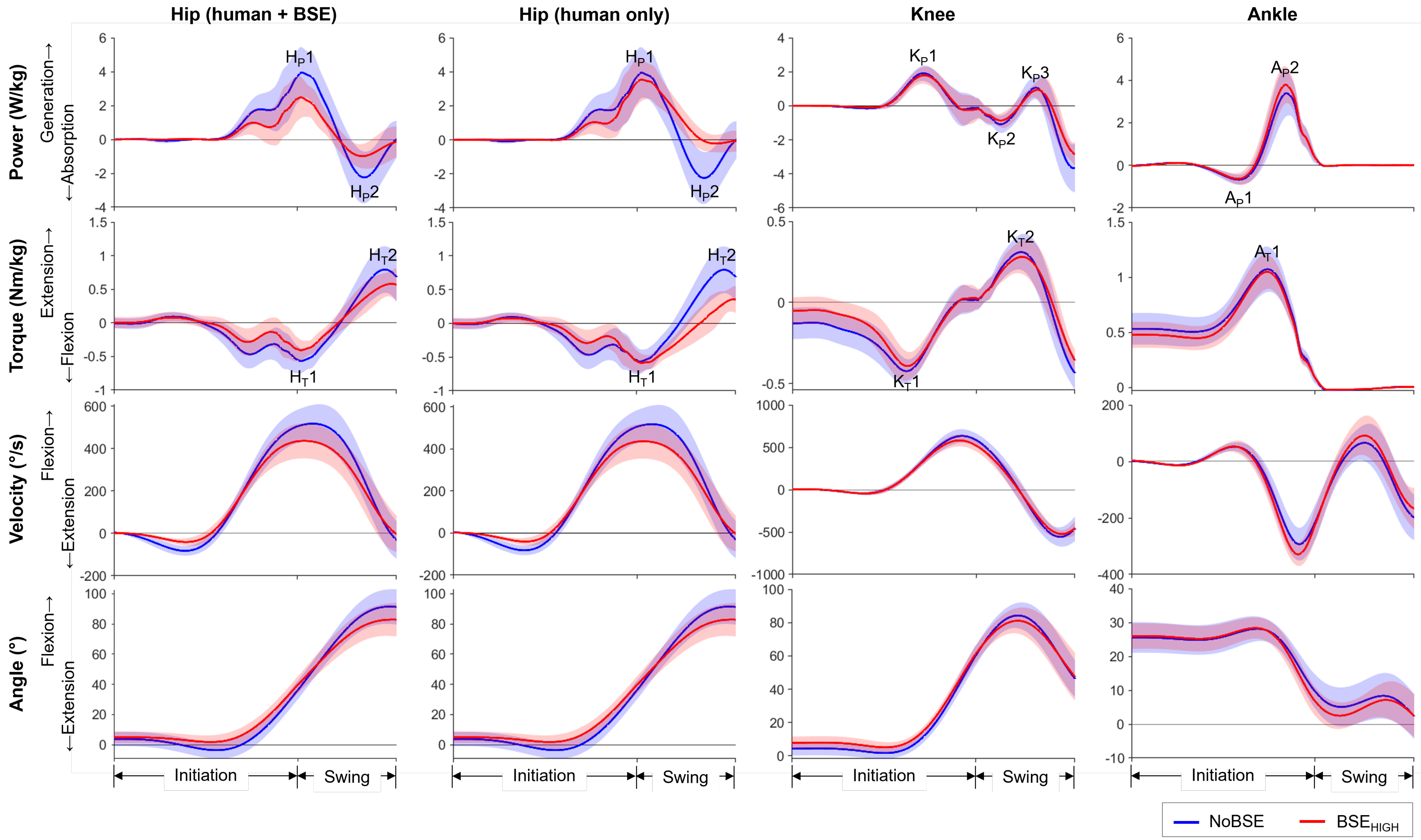
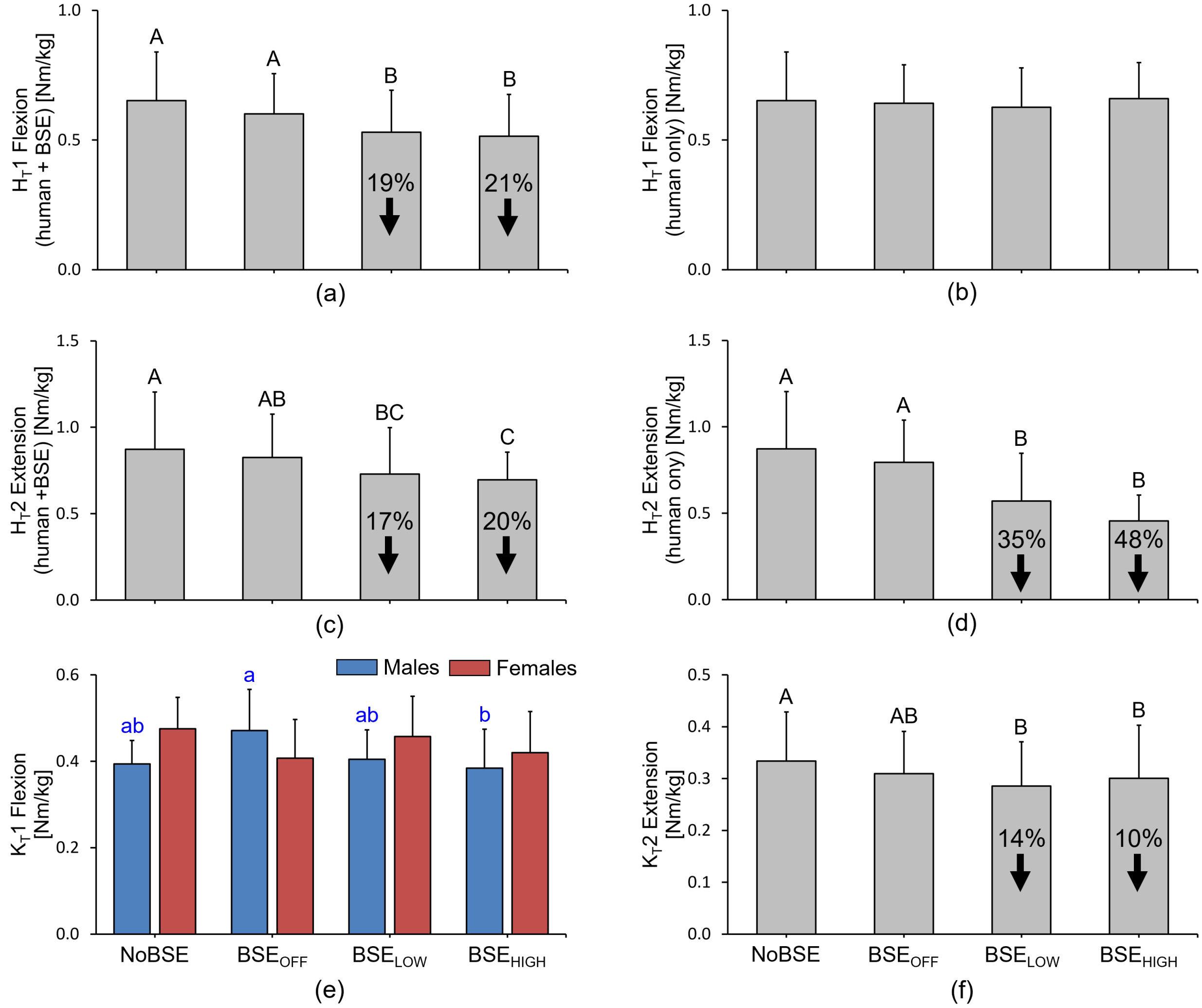
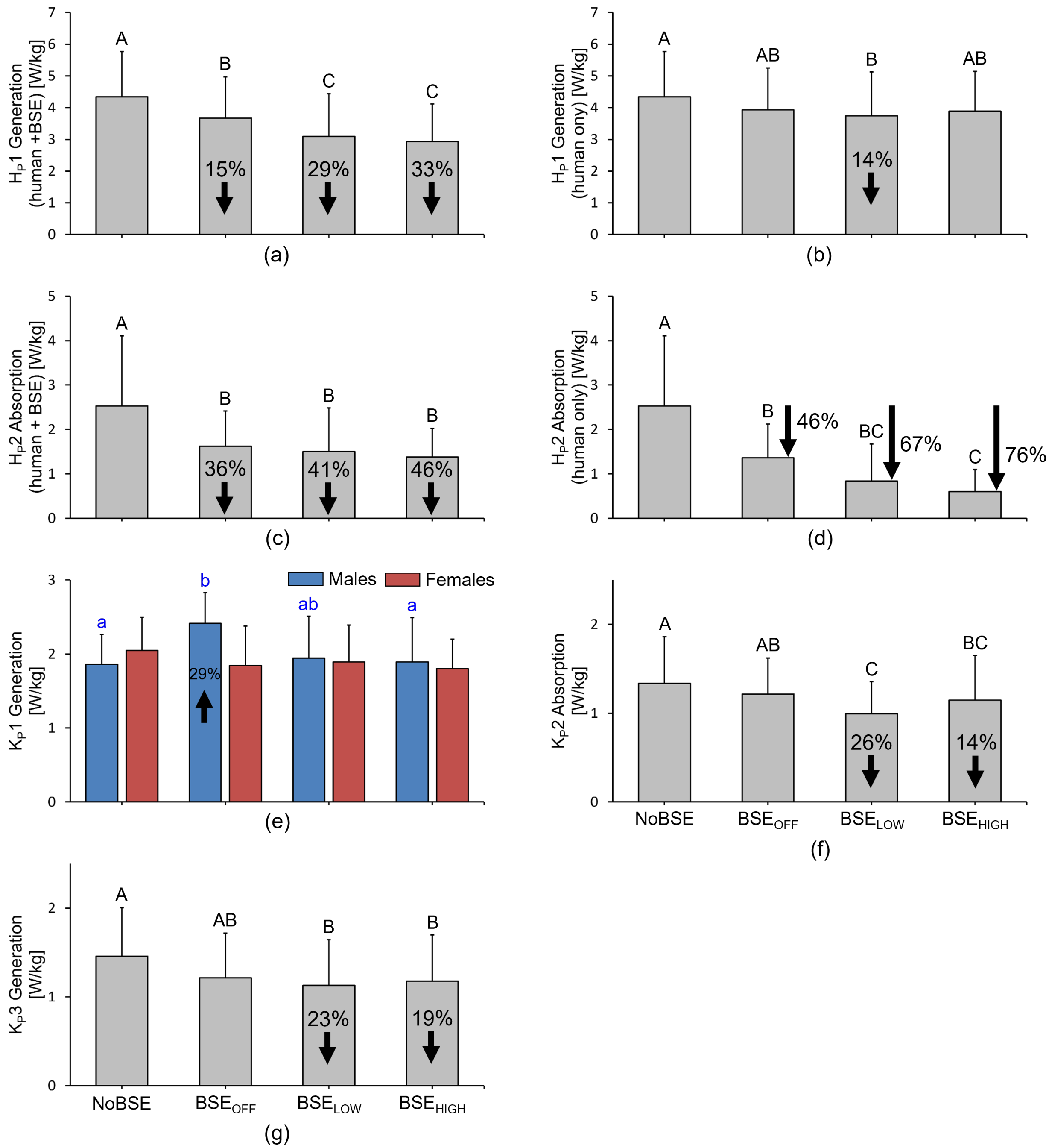

Hip (human + BSE)
Hip (human only)
Knee
Ankle
Power (W/kg)
←Absorption Generation→
Torque (Nm/kg)
←Flexion Extension→
Velocity (°/s)
←Extension Flexion→
Angle (°)
←Extension Flexion→
HP1
HP2
HT1
HT2
KP1
KP2
KP3
KT1
KT2
AP1
AP2
AT1
Initiation
Swing
NoBSE
BSEHIGH

Fig. 3.



1.0
0.5
0.0
H_T1 Flexion (human + BSE) [Nm/kg]
A
A
B
B
19%
21%
(a)
H_T1 Flexion (human only) [Nm/kg]
(b)
1.5
H_T2 Extension (human +BSE) [Nm/kg]
A
AB
BC
C
17%
20%
(c)
H_T2 Extension (human ony) [Nm/kg]
A
A
B
B
35%
48%
(d)
0.6
0.4
0.2
K_T1 Flexion [Nm/kg]
Males
Females
ab
a
ab
b
NoBSE
BSE_OFF
BSE_LOW
BSE_HIGH
(e)
0.3
0.1
K_T2 Extension [Nm/kg]
A
AB
B
B
14%
10%
(f)

Fig. 4.



H_P1 Generation (human +BSE) [W/kg]
A
B
C
C
15%
29%
33%
(a)
H_P1 Generation (human ony) [W/kg]
A
AB
B
AB
14%
(b)
H_P2 Absorption (human + BSE) [W/kg]
A
B
B
B
36%
41%
46%
(c)
H_P2 Absorption (human only) [W/kg]
A
B
46%
BC
67%
C
76%
(d)
Males
Females
K_P1 Generation [W/kg]
a
b
ab
a
29%
(e)
K_P2 Absorption [W/kg]
A
AB
C
BC
26%
14%
NoBSE
BSE_{OFF}
BSE_{LOW}
BSE_{HIGH}
(f)
K_P3 Generation [W/kg]
A
AB
B
B
23%
19%
NoBSE
BSE_{OFF}
BSE_{LOW}
BSE_{HIGH}
(g)

Fig. 5.

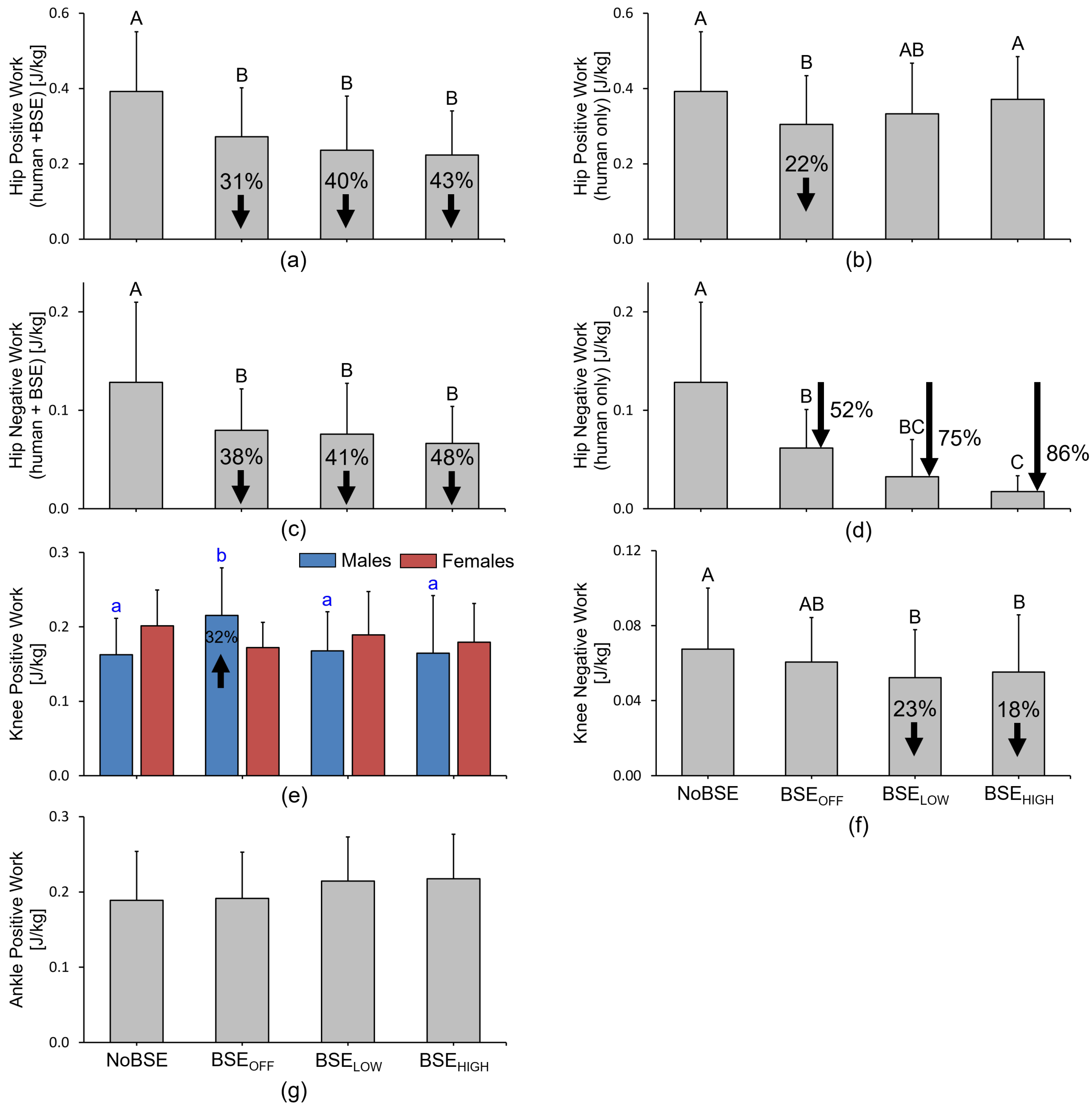
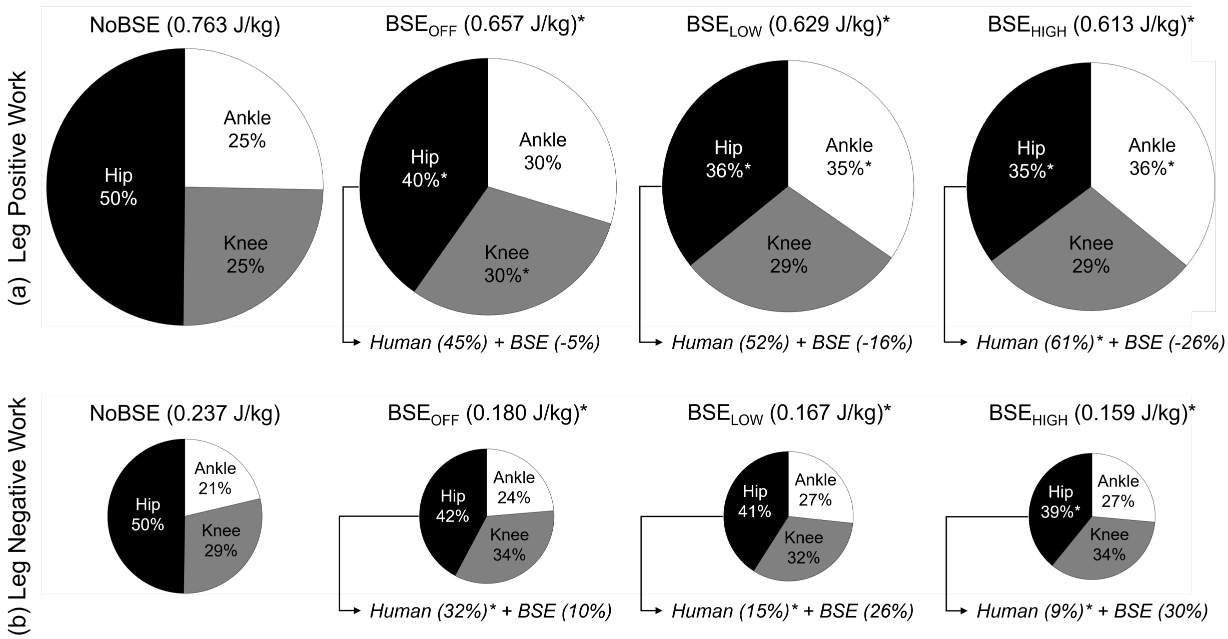
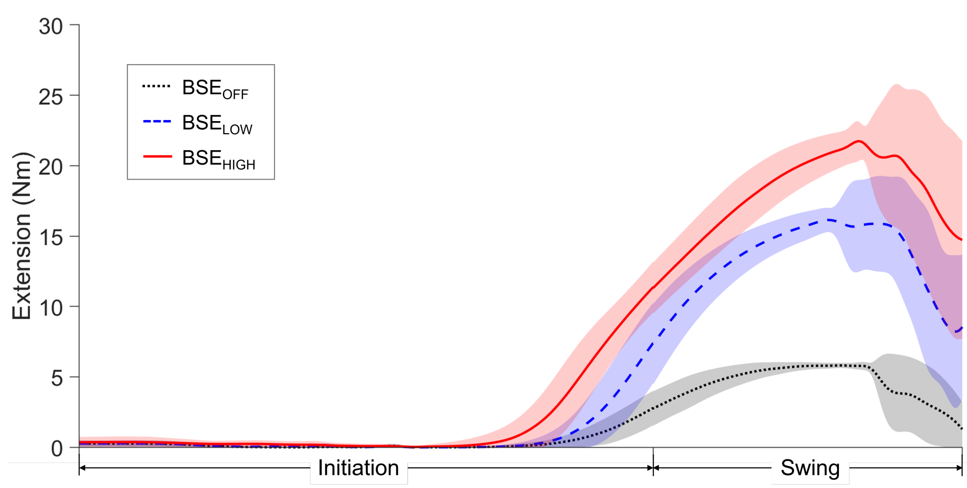

Hip Positive Work (human +BSE) [J/kg]
A
B
B
B
31%
40%
43%
(a)
Hip Positive Work (human only) [J/kg]
A
B
AB
A
22%
(b)
Hip Negative Work (human + BSE) [J/kg]
A
B
B
B
38%
41%
48%
(c)
Hip Negative Work (human only) [J/kg]
A
B
52%
BC
75%
C
86%
(d)
Knee Positive Work [J/kg]
Males
Females
a
b
a
a
32%
(e)
Knee Negative Work [J/kg]
A
AB
B
B
23%
18%
NoBSE
BSE OFF
BSE LOW
BSE HIGH
(f)
Ankle Positive Work [J/kg]
NoBSE
BSE OFF
BSE LOW
BSE HIGH
(g)

Fig. 6.



(a) Leg Positive Work
NoBSE (0.763 J/kg)
Hip 50%
Ankle 25%
Knee 25%
BSE$_{OFF}$ (0.657 J/kg)*
Hip 40%*
Ankle 30%
Knee 30%*
Human (45%) + BSE (-5%)
BSE$_{LOW}$ (0.629 J/kg)*
Hip 36%*
Ankle 35%*
Knee 29%
Human (52%) + BSE (-16%)
BSE$_{HIGH}$ (0.613 J/kg)*
Hip 35%*
Ankle 36%*
Knee 29%
Human (61%)* + BSE (-26%)
(b) Leg Negative Work
NoBSE (0.237 J/kg)
Hip 50%
Ankle 21%
Knee 29%
BSE$_{OFF}$ (0.180 J/kg)*
Hip 42%
Ankle 24%
Knee 34%
Human (32%)* + BSE (10%)
BSE$_{LOW}$ (0.167 J/kg)*
Hip 41%
Ankle 27%
Knee 32%
Human (15%)* + BSE (26%)
BSE$_{HIGH}$ (0.159 J/kg)*
Hip 39%*
Ankle 27%
Knee 34%
Human (9%)* + BSE (30%)

Fig. 7.



BSE$_{OFF}$
BSE$_{LOW}$
BSE$_{HIGH}$
Extension (Nm)
30
25
20
15
10
5
0
Initiation
Swing