**TITLE PAGE**

**Full title:**

Wearing a back-support exoskeleton impairs single-step balance recovery performance following a forward loss of balance – An exploratory study

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**Abstract**

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

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

## 1. Introduction

There is growing interest in back-support exoskeletons (BSEs) as a potential workplace intervention to control the risk of overexertion injuries associated with manual material handling (Kermavnar et al., 2021; Nussbaum et al., 2019). At present, passive BSEs that require no external energy input (vs. active that do) are relatively more mature for occupational use due to their lower cost and ease of implementation in the workplace (Nussbaum et al., 2019). BSEs include rigid structures and physical attachments surrounding or supporting the chest, waist, and/or thighs. Among marketed passive BSEs, masses range from 2.8 to 4.5 kg (Exoskeleton Report LLC, 2021; Voilqué et al., 2019), and extension torques of ~35 Nm can be produced at each hip (Koopman et al., 2019; Madinei, 2021; Näf et al., 2018). BSE characteristics, such as their rigid structure, added mass, and the external hip extension torque, may place increased demands on the postural control system. In fact, we have shown that BSE use decreases postural balance in bipedal stance (Park et al., 2021) and decreases gait performance and stability during level walking (Park et al., 2022). Based on this prior evidence, and the rapid emergence of this technology as a potential intervention in industrial settings, we believe that it is important to quantify the effects of BSE use on fall risk.

Most falls that occur while walking are due to external perturbations such as trips and slips (Berg et al., 1997; Heijnen and Rietdyk, 2016) that cause a loss of balance. In occupational settings, risk factors for such trips and slips include slippery or uneven surfaces, poor lighting, and unstable footwear (Afanuh et al., 2012). Hence, the ability to respond effectively to a loss of balance (i.e., reactive stepping) is a critical factor in preventing falls, by expanding and re-establishing the base-of-support (BoS) (Aftab et al., 2016). Results from prior studies suggest that wearing a BSE could increase the likelihood of unsuccessful balance recovery following postural perturbations. For example, increased constraints on the hip and trunk from wearing a corset inhibited the ability of an individual to effectively control their trunk following a forward loss of balance (Pretty et al., 2019). Similarly, studies on age-related differences in balance recovery have

shown that reduced recovery ability with aging is associated with increased weight transfer time (Wojcik et al., 1999), decreased step velocity (Wojcik et al., 1999), decreased lower limb joint torques (Madigan and Lloyd, 2005a; Wojcik et al., 2001), and decreased joint powers (Madigan, 2006). The added mass of a BSE might similarly increase weight transfer time. In particular, the external hip-extension torque provided by the device may decrease the lower limb flexion joint torques and powers needed to execute a successful balance recovery, while both the mass and the torque of the device could decrease step velocity during single-step balance recovery. However, to the best of our knowledge, the effects of a BSE on reactive balance recovery have not yet been reported.

One method to assess balance recovery ability is the tether-release protocol, which involves releasing an individual from a static, forward-leaning posture and having them attempt to recover their balance with a single step (Do et al., 1999; Thelen et al., 1997). Balance recovery responses can be quantified using spatiotemporal characteristics of stepping, joint kinematics and kinetics, and ground reaction forces (e.g., Carty et al., 2011; Ringhof et al., 2019). Rapid perturbations are induced in this protocol, hence the direction, amplitude, and speed of compensatory reactions need to be modulated in real time to account for the unpredictable body motion (Maki et al., 2003). Slower and shorter steps are typically associated with unsuccessful recovery due to failure in increasing the BoS area timely and adequately (Lee et al., 2014; Wojcik et al., 1999). Wojcik et al. (2001) reported that, as the lean angle increased, larger hip and knee ranges of motion (RoMs) were used in successful recoveries (Wojcik et al., 2001). Furthermore, older men exhibited smaller peak flexion velocities at the hip, knee, and ankle (plantarflexion), compared to young men (Madigan and Lloyd, 2005b). The additional external hip extension torque generated by a BSE may further decrease balance recovery in reaction to a perturbation, by impairing the hip flexion needed to generate a rapid stepping response.

To support a better understanding of whether BSE use increases fall risk, we performed an exploratory study to investigate the effects of wearing a BSE on balance recovery following a forward loss

of balance. A tether-release protocol was utilized to simulate a forward fall. Based on the characteristics of a BSE, we hypothesized that wearing a BSE would adversely affect balance recovery following a forward balance perturbation. Balance recovery ability was assessed directly using the maximum lean angle from which individuals could execute successful recovery using a single step in each experimental condition. We also obtained several recovery response metrics associated with recovery performance, including spatiotemporal measures (reaction time and step characteristics), lower limb joint kinematics (RoMs and peak flexion/extension velocities during recovery), and the margin of stability (MoS) for body center-of-mass (CoM). Since the additional external hip extension torque generated by a BSE may decrease balance recovery, by impairing the hip flexion needed to generate a rapid stepping response, we expected BSE use to cause shorter and slower steps, decreased hip RoM and peak angular velocity, and reduced MoS. We further expected to find an inverse relationship between BSE torque and recovery responses, with greater external hip extension torque generated by the BSE leading to a greater deterioration in balance recovery kinematics.

## 2. Methods

### 2. 1. Participants

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

## 2. 2. Exoskeleton selection and conditions

The backX™ model AC (US Bionics Inc., USA), designed to reduce physical demands on the back during forward bending, was used in this study (Fig. A1). This exoskeleton has mass = 4.5 kg and consists of a waist strap, chest support, and an external torque generator about each hip that is coupled with the waist strap and the chest support. Four exoskeleton (*EXO*) conditions were included to represent expected use cases: 1) NoExo (no exoskeleton, control condition); 2) $EXO_{OFF}$ (backX with no supportive torque); 3) $EXO_{LOW}$ (backX with low supportive torque [instant mode]); and 4) $EXO_{HIGH}$ (backX with high supportive torque [instant mode]). Torques applied by the backX in these different settings were measured in another study (Madinei et al., 2022), which involved an isokinetic hip flexion/extension test protocol in a dynamometer, using the exoskeleton mounted on a mannequin. Peak hip extension torque generated with $EXO_{LOW}$ and $EXO_{HIGH}$ are ~18 Nm and ~27 Nm, respectively.

## 2. 3. Experimental design and procedures

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

Forward loss of balance was induced by releasing participants from a static forward leaning posture (initial lean angle = 20°). Participants were asked to recover their balance using a single step of the dominant foot (i.e., the foot used to kick a ball) immediately following release. Successful recoveries were followed by repeated trials at increasing lean angles, in increments of 2.5°. Failed recoveries were followed by a second trial at the same lean angle. This process was repeated until participants failed to recover their

140 balance in two consecutive trials at the same lean angle (Madigan and Lloyd, 2005b; Thelen et al., 1997).
141 To minimize learning effects, the experimental protocol was begun after a practice session of 10-15 trials
142 of single step recovery.
143 Participants first stood with feet about shoulder width apart, with each foot on a force platform
144 (OR6-7-1000, AMTI, Watertown, MA, USA) and with their arms along the sides of the body. They were
145 then asked to lean forward using only ankle dorsiflexion with their feet flat on the floor (Fig. 1 (a)). One
146 end of a support rope was attached to the back of a belt worn by the participant, while the other end was
147 held in a releasable clasp affixed to a rigid structure. Rope tension was measured using an in-line loadcell
148 (Interface Force Measurement Solutions, Scottsdale, AZ, USA). Lean angle was controlled by adjusting
149 the length of the rope, and the resultant lean angle relative to upright standing posture was measured using
150 an inertial measurement unit (Mtw Awinda, Xsens Technologies B.V., the Netherlands) attached over the
151 L1 spinous process. To minimize any anticipatory movements prior to release, vertical ground-reaction
152 forces were visually confirmed to be roughly equal bilaterally.
153 At each lean angle, participants were released by the experimenter at a random time interval (2-10
154 s) by manually opening the clasp. To prevent falls to the floor, a full-body safety harness was worn and
155 attached with a lanyard to a sliding track along the ceiling (Fig. 1). Another loadcell (Cooper Instruments
156 and Systems, Warrenton, VA, USA), mounted in-line with the lanyard, was used to measure the force
157 applied to the harness. Any of three criteria were used to define a failed balance recovery (Madigan and
158 Lloyd, 2005b; Thelen et al., 1997): 1) when more than one step was taken with the dominant foot; 2) when
159 a force >30% body weight was applied to the lanyard at any point during recovery; and 3) when the trailing
160 foot took a step longer than 30% of individual stature. Another force platform (9090-15, Bertec Corporation,
161 Columbus, OH, USA) was placed directly in front of the participant such that in the event of a successful
162 recovery the leading foot would land on it (Fig. 1 (b)).
163

164 **2. 4. Instrumentation and outcome measures**

165 Whole body kinematics, ground reaction forces, and loadcell forces were collected during balance
166 recovery trials. Reflective markers (*n*=33) and four rigid marker clusters (Park et al., 2022) were sampled
167 at 120 Hz using a 13-camera optical motion capture system (Qualisys, Inc., Gothenburg, Sweden), then
168 low-pass filtered (6 Hz cutoff; $4^{th}$-order Butterworth; bidirectional) (Nevisipour and Honeycutt, 2020).
169 Triaxial ground reaction forces from the three force platforms and uniaxial force from two loadcells were
170 sampled at 1200 Hz, subsequently low-pass filtered (10 Hz cutoff; $4^{th}$-order Butterworth; bi-directional),
171 and down-sampled to 120 Hz.

172 Body segment masses, segment CoM locations, and joint centers were determined using the
173 methods suggested by Dumas et al. (2007). Joint angles were calculated as Cardan angles between adjacent
174 local segments with an order of rotation of flexion-extension, abduction-adduction, and internal-external
175 rotation. Whole-body CoM was calculated using a 13-segment model (bilateral foot, shank, thigh, upper
176 arm, and forearm, as well as the pelvis, trunk, and head).

177 Balance recovery ability was defined as the maximum lean angle from which participants could
178 recover their balance with a single step (Hsiao-Wecksler, 2008). Balance recovery responses that were
179 quantified during single-step balance recovery included spatiotemporal measures, lower limb joint RoMs
180 and peak angular velocities, and MoS measures.

181 *Spatiotemporal measures.* Temporal events were first extracted during a single-step recovery from
182 the in-line loadcell, force platforms, and whole body CoM data (King et al., 2005; Wojcik et al., 1999), as
183 illustrated in Fig. 2. Temporal measures included reaction time (release onset to the start of pushoff; Fig. 2)
184 and step time (liftoff to landing). Step length was the distance in the anteroposterior (AP) direction of the
185 calcaneus marker on the stepping foot between liftoff and landing.

186 *Joint RoM and peak angular velocities.* Lower limb joint RoMs and peak angular velocities were
187 obtained from sagittal plane kinematics of the stepping leg during the time interval between release onset

and balance recovery. Outcome measures included RoM at the hip, knee, and ankle, and peak velocities in flexion and extension at the hip, knee, and ankle.

*Margin of Stability (MoS).* Extrapolated whole-body CoM (XCoM) was obtained using the equation described in Hof et al. (2005). MoS was defined as the minimum distance between the XCoM and BoS borders in the AP direction (Hof et al., 2005). The BoS was defined as the boundary made by markers attached on the 2$^{nd}$ distal phalanx of the stepping foot, calcaneus of the non-stepping foot, 1$^{st}$ metatarsal head of the non-stepping foot, and 5$^{th}$ metatarsal head of the stepping foot. MoS measures analyzed here were $MoS_{LANDING}$ (i.e., MoS at the instant of landing) and $MoS_{RECOVERY}$ (i.e., MoS at the instant of balance recovery).

**2. 5. Statistical analysis**

A two-way, mixed-factor analysis of variance (ANOVA) was performed on maximum lean angle, with *EXO* as a within-subject factor and gender (*GEN*) as between-subjects factor. Separate three-way, mixed-factor ANOVAs were performed for each balance recovery response, with *EXO* and lean angle (*LEAN*) as within-subject factors and *GEN* as a between-subjects factor. Lean angle included four levels (Lev1: 20°, Lev2: 22.5°, Lev3: 25°, Lev4: maximum lean angle). Parametric model assumptions were assessed using Shapiro-Wilk tests, and data transformations were performed to meet these assumptions as relevant (Table 1). Significant effects were followed by *post hoc* pairwise comparisons using Tukey's HSD tests, and significant interaction effects were further examined using simple effects analyses. Given the study goals, the subsequent presentation of results and the discussion emphasizes the main and interaction effects of *EXO*. If interactions were ordinal, (i.e., no crossovers 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. Partial eta-squared ($\eta_p^2$) was used to quantify effect sizes for significant main/interaction effects. 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 (v. 15.0, SAS Institute Inc., Cary, NC), using the restricted maximum likelihood (REML) method and with statistical significance concluded when $p < 0.05$.

## 3. Results

Mean (SD) of maximum lean angles was 30.9° (4.8°), and these angles were not significantly affected by *EXO* ($p$=0.991), gender ($p$=0.712), or their interaction ($p$=0.135). Substantial inter-individual differences were evident: compared to the control condition, maximum lean angle in the $EXO_{HIGH}$ condition was lower among 5/16 individuals, the same among 7/16 individuals, and higher among 4/16 participants. In contrast, there were significant main effects of *EXO* on most balance recovery responses (Table 1), and these results are presented in more detail below for each group of outcomes. Tables A1 and A2 in the appendix provide descriptive summaries of each balance recovery response and a summary of *post hoc* comparisons for the significant main effects of *EXO*, respectively.

### 3. 1. Spatiotemporal measures

There were significant main effects of *EXO* on all spatiotemporal measures (Table 1, Fig. 3). Reaction time was higher in the three conditions when wearing the BSE compared to the NoExo condition. Step time was lower in both the $EXO_{OFF}$ and $EXO_{LOW}$ conditions than in the NoExo condition. Furthermore, step length was lower in the three conditions when wearing the BSE than in the NoExo condition.

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

Fig. 4 presents sample joint angles and angular velocities during a single-step recovery. There were no significant main or interaction effects of *EXO* on ankle RoM. In contrast, both hip and knee RoMs were

lower in the three conditions when wearing the BSE compared to the NoExo condition (Table 1 and Fig. 5).

There were significant main effects of *EXO* on peak velocities in flexion and extension at both the hip and knee (Table 1). While peak hip flexion velocity was lower in the three BSE conditions than the NoExo condition, peak hip extension velocity was lower in the $EXO_{HIGH}$ condition compared to the NoExo condition (Fig. 6). There was a significant interaction effect of *EXO* and *LEAN* on peak hip extension velocity, which was lower in the $EXO_{HIGH}$ compared to other conditions only with a 20° lean angle. Peak knee flexion velocity was lower in the $EXO_{LOW}$ and $EXO_{HIGH}$ conditions compared to NoExo, and it was also lower in $EXO_{HIGH}$ than in $EXO_{OFF}$ (Fig. 6). Peak knee extension velocity was lower only in the $EXO_{OFF}$ condition compared to NoExo (Fig. 6).

### 3. 3. Margin of stability measures

There were significant main effects of *EXO* on both $MoS_{LANDING}$ and $MoS_{RECOVERY}$ (Table 1). Both $MoS_{LANDING}$ and $MoS_{RECOVERY}$ were lower in $EXO_{HIGH}$ compared to the NoExo and $EXO_{OFF}$ conditions (Fig. 7).

## 4. Discussion

### 4. 1. Balance recovery ability

The maximum lean angle from which individuals could successfully recover balance did not differ between *EXO* conditions. Rather, we observed large inter-individual differences, which may have been related to individual variability in lower limb strength, agility, and/or motor learning ability. We further examined the effects of *EXO* on the magnitude of overshoot of exerted force as a percent increase from the baseline (Fig. 2 (a)), measured from the right force platform immediately following pushoff. However, no

259 significant main effect of *EXO* was found ($p$=0.091), implying that participants did not exert more muscular
260 effort to recover balance in conditions in which they wore the exoskeleton. Maximum lean angle, or our
261 methodology to quantify this measure, may not be sufficiently sensitive to differentiate between the effects
262 of the various *EXO* conditions included, especially given that there were *EXO* effects on balance recovery
263 responses.
264
265 **4. 2. Balance recovery responses**
266 BSE use decreased peak hip flexion velocity by 5-13% (Fig. 6). Changes observed in the $EXO_{OFF}$
267 condition may have been due to constrained movement and/or added mass when wearing the BSE (the
268 device produced no supportive torque in this condition), whereas further decreases in the $EXO_{HIGH}$
269 condition likely resulted from the external hip extension torque provided by the device. Wearing the BSE
270 also led to an 8-11% decrease in hip RoM (Fig. 5), by decreasing peak hip flexion (by 6-10%). Decreases
271 in both peak hip flexion velocity and hip RoM indicate that wearing the BSE impeded hip flexion
272 kinematics during reactive stepping.
273 From Fig. 4, it can be observed that between rope release and landing (i.e., stepping), both the hip
274 and the knee were flexed. Therefore, changes in hip kinematics could have caused a corresponding decrease
275 in peak knee flexion velocity and knee RoM (Figs. 5 and 6). Decreased peak hip flexion velocity may also
276 be the primary cause of decreased step time (3-4%) and step length (8-10%) when wearing the BSE (Fig.
277 3). Note that the spatiotemporal measures did not differ between the three BSE conditions, suggesting that
278 simply wearing the BSE affected these measures, but the magnitude of external hip extension torque did
279 not.
280 Since stepping responses are associated with balance recovery performance, our results regarding
281 changes in spatiotemporal measures, joint RoMs, and peak joint velocities during stepping imply that
282 wearing the BSE may impair balance recovery. For example, Madigan and Lloyd (2005b) reported that

283 older adults had lower peak hip (26% decrease) and knee (30% decrease) flexion velocities during balance
284 recovery. Furthermore, large RoMs at the hip and knee were found as critical for successful balance
285 recovery (Wojcik et al., 1999, 2001).

286 Reaction time increased by 10-12% in the three BSE conditions (Fig. 3), which we think could have
287 one of three potential causes. First, the BSE straps and structures that surround the chest, waist, and/or thigh
288 may have altered sensory feedback, thereby interfering with proprioception (Stirling et al., 2020). Second,
289 there may have been increased postural anxiety when wearing the BSE, and such anxiety can lead to reduced
290 spinal reflex excitability (Sibley et al., 2007). Third, Do et al. (1999) noted that control of the short swing
291 phase duration in reactive stepping following a loss of balance can reduce reaction time. Hence, it can be
292 speculated that participants programmed a longer swing phase duration prior to release, to increase the BoS
293 when wearing the BSE, and this in turn increased reaction time. Whatever the underlying cause(s) may be,
294 successful recovery from unexpected balance loss when wearing a BSE in real life may not allow
295 individuals the affordance to react more slowly.

296 Both $MoS_{LANDING}$ and $MoS_{RECOVERY}$ decreased (10-13%) in the $EXO_{HIGH}$ condition (Fig. 7),
297 indicating that the high supportive torque impaired dynamic stability during balance recovery following a
298 forward loss of balance (Hof et al., 2005; Ringhof et al., 2019). XCoM is located inside of the BoS
299 immediately after landing (Fig. 8), explaining why MoS values are always positive at both landing and
300 balance recovery in this study. Fig. 8 also shows that XCoM gradually moves anteriorly after rope release,
301 then moves back (posteriorly) after landing (i.e., stepping). Hence, both decreased step length and increased
302 step time seem to be associated with decrements in MoS (Hsiao-Wecksler and Robinovitch, 2007), as the
303 former can lead to a decrease in BoS and the latter changes the direction of XCoM movement (i.e., anterior
304 to posterior direction) more slowly. These associations further explain why MoS values decreased only in
305 the $EXO_{HIGH}$ condition, since compared to the control condition step length decreased in the three conditions
306 when wearing the BSE and step time decreased only in the $EXO_{OFF}$ and $EXO_{LOW}$ conditions (Fig. 3).

### 4. 3. Limitations

This study has some limitations common with most laboratory-based research conducted using exoskeletons. Such limitations include individual differences in BSE fit/adjustments that may have influenced our findings, a focus only on the short-term effects of BSE use, and inclusion of only healthy and young individuals. Further, although we minimized any anticipatory adjustments before releasing the participants (by visually checking vertical ground reaction forces), unlike falls outside of the laboratory participants were still expecting a fall during the experimental protocol. In addition, we tested only one BSE, and it unclear whether the current results will generalize to several other BSE designs that are available.

### 4. 4. Conclusions

We evaluated the effects of wearing backX on balance recovery following a loss of balance. The maximum lean angle was not altered when wearing the BSE in any of the conditions with varying levels of supportive torque. However, wearing the BSE negatively affected balance recovery responses at all lean angles. Specifically, wearing the BSE increased reaction times and impeded hip flexion kinematics during reactive stepping, and decreased hip flexion, which in turn decreased peak knee flexion velocity, knee RoM, and step length, all suggesting impaired balance recovery. While constrained movement and/or added mass of the exoskeleton may be the primary factor that decreases balance recovery performance, margin of stability measures suggested that increasing external hip extension torque from the BSE can further impair balance recovery. Although differences in recovery responses did not lead to a resultant loss of 'recovery ability' in our young and healthy sample, future work is needed to understand if other demographic groups (e.g., older workers) may also be able to demonstrate successful balance recovery with the use of BSEs in

occupational environments. Future work extending kinetic analyses to recovery responses would offer a more complete picture of how a BSE could impact balance recovery following a loss of balance.

**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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## Tables

**Table 1.** Summary ($p$-values) of ANOVA results for the main and interaction effects (*EXO*: *EXO* condition, *LEAN*: lean angle, *GEN*: gender) on each balance recovery outcome measure. No three-way interactions (i.e., *EXO*×*LEAN*×*GEN*) were statistically significant (all $p$-values>0.3). For each effect, entries are $p$ values ($\eta_p^2$), and significant effects are highlighted in bold font ($p < 0.05$).

| Outcome measures | Data transformation performed | Main and interaction effects | | | | | |
|---|---|---|---|---|---|---|---|
| | | *EXO* | *LEAN* | *GEN* | *EXO*×*LEAN* | *EXO*×*GEN* | *LEAN*×*GEN* |
| *Spatiotemporal measures* | | | | | | | |
| Reaction time | - | **<0.001 (0.131)** | **0.035 (0.043)** | 0.177 (0.035) | 0.761 (0.028) | 0.869 (0.004) | **0.037 (0.040)** |
| Step time | Log | **0.004 (0.064)** | **<0.001 (0.186)** | 0.513 (0.058) | 0.277 (0.052) | 0.541 (0.011) | 0.672 (0.007) |
| Step length | - | **<0.001 (0.135)** | **<0.001 (0.593)** | 0.333 (0.047) | 0.953 (0.016) | 0.426 (0.014) | 0.655 (0.008) |
| *RoM* | | | | | | | |
| Hip | - | **<0.001 (0.222)** | **<0.001 (0.511)** | 0.078 (0.075) | 0.984 (0.012) | 0.288 (0.018) | 0.687 (0.008) |
| Knee | - | **<0.001 (0.161)** | **<0.001 (0.436)** | 0.700 (0.006) | 0.956 (0.016) | **0.023 (0.045)** | 0.795 (0.006) |
| Ankle | Square root | 0.294 (0.018) | **<0.001 (0.386)** | 0.826 (0.003) | 0.603 (0.036) | 0.471 (0.013) | 0.561 (0.010) |
| *Peak angular velocity* | | | | | | | |
| Hip flexion | - | **<0.001 (0.297)** | **<0.001 (0.355)** | **0.038 (0.260)** | 0.828 (0.024) | 0.212 (0.022) | 0.500 (0.012) |
| Hip extension | Log | **0.015 (0.051)** | 0.170 (0.024) | 0.172 (0.084) | **0.008 (0.103)** | 0.492 (0.012) | 0.579 (0.010) |
| Knee flexion | Square | **<0.001 (0.203)** | **<0.001 (0.351)** | **0.023 (0.296)** | 0.710 (0.030) | **0.029 (0.043)** | 0.183 (0.024) |
| Knee extension | - | **0.028 (0.044)** | **<0.001 (0.338)** | 1.000 (0.000) | 0.538 (0.038) | 0.601 (0.009) | 0.146 (0.027) |
| Ankle dorsiflexion | Square root | 0.104 (0.031) | **<0.001 (0.099)** | 0.528 (0.041) | 0.180 (0.061) | 0.451 (0.013) | 0.876 (0.003) |
| Ankle plantarflexion | Square root | 0.784 (0.005) | **<0.001 (0.257)** | 0.885 (0.002) | 0.856 (0.023) | 0.236 (0.021) | 0.983 (0.001) |
| *MoS* | | | | | | | |
| $MoS_{LANDING}$ | - | **0.023 (0.047)** | **<0.001 (0.520)** | **0.003 (0.187)** | 0.481 (0.041) | 0.171 (0.024) | 0.832 (0.004) |
| $MoS_{RECOVERY}$ | Square | **0.005 (0.063)** | **<0.001 (0.130)** | 0.128 (0.082) | 0.208 (0.058) | 0.151 (0.026) | 0.936 (0.002) |

## Figure captions

**Fig. 1.** Illustration of the experimental setup during (a) initial tethered lean and (b) balance recovery after release. Numbers indicate: ① in-line loadcell, ② lanyard loadcell, ③ left force platform, ④ right force platform, ⑤ front force platform, and ⑥ backX exoskeleton.

**Fig. 2.** Sample time histories of forces (in-line loadcell, right force platform [$F_Z$], and front force platform [$F_Z$]) and whole-body CoM displacement in the anteroposterior (AP) direction during a single-step recovery. As all participants used their right foot as the dominant (leading) foot for stepping, vertical force from the right force platform was used to detect onsets of pushoff and liftoff.

**Fig. 3.** Spatiotemporal measures in the four *EXO* conditions during a single-step recovery following a forward loss of balance. 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. 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 upward (or downward) pointing arrowheads and accompanying numbers indicate a significant increase (or decrease) from the NoExo (control) condition and the corresponding percentage change in a dependent measure.

**Fig. 4.** Sample joint angles and angular velocities during a single step recovery. Zero joint angle corresponds to the anatomical position, and positive values indicate hip flexion, knee flexion, and ankle dorsiflexion.

**Fig. 5.** Joint RoMs in the four *EXO* conditions during a single-step recovery following a forward loss of balance. 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. Means of pairs of conditions not sharing the same letters are significantly different (e.g., A is significantly different from B). The downward pointing arrowheads and accompanying numbers indicate a significant decrease from the NoExo (control) condition and the corresponding percentage change in a dependent measure.

**Fig. 6.** Peak joint angular velocities in the four *EXO* conditions during a single step-recovery following a forward loss of balance. 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. Means of pairs of conditions not sharing the same letters are significantly different (e.g., A is significantly different from B; C is not significantly different from BC). The downward pointing arrowheads and accompanying numbers indicate a significant decrease from the NoExo (control) condition and the corresponding percentage change in a dependent measure.

**Fig. 7.** MoS measures in the four *EXO* conditions during a single step-recovery following a forward loss of balance. 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. 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 arrowheads and accompanying numbers indicate a significant decrease from the NoExo (control) condition and the corresponding percentage change in a dependent measure.

**Fig. 8.** An example showing temporal changes in anteroposterior (AP) positions of the base-of-support (BoS), whole body center-of-mass (CoM), and extrapolated whole-body CoM (XCoM). Zero AP position indicates the origin of the laboratory reference frame.

**Figures**

**Fig. 1.**

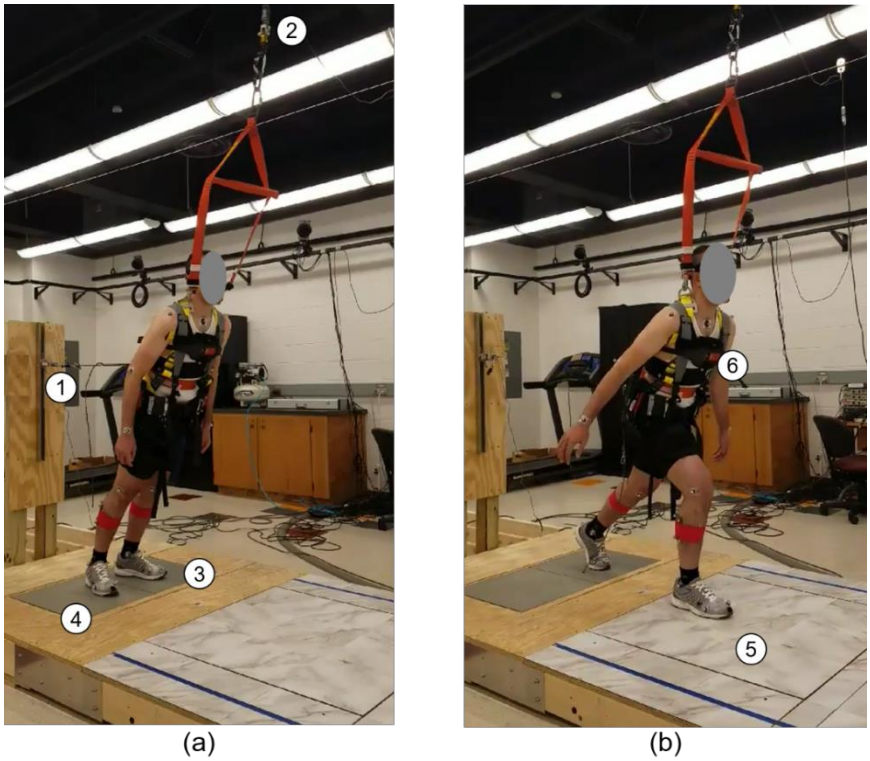



**Fig. 2.**

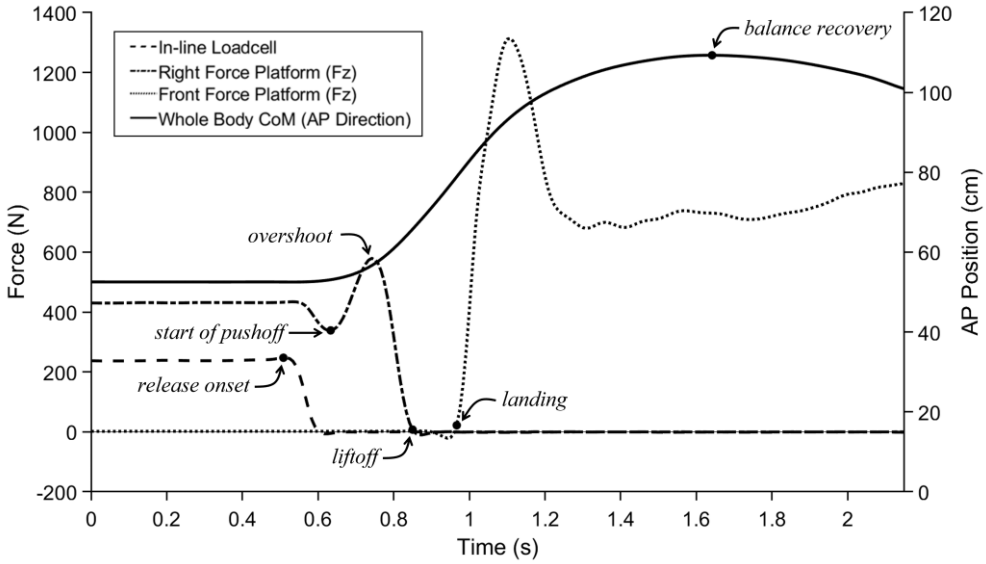


**Fig. 3.**

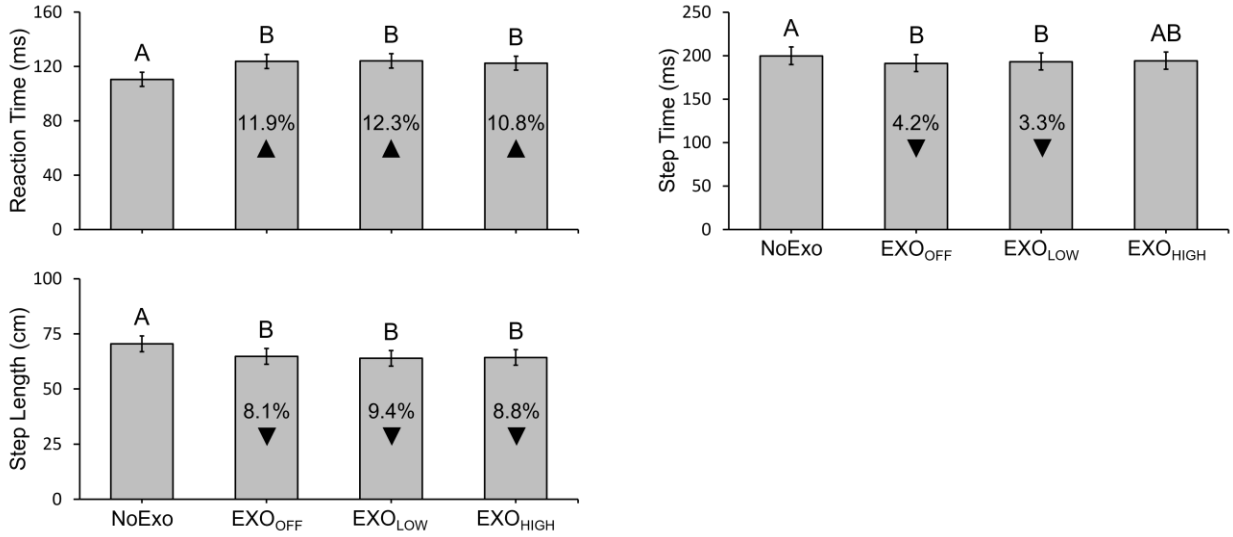

Reaction Time (ms)
160
120
80
40
0
A
B
B
B
11.9%
12.3%
10.8%
Step Time (ms)
250
200
150
100
50
0
A
B
B
AB
4.2%
3.3%
Step Length (cm)
100
75
50
25
0
A
B
B
B
8.1%
9.4%
8.8%
NoExo
EXO$_{OFF}$
EXO$_{LOW}$
EXO$_{HIGH}$

**Fig. 4.**

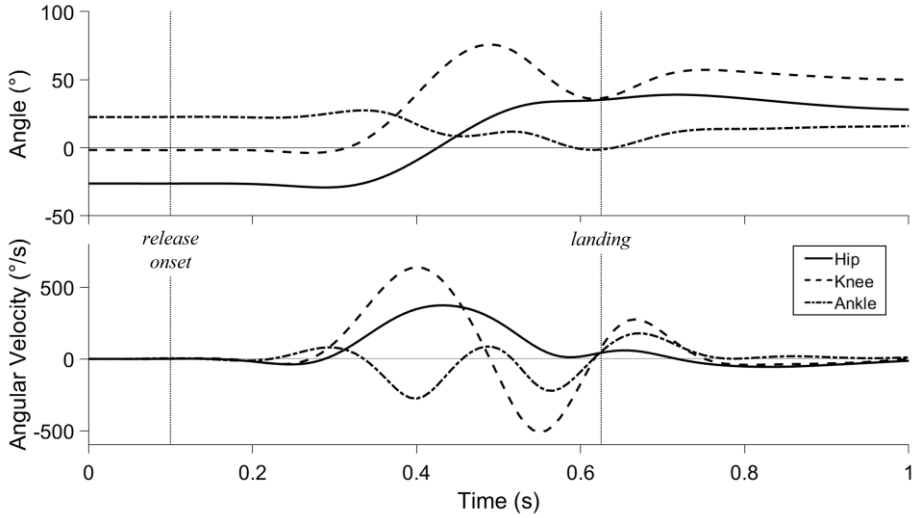

Angle (°)
100
50
0
-50
release onset
landing
Angular Velocity (°/s)
500
0
-500
Hip
Knee
Ankle
0
0.2
0.4
0.6
0.8
1
Time (s)

**Fig. 5.**

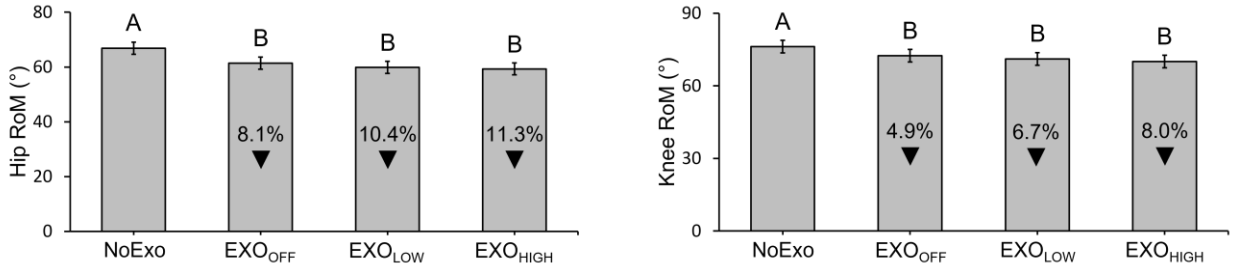



**Fig. 6.**

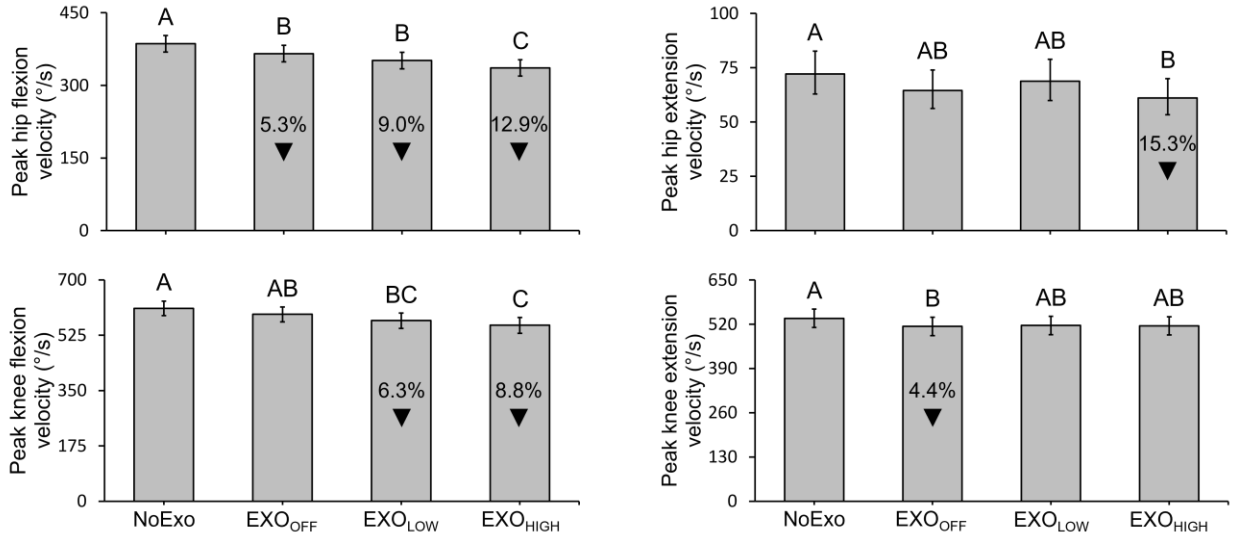



**Fig. 7.**

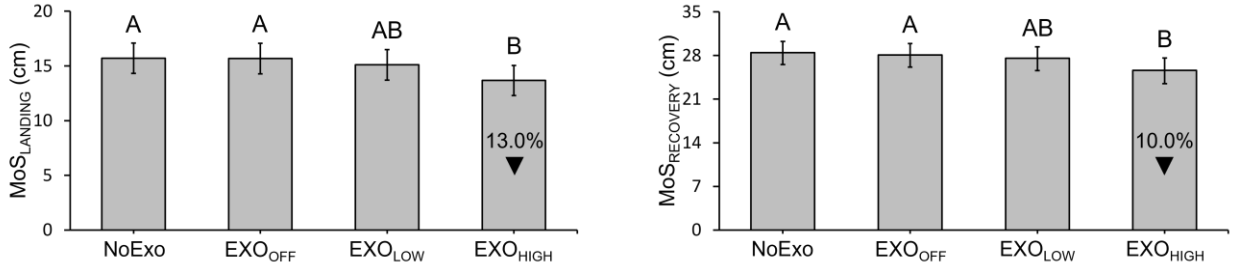
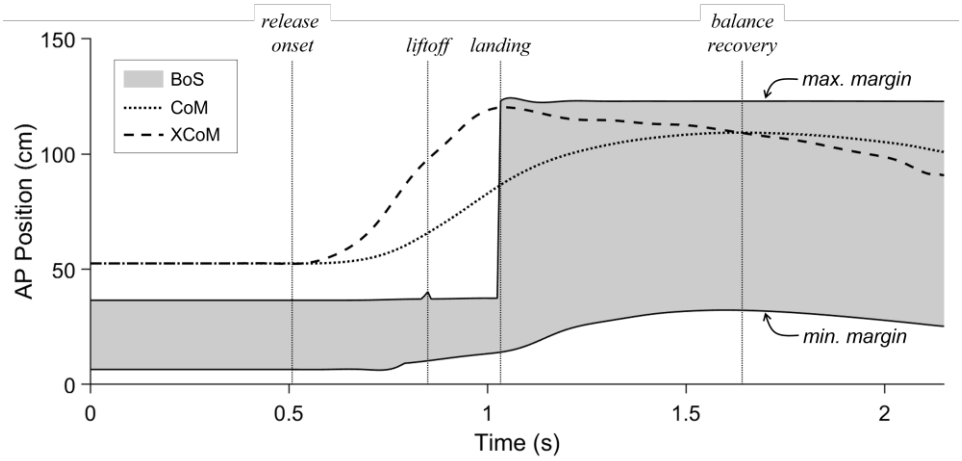


**Fig. 8.**


release onset
liftoff
landing
balance recovery
BoS
CoM
XCoM
max. margin
min. margin
AP Position (cm)
Time (s)
150
100
50
0
0
0.5
1
1.5
2

## Appendices

**Table A1.** Least squares mean (95% confidence intervals) of each outcome measure for balance recovery responses in 20° (i.e., minimum lean angle) and maximum lean angle.

| Outcome measures | Lean angle = 20° | | | | Lean angle = maximum | | | |
|---|---|---|---|---|---|---|---|---|
| | NoExo | $EXO_{OFF}$ | $EXO_{LOW}$ | $EXO_{HIGH}$ | NoExo | $EXO_{OFF}$ | $EXO_{LOW}$ | $EXO_{HIGH}$ |
| *Spatiotemporal measures* | | | | | | | | |
| Reaction time (ms) | 104.7 (90.6, 118.8) | 126.6 (112.5, 140.6) | 125.5 (111.5, 139.6) | 117.2 (103.1, 131.3) | 118.2 (111.8, 124.6) | 125.5 (119.1, 131.9) | 127.1 (120.7, 133.5) | 127.1 (120.7, 133.5) |
| Step time (ms) | 201.5 (188.7, 215.1) | 189.4 (177.4, 202.2) | 183.5 (171.9, 195.9) | 186.6 (174.8, 199.3) | 207.5 (198.5, 217.1) | 199.1 (190.4, 208.2) | 205.2 (196.2, 214.6) | 204.6 (195.6, 213.9) |
| Step length (cm) | 62.0 (57.1, 67.0) | 55.4 (50.5, 60.4) | 53.0 (48.0, 57.9) | 54.1 (49.1, 59.1) | 82.9 (78.9, 86.9) | 77.5 (73.5, 81.6) | 77.0 (73.0, 81.0) | 75.2 (71.2, 79.2) |
| *RoM* | | | | | | | | |
| Hip (°) | 60.1 (56.4, 63.8) | 55.8 (52.1, 59.5) | 52.6 (48.9, 56.2) | 52.0 (48.3, 55.7) | 75.9 (73.0, 78.7) | 69.8 (66.9, 72.6) | 68.7 (65.9, 71.6) | 67.2 (64.3, 70.1) |
| Knee (°) | 69.4 (65.5, 73.3) | 68.4 (64.4, 72.3) | 65.2 (61.3, 69.2) | 64.2 (60.3, 68.2) | 83.6 (80.0, 87.1) | 79.4 (75.8, 82.9) | 77.9 (74.3, 81.4) | 76.6 (73.0, 80.1) |
| Ankle (°) | 28.2 (25.3, 31.2) | 25.7 (23.0, 28.6) | 26.5 (23.7, 29.4) | 25.5 (22.7, 28.3) | 36.2 (34.2, 38.4) | 35.3 (33.2, 37.4) | 34.9 (32.8, 37.0) | 34.8 (32.7, 37.0) |
| *Peak angular velocity* | | | | | | | | |
| Hip flexion (°/s) | 346.3 (319.9, 372.7) | 333.3 (306.9, 359.8) | 328.7 (302.2, 355.1) | 303.9 (277.4, 330.3) | 413.9 (395.7, 432.2) | 393.5 (375.3, 411.8) | 373.6 (355.3, 391.8) | 356.2 (337.9, 374.4) |
| Hip extension (°/s) | 72.0 (59.5, 87.1) | 66.6 (55.1, 80.7) | 73.1 (60.4, 88.5) | 46.9 (38.7, 56.7) | 76.8 (64.6, 91.3) | 67.3 (56.6, 79.9) | 69.7 (58.7, 82.8) | 72.0 (60.6, 85.5) |
| Knee flexion (°/s) | 549.6 (516.1, 581.3) | 546.5 (512.8, 578.3) | 535.2 (500.6, 567.6) | 514.5 (478.4, 548.1) | 643.4 (612.8, 672.5) | 631.4 (600.3, 661.1) | 600.9 (568.1, 632.0) | 588.2 (554.7, 620.0) |
| Knee extension (°/s) | 487.2 (451.2, 523.2) | 466.3 (430.3, 502.3) | 470.9 (434.9, 506.9) | 464.9 (428.9, 500.9) | 581.3 (545.7, 616.9) | 564.5 (528.9, 600.1) | 551.6 (516.1, 587.2) | 538.3 (502.7, 573.9) |
| Ankle dorsiflexion (°/s) | 160.0 (141.6, 179.5) | 147.7 (130.0, 166.4) | 151.2 (133.3, 170.2) | 147.2 (129.6, 166.0) | 157.6 (139.3, 177.1) | 173.1 (153.9, 193.6) | 175.5 (155.8, 196.3) | 179.9 (159.9, 201.0) |
| Ankle plantarflexion (°/s) | 267.4 (236.6, 300.2) | 259.9 (229.5, 292.2) | 268.2 (237.4, 301.0) | 261.3 (230.8, 293.7) | 307.5 (284.8, 331.0) | 320.0 (296.9, 344.0) | 331.7 (307.6, 356.7) | 333.6 (309.5, 358.7) |
| *MoS* | | | | | | | | |
| $MoS_{LANDING}$ (cm) | 22.3 (20.0, 24.6) | 20.3 (18.0, 22.7) | 19.4 (17.1, 21.8) | 17.6 (15.3, 19.9) | 9.5 (7.1, 12.0) | 11.1 (8.7, 13.6) | 9.2 (6.7, 11.7) | 7.2 (4.7, 9.6) |
| $MoS_{RECOVERY}$ (cm) | 31.4 (28.9, 33.8) | 29.8 (27.2, 32.3) | 29.2 (26.5, 31.7) | 26.1 (23.0, 28.9) | 25.9 (22.9, 28.5) | 26.4 (23.5, 29.0) | 24.9 (21.8, 27.6) | 21.7 (18.1, 24.8) |

**Table A2.** Summary of *post hoc* pairwise comparisons (Tukey's HSD tests) for the significant main effects of exoskeleton on each outcome measure for balance recovery responses. Table entries are *p* values, and significant differences are highlighted in bold ($p < 0.05$).

| Outcome measures | *EXO* condition | | | | | |
|---|---|---|---|---|---|---|
| | NoExo vs. $EXO_{OFF}$ | NoExo vs. $EXO_{LOW}$ | NoExo vs. $EXO_{HIGH}$ | $EXO_{OFF}$ vs. $EXO_{LOW}$ | $EXO_{OFF}$ vs. $EXO_{HIGH}$ | $EXO_{LOW}$ vs. $EXO_{HIGH}$ |
| *Spatiotemporal measures* | | | | | | |
| Reaction time | **<0.001** | **<0.001** | **<0.001** | 0.999 | 0.970 | 0.938 |
| Step time | **0.003** | **0.035** | 0.095 | 0.854 | 0.605 | 0.974 |
| Step length | **<0.001** | **<0.001** | **<0.001** | 0.916 | 0.983 | 0.992 |
| *RoM* | | | | | | |
| Hip | **<0.001** | **<0.001** | **<0.001** | 0.506 | 0.234 | 0.962 |
| Knee | **0.003** | **<0.001** | **<0.001** | 0.559 | 0.107 | 0.775 |
| *Peak angular velocity* | | | | | | |
| Hip flexion | **0.002** | **<0.001** | **<0.001** | 0.068 | **<0.001** | **0.036** |
| Hip extension | 0.175 | 0.819 | **0.013** | 0.650 | 0.754 | 0.136 |
| Knee flexion | 0.072 | **<0.001** | **<0.001** | 0.074 | **<0.001** | 0.280 |
| Knee extension | **0.046** | 0.097 | 0.065 | 0.991 | 0.999 | 0.999 |
| *MoS* | | | | | | |
| $MoS_{LANDING}$ | 1.000 | 0.849 | **0.035** | 0.872 | **0.041** | 0.233 |
| $MoS_{RECOVERY}$ | 0.966 | 0.685 | **0.005** | 0.920 | **0.023** | 0.117 |

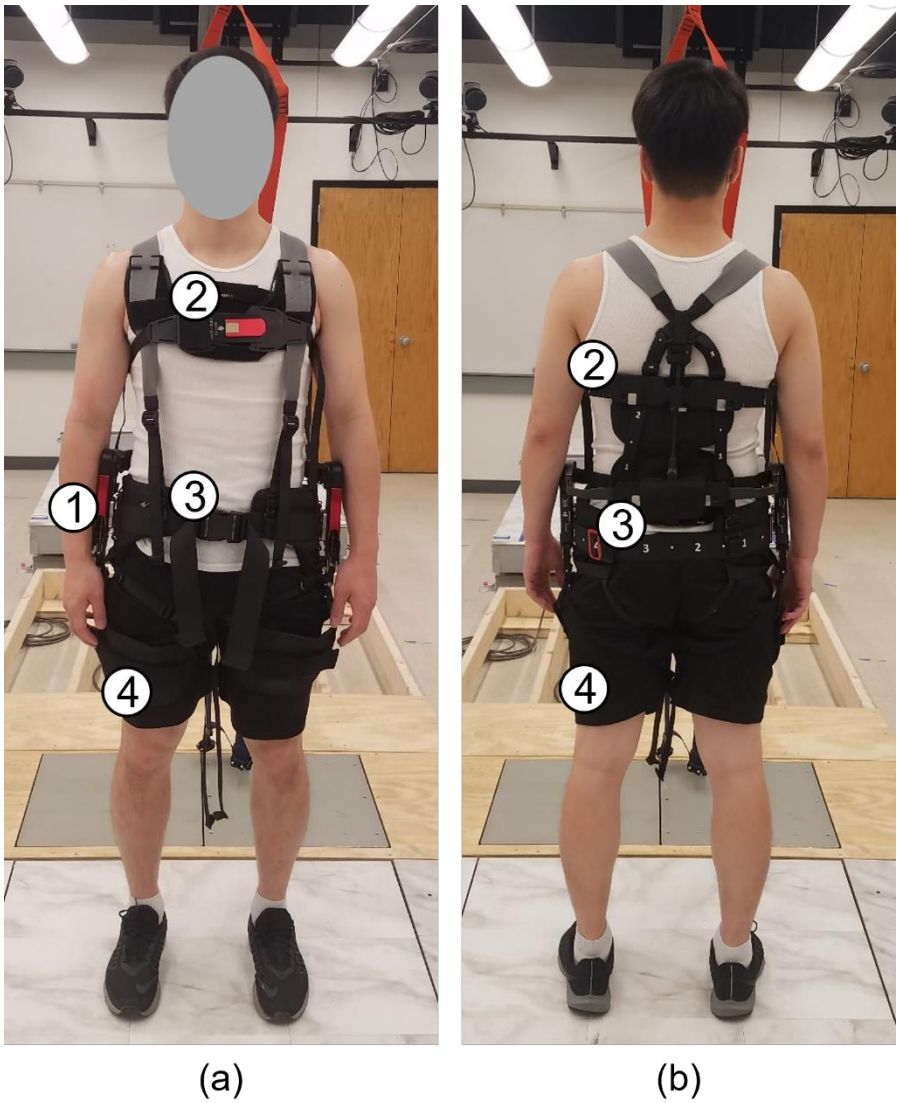



**Fig. A1.** The backX (www.suitx.com) used in the study. Numbers indicate: ① torque generator, ② chest pad and strap, ③ hip belt, ④ thigh strap.