The Effects of Exoskeleton Assistance at the Ankle on Sensory Integration During Standing Balance

Santiago Canete, Elizabeth B. Wilson, W. Geoffrey Wright, and Daniel A. Jacobs, *Member, IEEE*

***Abstract*— Exoskeleton devices can reduce metabolic cost, increase walking speed, and augment load-carrying capacity. However, little is known about the effects of powered assistance on the sensory information required to achieve these tasks. To learn how to use an assistive device, humans must integrate novel sensory information into their internal model. This process may be disrupted by challenges to the sensory systems used for posture. We investigated the exoskeleton-induced changes to balance performance and sensory integration during quiet standing. We asked 11 unimpaired adults to perform a virtual reality-based test of sensory integration in balance (VRSIB) on two days while wearing the exoskeleton either unpowered, using proportional myoelectric control, or with regular shoes. We measured postural biomechanics, muscle activity, equilibrium scores, postural control strategy, and sensory ratios. Results showed improvement in balance performance when wearing the exoskeleton on firm ground. The opposite occurred when standing on an unstable platform with eyes closed or when the visual information was non-veridical. The balance performance was equivalent when the exoskeleton was powered versus unpowered in all conditions except when both the support surface and the visual information were altered. We argue that in stable ground conditions, the passive stiffness of the device dominates the postural task. In contrast, when the ground becomes unstable the passive stiffness negatively affects balance performance. Furthermore, when the visual input to the user is non-veridical, exoskeleton assistance can magnify erroneous muscle inputs and negatively impact the user's postural control.**

***Index Terms*— Assistive devices, ankle exoskeleton, sensory information, postural control, quiet standing.**

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I. Introduction

TO CONTROL movement, humans must integrate the sensory information from the visual, vestibular, and somatosensory systems. When information from the sensory systems is degraded due to aging, disease, or environment, the central nervous system (CNS) undergoes a process of sensory reweighting [1], [2], [3] to maintain stability. For example, when the somatosensory information is degraded by standing on an unstable surface, humans may compensate by increasing the contribution of the vestibular and visual information to the state estimation [4], [5], [6], [7].

When standing quietly, people primarily rely on hip and ankle torques to maintain balance [8], [9]. Studies of unimpaired adults show a preference for utilizing an ankle strategy [10], [11], which prioritizes torque at the most distal joint. Reductions in ankle torque generation, such as those arising from motor loss due to aging, or changing external environment to a more narrow or compliant surface, lead to increased hip torques for quick center of mass (CoM) adjustments [12], [13]. Powered ankle-foot orthoses (AFOs) could potentially compensate for motor loss at the joint, but it is unclear how the exoskeleton assistance interacts with the sensorimotor system.

While sensory reweighting has been well studied in natural movements, especially in postural tasks, we do not yet understand how assistive devices affect the dynamics of this process. Identifying the relationship between robotic assistance and sensorimotor processes could lead to improved methods for augmenting human performance and improving rehabilitation. Providing torque assistance changes the relationship between the motor signal and motion at the assisted joint, potentially acting as a source of sensory error that could trigger the reweighting process during assisted balance [14], [15]. Because learning to use the exoskeleton requires the user to incorporate novel physical and sensory information into their internal model, novice users may have difficulty coordinating movement and responding to perturbations in the somatosensory channel. For example, older adults with age-related vestibular deterioration have been shown to use somatosensory feedback to compensate if vision is unreliable

[16]. Thus if the person has not learned how to use the device, wearing the exoskeleton may negatively impact the somatosensory feedback needed to maintain balance.

Previous studies of standing balance with passive AFOs have reported conflicting outcomes in postural tasks. People with peripheral neuropathy could improve postural control using the auxiliary cutaneous cues of a passive AFO device [17]. In contrast, several other studies with passive AFOs showed that restricting the ankle movement decreased the ability of the user to coordinate joint torques to respond to perturbations [18] which can have an adverse effect on postural control [19], [20]. For powered devices, there is little reported information on standing balance, one study showed that when perturbed unimpaired young adults could reduce their biological ankle torques with a controller designed specifically for balance [21], and a Parkinson's disease patient could reduce the center of pressure displacement [22]. However, it is unclear if the effects would remain under different sensory conditions.

One challenge in developing interfaces for balance assistance is understanding how different components (i.e. physical characteristics and control strategy) affect performance. For instance, the added friction at the exoskeleton joints could lead to increases in passive stiffness at the user's joints, requiring greater torques to control posture. During quiet standing, changes in passive stiffness at the joints have been shown to play an important role in balance performance [23], [24]. When sway angles remain small a greater ankle stiffness can result in smaller center of gravity velocities [25]. On the contrary, when disturbances or large sway angles are present, increased passive stiffness at the joints can have an adverse effect on postural sway [26].

Similarly, the assistive torques exerted at the joints by the exoskeleton could interfere with the somatomotor control loop. It is known that differences between sensory information and our internal predictions can reduce the reliability of the affected sensory channel and will require sensory reweighting [7], [27]. When a naive user wears an assistive device they must incorporate the tool into their internal model through experience. Thus, in the initial phases of learning the device, one could expect a mismatch between the somatosensory information and the user's internal model, triggering a greater reliance on visual and vestibular information. When learning how to use a robotic exoskeleton people may modify their gait strategies [28] or increase the priority of step width regulation [29] in order to preserve stability. These effects during the initial stages of adaptation to the device could be a result of the novel sensory information. Furthermore, additional sources of noise in the somatosensory channel could arise from time delays in the actuation process, due to signal filtering or intrinsic delays in the biological measurements used for control (e.g. sEMG [30]). Although we do not currently know if time delays in assistance reduce the reliability of the somatosensory information, it has been shown that time delays in the torque-generation process can have a significant destabilizing effect [31] and that exoskeleton assistance at the ankle has to respond faster than physiological movement in order to improve standing balance [32].

An important consideration is that robotic exoskeletons for partial assistance are commonly designed for voluntary control, where the goal of the device is to augment the user's intention in a transparent manner (i.e. minimal interference). This form of control assumes that the user's strategy is adequate for the task being performed. However, when there is conflicting sensory information a person may be unable to attenuate the automatic postural muscle adjustments to control balance [5], [33], resulting in increased postural sway. Thus, an assistive device that augments such muscle inputs could have a negative effect on the user's ability to control posture.

In this study, we used a virtual reality-based sensory integration in balance (VRSIB) test to measure visual sensory reliance by introducing dynamic visual information. Optic flow information interpreted by the visual system is integral to the perception of self-motion [34] and subsequently directing postural adjustments [6]. When presented in physical [35] or virtual reality (VR) [36] environments, vection, which is induced by optic flow, elicits a postural response opposing the perceived direction of sway in an attempt to restore static stability. By employing this method, we can observe the effects on postural control of torque augmentation at the ankle when the user attempts to correct the mismatched visual and vestibular information.

By performing the VRSIB test, we aim to characterize 1) the broad exoskeleton-induced changes in standing balance performance, and 2) the individual sensory system adaptations that are associated with these effects. Our primary hypothesis is that powered exoskeleton assistance will lead to decreases in balance performance in naive users across sensory conditions initially, but the magnitude of the decrease will be lower after the second day of training because of the gained experience. Our secondary hypothesis is that any decrease in balance performance from the exoskeleton will be greater in unstable ground conditions because it will be an additional somatomotor challenge, negatively affecting the user's ability to modulate postural balance using the available visual and vestibular information.

II. Methods

A. Data Collection

We recruited 11 young adults with no history of neurological or musculoskeletal impairment, and no prior experience using a powered exoskeleton. The subjects in this study were: 6 Females and 5 Males, age 24.34 ± 3.63 years, mass 71.12 ± 13.22 kg, height 1.74 ± 0.1 meters. Prior to this experiment, participants had not previously performed the VRSIB test. All participants provided informed written consent in accordance with the Temple University Institutional Review Board (IRB:28448).

We tracked the motion of the subjects using 16 motion capture cameras (sample rate: 120 Hz; Qualisys, Goteborg, Sweden) and 39 reflective markers (34 lower body, 5 upper body). We measured ground reaction forces using a split-belt instrumented treadmill (sample rate: 1200 Hz; Bertec, Ohio, USA). We collected surface electromyography (sEMG) from muscles in both legs for the exoskeleton's myoelectric controller (i.e. soleus, tibialis anterior) using a wired amplifier system (Delsys, Massachusetts, USA).

B. Experimental Protocol

Subjects performed 3 blocks of a custom VR-based test of sensory integration in balance (VRSIB, Fig. 1B). The VRSIB is a modification of the clinically validated sensory organization test (SOT) protocol [37], [38], [39], in which the visual and somatosensory components of the sensorimotor integration process are systematically perturbed.

To systematically perturb the visual information provided to the subject, the VRSIB employed an immersive virtual reality headset to modify optic flow in the visual surround. Virtual environments (VE) presented using both TV displays [40], [41], and head-mounted displays (HMD) [42], [43], [44] have been validated by previous studies for this application in substitution of the large projected screen of the traditional SOT protocol. An HMD device (Oculus Quest, Facebook Technologies, California, USA) with a refresh rate of 72 Hz displayed an immersive VE created with the Unity game development platform (Unity Technologies, California, USA) for the VRSIB. In the VE, participants looked down along a virtual hallway with the scene tracking the user's head movement. To further challenge the visuomotor process, a sinusoidal perturbation about the medial-lateral axis (i.e. pitch) of 20 deg amplitude and 0.25 Hz, centered about the eyes was added to the self-generated motion created by the standard tracking of the user's head movement [45] for specific conditions in the VRSIB.

The somatosensory information from the foot and ankle was perturbed by having the subject balance upon a soft foam block (Balance Pad, Airex AG, Switzerland). The foam block is a commonly used substitution in clinical settings for the sway-reference platform in the original SOT protocol.

In the VRSIB, we repeated the SOT protocol in three footwear conditions in order to identify how both passive and active assistance modifies the user's balance performance. In the baseline conditions, subjects wore their regular shoes. In the unpowered condition, the actuation systems of the exoskeleton were turned off, therefore only the inherent stiffness and friction in the orthosis remained. In the powered condition, the exoskeleton provided powered assistance in the plantarflexion directly only.

Therefore, in total, the VRSIB consisted of 18 conditions, the product of the 6 ground and vision conditions of the original SOT and the 3 footwear conditions. Subjects were instructed to stand with their gaze facing forward and maintain a still, upright position for the entire duration of each trial. For each sensory condition, we recorded three 30-second trials. The support conditions were: 1) firm ground (Firm) and 2) foam ground (Foam), where the subject stood on the balance board. The visual conditions were: 1) Eyes open (EO), where the subject viewed the room without the headset, eyes-closed (EC), the subject removed the headset and kept the eyes closed, and virtual reality (VR), where the sinusoidal VE perturbation was added to the natural head motion. The footwear conditions were: 1) regular shoes (shoes), 2) exoskeleton unpowered (ExoUnpowered), and 3) exoskeleton powered (ExoPowered).

To control for order effects in the repeated experiments, the presentation of the 18 conditions was partially counterbalanced. We chose not to fully counterbalance the study because of the substantial time required to repeatedly don and doff the exoskeleton. At the start of the study, subjects were randomly assigned to either the shoes or the two exoskeleton block conditions. For the exoskeleton blocks, they were randomly assigned to either the unpowered or powered conditions first. In each of the footwear blocks, the EO Firm combination was performed first to establish a baseline for static balance performance. All the remaining combinations for the visual and ground conditions were then presented in a randomized order. To identify learning effects, the VRSIB was repeated on two separate days within 18.45±10.00 days (mean ± std) apart from each other.

C. Powered Ankle Exoskeleton

1) Hardware: We used custom bilateral ankle-foot orthoses with one rotational degree of freedom in the dorsoplantar direction (Fig. 1A). The shank component is made out of carbon fiber and connects to the shoe through two steel rods. The shoes are standard orthotic shoes with a steel plate inlaid in the midsole. The shoe and shank components are connected by two pin joints at the medial and lateral sides. The exoskeleton is actuated through an off-board cable-driven system (Humotech, Pennsylvania, USA) with a flexible Bowden cable transmission. The force exerted at the heel is measured by an inline load cell (Omega Engineering, Stamford, Connecticut). To measure the dorsoplantar flexion angle of the exoskeleton, we attached an absolute magnetic encoder (MAE3, US Digital, West Virginia, USA) on the lateral side of the pin-joint at the ankle. The control signals to the actuators are generated through a Performance Real-Time target machine (Speedgoat, Maryland, USA) using Simulink 2020b (MathWorks, Maryland, USA).

2) Control: The exoskeleton uses proportional myoelectric control in closed-loop, and it is based on the sEMG difference between the soleus and the tibialis anterior muscles. The closed-loop system uses a proportional controller with damping injection [46] to compensate for the error between the user's desired torque and the measured ankle torque from the load cell and ankle encoder. The raw sEMG signals were processed in real-time. We applied a high-pass filter (2nd order Butterworth, cutoff frequency 80 Hz), followed by full wave rectification. Then, low-pass filtered the rectified signal (2nd order Butterworth, cutoff frequency 4 Hz) to get the linear envelope. Then we multiplied by a static gain to achieve a 40% level of assistance based on the peak torques for each individual subject. The scaled, filtered, and rectified signal was used as the desired torque profile at the ankle.

D. Data and Statistical Analyses

1) Measures of Postural Sway and Sensory Ratios: As a measure of balance performance we computed the equilibrium scores [17], [38], [39], [41], [47], [48], [49] which are the

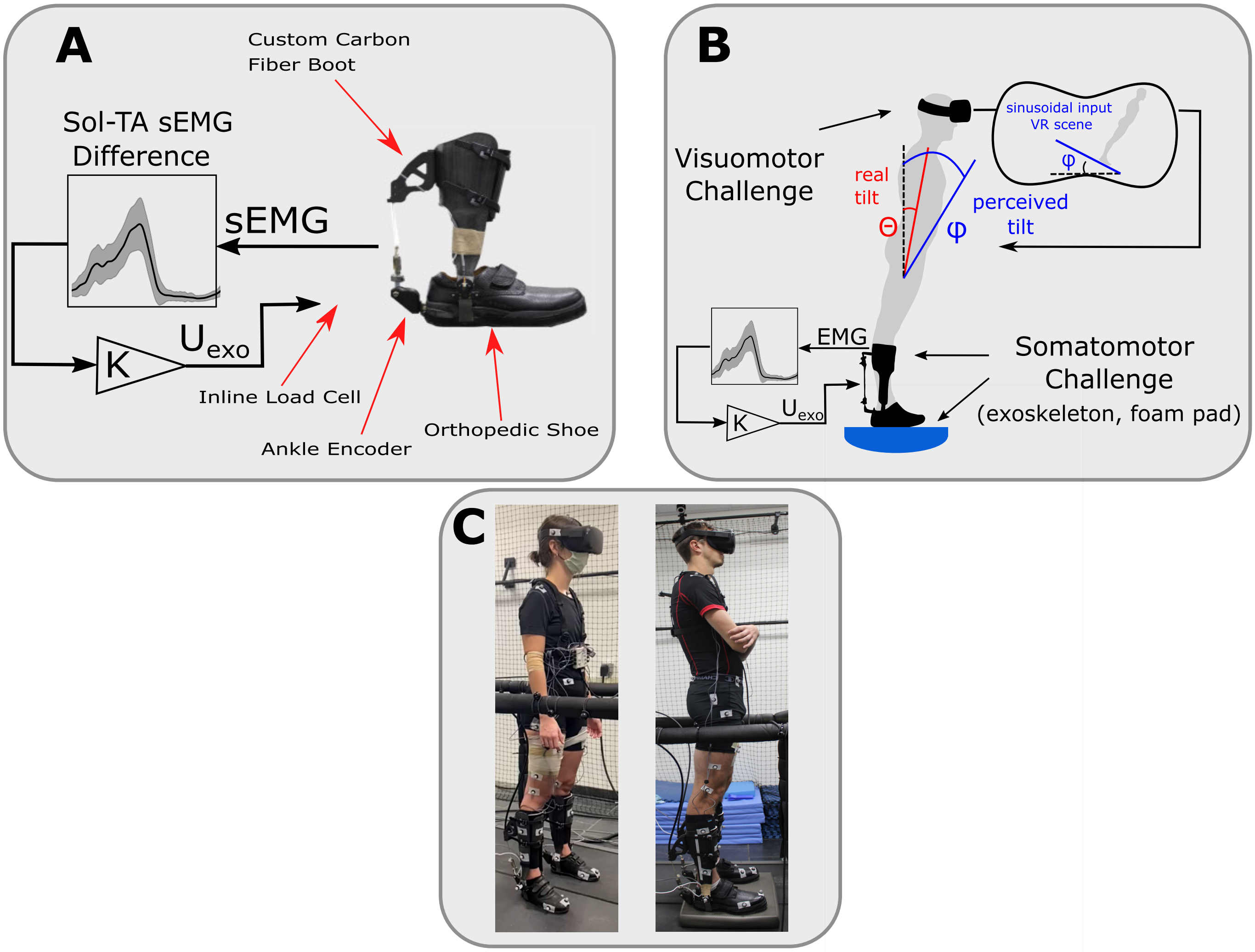

Fig. 1. A) During the exoskeleton trials participants used custom bilateral ankle-foot orthoses. When powered the exoskeleton was myoelectrically controlled operating in a closed loop using the sEMG difference between the soleus and tibialis anterior muscles as torque input and the measured torque at the ankle as feedback. B) Experimental paradigm: By altering the sensory information available to participants, we aimed to identify the effects that exoskeleton assistance has on sensory integration and postural control. Similar to the effects caused by mismatched visual information, we expected that the mismatch in somatosensory information would negatively affect balance performance. In the initial stages of learning the device users would need to update their internal model to incorporate the novel information. Furthermore, in the condition where visual information was altered, exoskeleton assistance could result in a magnification of incorrect muscle adjustments. C) Both pictures show participants during the powered exoskeleton trial and wearing the virtual reality headset. The one on the left was on firm ground and the one on the right was on the foam pad.

standard measurement used in the SOT. Equilibrium scores measure how well a participant's center of mass (COM) sway angle remains inside a theoretical limit of stability in the anterior-posterior direction. If a person sways more than the theoretical limit the equilibrium score will be set to 0 for that condition and trial.

Sway angles were calculated using a single rigid body model connecting the feet and the COM, and assuming one rotational joint at the ankle. The COM kinematics were calculated using a scaled musculoskeletal model and the OpenSim 4.0 API (AnalyzeTool, model: gait2354) [50]. The location of the ankle joints in space was measured using the bilateral malleoli markers on both limbs. The sway angle values were calculated between the vertical projection of the ankle-to-COM distance, and the leaning vector to the COM at each time point.

The equilibrium scores for each trial condition were then calculated using:

$$ES = \frac{12.5^{\circ} - [\theta_{max}(ant) - \theta_{max}(pos)]}{12.5^{\circ}}$$

where $\theta_{max}(ant)$ and $\theta_{max}(pos)$ are the maximum sway angles in the anterior and posterior directions respectively for each trial. The 12.5° is the theoretical sway angle limit (degrees) in the anterior-posterior direction clinically validated by the standard SOT.

The sensory ratios were computed according to the sensory organization test. Where the equilibrium scores (ES) for each condition are used for calculating the sensory ratios:

$$Somatosensory = \frac{ES_{ECFirm}}{ES_{EOFirm}}$$

$$Visual = \frac{ES_{EOFoam}}{ES_{EOFirm}}$$
$$Vestibular = \frac{ES_{ECFoam}}{ES_{EOFirm}}$$
$$Preference = \frac{ES_{VRFirm} + ES_{VRFoam}}{ES_{ECFirm} + ES_{ECFoam}}$$

where each ratio (i.e., Somatosensory, Visual, Vestibular) represents the ability of the participants to utilize the sensory information from each of the three systems to maintain stability. The Preference Ratio represents the degree to which the participants rely on visual information, even when it is incorrect.

2) Joint Kinematics and Kinetics: The force data were filtered using a fourth-order zero-lag low-pass Butterworth (15Hz). The joint kinematics and kinetics were obtained using the OpenSim 4.0 API [50]. The joint torques were normalized in magnitude to each subject's mass. We calculated the RMS joint torques for the ankle, knee, and hip for each trial.

3) Postural Control Strategies: To determine the percentage of the time in which participants used an ankle, knee, or hip strategy the moving correlations were computed between adjacent joints; ankle-knee, and knee-hip torques [8], [51]. This process was used based on the evidence that humans use mixed postural strategies to maintain posture [10], [52]. The strategy was assumed to be "ankle" when the correlations between knee-hip and ankle-knee torques were positive. The strategy was "knee" when the correlation of ankle-knee torque was negative and the correlation of knee-hip torques was positive. Finally, the strategy was "hip" when the correlation of ankle-knee torque was positive and the correlation of knee-hip torques was negative.

4) Statistical Analyses: A linear, mixed-effects model was used to test the effect of the exoskeleton on the equilibrium scores, sensory ratios, and postural strategies. The fixed effects were the exoskeleton, visual, and support conditions and the random effect was the subjects. The linear mixed model contained a full factorial design to account for interaction effects. If a significant effect was found, then a pairwise comparison was done using Tukey's Honest Significant Difference (Tukey HSD) test. All statistical analyses were performed in JMP® Pro 15 (SAS Institute Inc., Cary, NC, 1989-2022). All statistical tests were set to a significance level of 0.05.

III. Results

A. Equilibrium Scores

There was a significant fixed effect for support (DF = 1, F-Ratio = 1070.7, $p < 0.0001$) and visual (DF = 2, F-Ratio = 423.4, $p < 0.0001$) conditions, and there was a significant random effect for the subject ($p = 0.039$). All the fixed effect interactions are shown in Tab. I. In general, there was a significant interaction between the exoskeleton and support condition (DF = 2, F-Ratio = 40.0, $p < .0001$), and exoskeleton and visual condition (DF = 4, F-Ratio = 2.6, $p = 0.033$). There was no significant interaction between exoskeleton and day, while the interaction was significant for both support and day and visual and day.

The means and 95% confidence intervals are shown in Tab. II. We observed a statistically significant increase in equilibrium scores in the unpowered and powered exoskeleton conditions compared to shoes in the eyes closed and stable support conditions. For the tilting VR scene and eyes closed conditions, we observed a significant decrease in equilibrium scores for the exoskeleton unpowered and powered with respect to shoes. This effect changed for the unstable support when the visual input was also altered, in this case, exoskeleton powered was significantly different from unpowered and shoes (Fig. 2).

Additionally, there was a significant difference between the baseline condition (EO-Firm) and all other sensory conditions, with the largest decrease in equilibrium scores happening with respect to the VR-Foam condition (−27.69%).

TABLE I
Fixed Effect Test: Equilibrium Scores

Source (Conditions)	DF	F-Ratio	Prob>F
Day	1	0.0620	0.8034
Exo	2	1.5116	0.2210
Day-Exo	2	0.8710	0.4188
Support	1	1070.654	<.0001*
Day-Support	1	10.1598	0.0015*
Exo-Support	2	40.0275	<.0001*
Day-Exo-Support	2	0.5197	0.5949
Visual	2	423.3910	<.0001*
Day-Visual	2	7.7239	0.0005*
Exo-Visual	4	2.6341	0.0328*
Day-Exo-Visual	4	0.5173	0.7230
Support-Visual	2	180.1313	<.0001*
Day-Support-Visual	2	10.5616	<.0001*
Exo-Support-Visual	4	3.7856	0.0046*
Day-Exo-Support-Visual	4	0.4393	0.7803

B. Sensory Ratios

There was a significant effect of exoskeleton condition on the visual (DF = 2, F-Ratio = 6.82, $p = 0.002$) and vestibular (DF = 2, F-Ratio = 17.23, $p < 0.0001$) ratios, and a significant effect of day on the preference ratios (DF = 1, (DF = 1, F-Ratio = 5.66, $p = 0.02$). For the pairwise comparisons, we observed a significant reduction in the visual ratio when the exoskeleton was powered compared to regular shoes. We also observed a significant reduction in the vestibular ratio for both exoskeleton conditions with respect to shoes. We observed no significant differences in somatosensory or preference ratios (Fig. 2, Tab. II).

C. Ankle Torque

There was a significant fixed effect on the peak ankle torque from the exoskeleton (DF = 2, F-Ratio = 9.5, $p < 0.0001$), support (DF = 1, F-Ratio=889.9, $p < 0.0001$), and visual (DF = 2, F-Ratio = 336.9, $p < 0.0001$) conditions, and there was a significant random effect for the subject ($p = 0.026$). There was a significant interaction between the exoskeleton and and support condition (DF = 2, F-Ratio = 34.6,

TABLE II
EQUILIBRIUM SCORES & SENSORY RATIOS

Condition	Means (95% CI) Shoes	ExoUnpowered	ExoPowered	Difference (p-value) Shoes-ExoUnpowered	Shoes-ExoPowered	ExoUnpowered-ExoPowered
EO-Firm	90.8 (88.3, 93.3)	93.4 (90.9, 95.9)	94.3 (91.8, 96.8)	-2.6 (p=0.61)	-3.6 (p=0.11)	-0.9 (p=1)
EC-Firm	87.2 (84.7, 89.7)	91.4 (88.9, 93.9)	91.6 (89.1, 94.1)	*-4.2 (p=0.01)	*-4.4 (p=0.008)	-0.2 (p=1)
VR-Firm	85.9 (83.4, 88.4)	88.5 (86.1, 91.0)	89.2 (86.7, 91.7)	-2.6 (p=0.61)	-3.3 (p=0.2)	-0.7 (p=1)
EO-Foam	88.0 (85.5, 90.5)	88.6 (86.2, 91.1)	87.6 (85.2, 90.1)	-0.6 (p=1)	0.4 (p=1)	1.0 (p=1)
EC-Foam	82.8 (80.4, 85.3)	77.8 (75.3, 80.3)	76.3 (73.8, 78.8)	*5.0 (p<0.001)	*6.5 (p<0.001)	1.5 (p=0.99)
VR-Foam	68.6 (66.1, 71.1)	65.5 (63.0, 68.0)	61.3 (58.8, 63.8)	3.1 (p=0.31)	*7.3 (p<0.001)	*4.2 (p=0.01)
Somatosensory	0.97 (0.95, 0.98)	0.98 (0.96, 1.0)	0.97 (0.96, 0.99)	-0.012 (p=0.27)	-0.005 (p=0.67)	0.007 (p=0.5)
Visual	0.97 (0.96, 0.99)	0.95 (0.94, 0.97)	0.93 (0.91, 0.95)	0.018 (p=0.23)	*0.04 (p=0.001)	0.02 (p=0.11)
Vestibular	0.92 (0.89, 0.95)	0.86 (0.84, 0.88)	0.82 (0.79, 0.85)	*0.06 (p=0.002)	*0.1 (p<0.001)	0.04 (p=0.074)
Preference	0.92 (0.90, 0.94)	0.93 (0.91, 0.96)	0.92 (0.89, 0.94)	-0.01 (p=0.77)	0.003 (p=0.98)	0.02 (p=0.64)

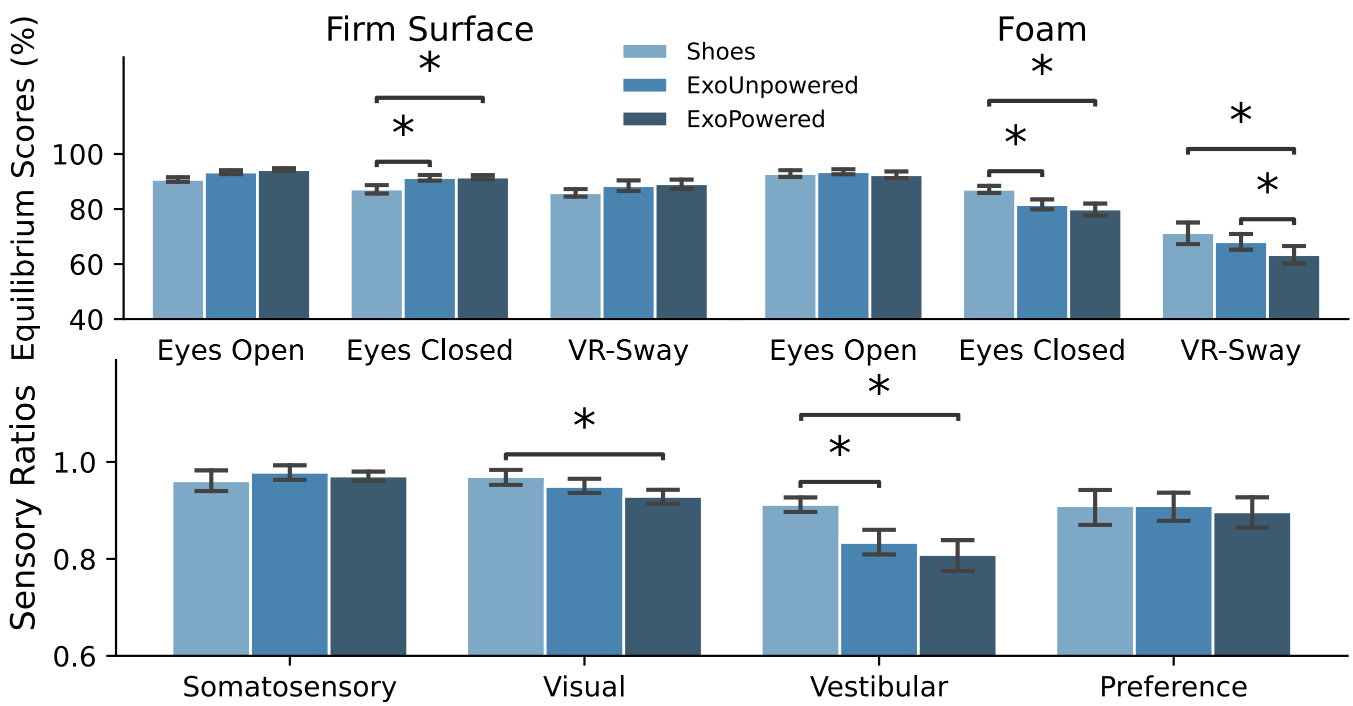

Fig. 2. Mean Equilibrium Scores and sensory ratios for 11 unimpaired subjects performing a virtual reality-based test of sensory integration in balance (VRSIB). The plots contain the joined data for the two days in which the experiment was performed. The error bars represent the 95% confidence intervals, and the asterisks represent a significant pairwise difference determined using a Tukey HSD test ($p < 0.05$). In the stable support conditions, there was a slight increase in equilibrium scores when the participants wore the exoskeleton. The only difference observed between the exoskeleton's unpowered and powered conditions was when the support surface and the perturbed optic flow were altered.

$p < .0001$), exoskeleton and visual condition (DF = 4, F-Ratio = 3.5, $p = 0.007$), and between support and visual condition (DF = 2, F-Ratio = 72.1, $p < 0.0001$).

We observed a statistically significant increase in peak magnitude and standard deviation of ankle torque (expressed in Nm/kg, Fig. 3) with respect to wearing their regular shoes only when participants wore the exoskeleton powered in the EC-Foam condition (peak: difference = 0.11, $p < 0.0001$; Std: difference = 0.030, $p < 0.0001$), and in the VR-Foam condition (peak: difference = 0.14, $p < 0.0001$; Std: difference = 0.041, $p < 0.0001$). A significant increase in peak ankle torque was also observed with respect to the exoskeleton unpowered unpowered condition in the VR-Foam condition (difference = 0.12, $p < 0.0001$). The peak ankle torque was, for all trials, in the direction of the plantar-flexion assistance of the exoskeleton.

The percentage of ankle torque produced by the exoskeleton in each condition was: in EO-Firm 10.95%, EC-Firm 10.72%, VR-Firm 10.90%, EO-Foam 10.92%, EC-Foam 11.63%, VR-Foam 15.15%. There was a statistically significant increase in the percentage of ankle torque generated by the exoskeleton in the VR-Foam condition with respect to all other assisted trials. However, there was no difference between any of the other conditions. The RMS tracking error between the torque input and actuator output from the exoskeleton was 0.58 ± 0.16 Nm ($mean \pm std$) with a delay in the tracking of

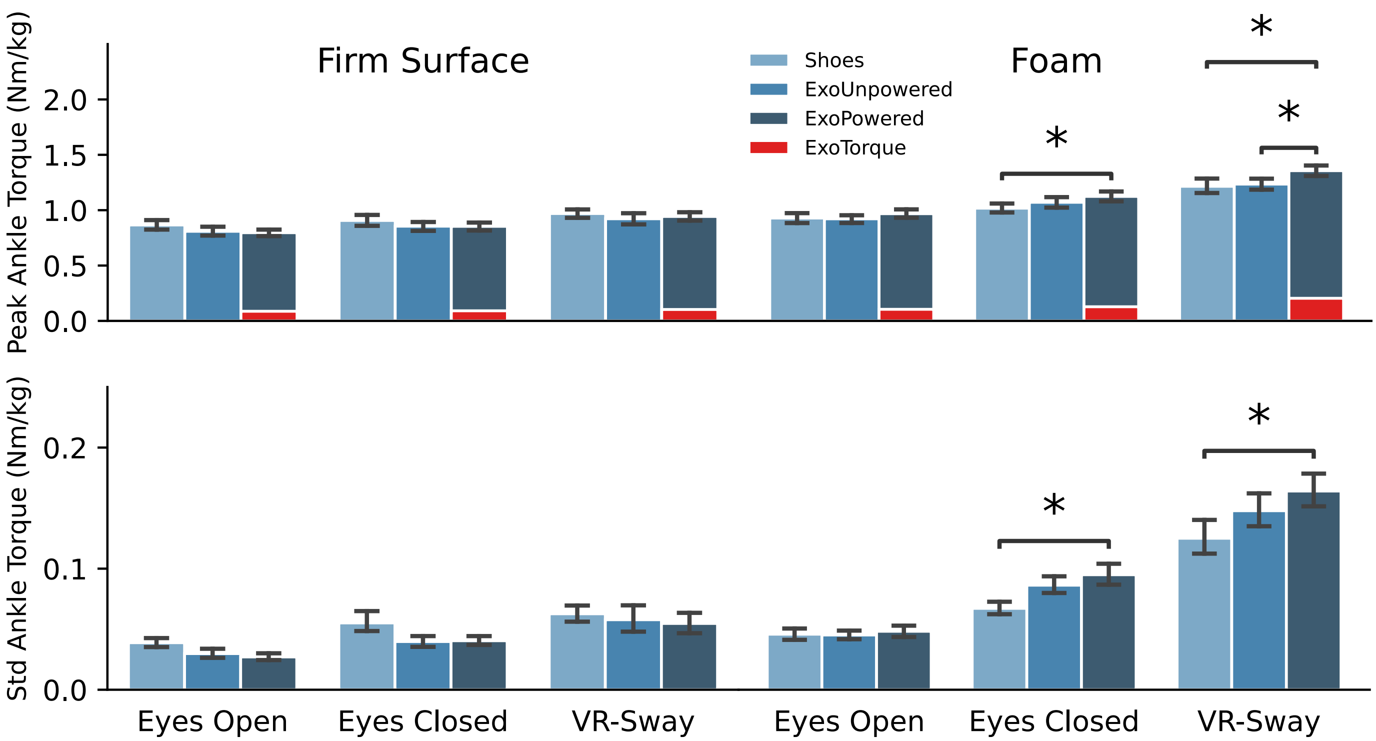

Fig. 3. Peak torques at the ankle, exoskeleton torque contributions, and standard deviation (Std) in ankle torque for 11 unimpaired subjects performing a virtual reality-based test of sensory integration in balance (VRSIB). The plots contain the joined data for the two days in which the experiment was performed. The bars show the mean, the error bars represent the 95% confidence intervals, and the asterisks represent a pairwise significant difference determined using a Tukey HSD test ($p < 0.05$). Participants only significantly increased their ankle torque in the exoskeleton powered conditions when the support was unstable and the visual information was either blocked or non-veridical. In these conditions, there was also a significant increase in ankle torque variability when the exoskeleton was turned on compared to when wearing regular shoes.

10.2 ± 3.5 mSec ($mean \pm std$). The delay between biological ankle torque and exoskeleton torque was 120.2 ± 59.2 mSec ($mean \pm std$) (Fig. 4), which is within the usual physiological delay in sEMG signals [32], [53].

D. Joint Strategies

The ankle strategy was predominant in all exoskeleton and sensory conditions, being used in shoes $97.45\% \pm 3.9$, in the unpowered $96.42\% \pm 5.4$, and when powered $97.16\% \pm 4.1$ of the trial time. We did not find significant differences in the percentage of ankle strategy used within trials across exoskeleton or sensory conditions.

IV. Discussion

We aimed to characterize the effects of an ankle exoskeleton on balance performance and sensory reweighting under different sensory conditions. For this, we used a virtual reality-based test of sensory integration in balance (VRSIB). In response to powered ankle assistance, our population's ability to regulate their postural balance using visual and vestibular information dropped compared to when wearing their regular shoes, which was in accordance with our secondary hypothesis. However, the mechanism was different than expected because our results showed that the somatosensory ratio was not significantly different between exoskeleton conditions. Instead, we found complex interactions between the exoskeleton and sensory conditions that contributed to both increases and decreases in equilibrium scores.

In addition, significant differences were found in equilibrium scores between the powered and unpowered exoskeleton conditions suggesting differential impacts of passive and active assistance depending on the available sensory information. Our results also showed that repeated exposure (i.e. training day) was not a significant fixed effect by itself nor was the interaction between the exoskeleton and day significant. Therefore, we cannot conclude if the equilibrium scores would be improved with sufficient time to learn the exoskeleton. Our findings have implications for understanding the role of assistance-induced sensory effects and the design of assistive devices for populations with impaired function.

In unimpaired populations, loss of balance has been shown to occur primarily when both visual and somatosensory information is affected leaving only vestibular input available for postural control. Our results with the exoskeleton agreed with previous reports of unassisted balance with subjects showing a significant decrease in postural performance under unstable ground conditions (foam), and when visual information was missing (eyes closed) or unreliable (virtual reality). However, the exoskeleton alone did not have a fixed effect on equilibrium scores but did have significant interaction with the support surface and visual condition. When visual information was reliable (eyes open) or somatosensory was reliable (firm ground), there were no significant differences in equilibrium score between the Shoes, ExoUnpowered, or ExoPowered conditions. This shows that the exoskeleton primarily impacted

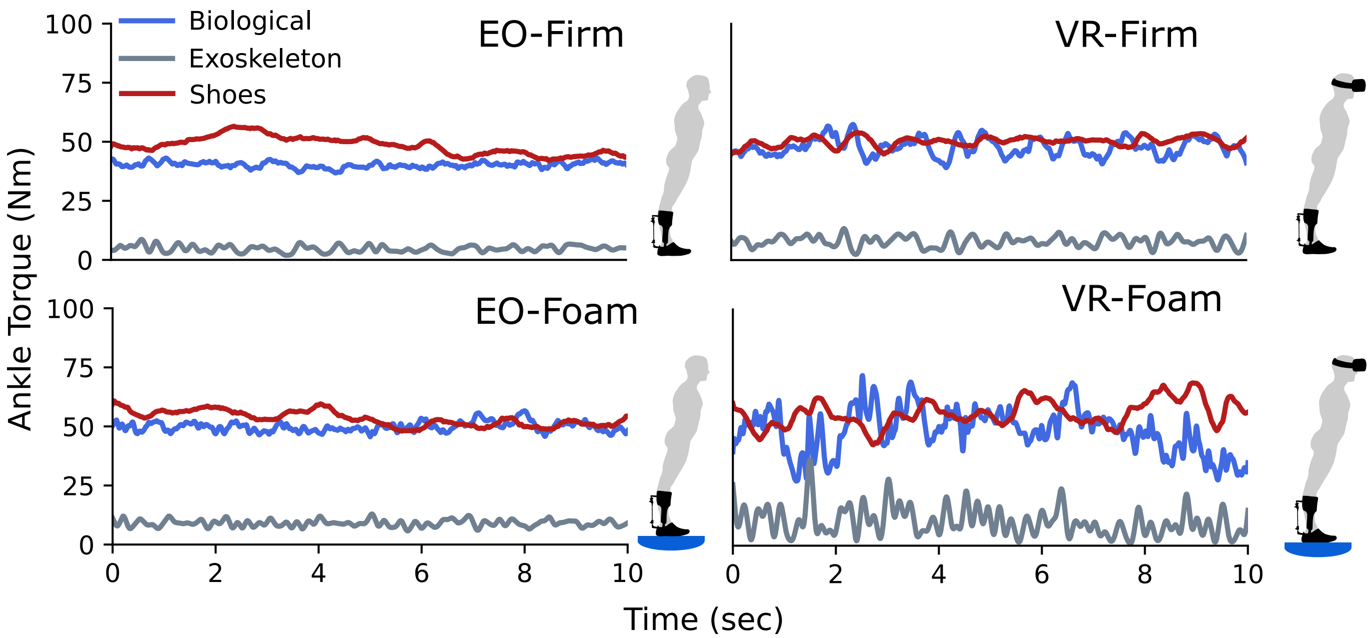

Fig. 4. Biological, and exoskeleton ankle torque time-series for the exoskeleton powered condition and ankle torque in the shoes condition for a representative subject in four of the tested sensory conditions: eyes open on firm ground (EO-Firm), eyes open on the foam pad (EO-Foam), wearing the virtual reality headset on firm ground (VR-Firm), and wearing the virtual reality headset on the foam pad (VR-Foam). The increase in magnitude and variability is consistent with the observed significant interaction of the exoskeleton with the support surface and visual conditions.

balance performance in situations where the visual system was unreliable and could not help compensate for disturbances in the somatosensory channel due to the foam support.

In contrast to our hypothesis, the exoskeleton did not negatively affect the use of the somatosensory system for balancing. Rather, in all the firm ground conditions where the orientation information from the somatosensory system was reliable, equilibrium scores increased in both exoskeleton conditions compared to Shoes. However, the differences were only significant in the eyes-closed, firm support condition, which may have been because of the statistical power of our sample size (Tab. II). Because the increases in equilibrium score in the firm ground condition were similar in magnitude for both the ExoPowered and ExoUnpowered conditions, it suggests that the impact from the exoskeleton was not because of the active assistance. Across the support and vision combinations, the ExoUnpowered and ExoPowered conditions were only significantly different for the foam, VR condition which shows that the impact of powered assistance was greatest when both vision and somatosensation were challenged.

As a result, we found significant decreases in the visual and vestibular ratio across exoskeleton conditions because the visual and vestibular systems could not fully compensate for the additional impact of the exoskeleton in the somatomotor loop. The visual ratio significantly decreased in the powered exoskeleton condition with respect to shoes because the main contribution was the increase in the baseline (EO-Firm) condition while the EO-Foam condition stayed constant. Similarly, the increase in baseline (EO-Firm) performance approached significance for the ExoPowered condition and facilitated a corresponding drop in vestibular ratio. However, sensory ratios demonstrated a deteriorated ability of participants to utilize vestibular information that was significant in both exoskeleton conditions. This is the result of a second and more dominant effect of the exoskeleton, both actuated and passive, observed in the lower dual-challenge sensory condition (EC-Foam) equilibrium scores that further decreased this value.

Our results may provide insight into previous findings in studies of postural control with passive ankle-foot orthoses that showed seemingly contradictory results, e.g. improvements in balance performance [17] and greater postural sway [18]. We surmise that the increase in equilibrium score in the firm ground condition resulted from the passive effects of the device because those conditions were not sufficiently challenging for our relatively young and unimpaired adult population. This is further supported by the pattern of unchanged peak ankle torques and the lower ankle torque variability on the firm ground across exoskeleton conditions. In the firm ground conditions, there was a small but non-significant decrease in the combined peak and standard deviation in ankle torque from the person and exoskeleton which shows that less active control was needed. However, in the eyes-closed and VR-sway foam conditions, there were significant increases in peak exoskeleton torque and ankle torque variability accompanying the decrease in equilibrium score. This interpretation aligns with previous studies that showed under unstable conditions the increase in stiffness can be detrimental to postural control [19], [20]. These results point to a critical need for further investigation of exoskeletons in balance control because previous studies with the SOT have shown that the altered somatosensation and vision of the VR-Foam condition is the most sensitive for predicting falls [44], [54].

While we observed a significant fixed effect interaction between day-visual condition and day-support condition, we did not see a significant interaction between day-exoskeleton condition. This implies that participants became familiar with the VRSIB conditions, but not necessarily with the exoskeleton in the two-day testing period. Previous reports have shown that there are learning effects associated with repetitive administrations of the sensory organization test [55], [56]. Improved performance when wearing an assistive device has also been shown to be associated with training time and learning effects [57], [58], [59], where participants exhibited good retention after 7 to 15 days. However, our results show no significant improvements in the use of the powered exoskeleton on the second day of testing. We believe the lack of assistance-related improvement in performance on the second day to be a consequence of the altered sensory conditions and not the time between days. One study that tested retention after an extended period of time showed little reduction in the learned motor skills after 2-3 months of the initial training period [60].

This study showed no difference in postural strategies across exoskeleton or sensory conditions, with ankle strategy largely dominating. The prevalence of the ankle strategy, in our study, justifies the use of a single-segment postural model for the calculation of the equilibrium scores. Ankle strategies are sufficient to correct small center of mass deviations and are dominant in less demanding balance tasks [10], [11]. In future work, we plan to investigate sensory reweighting in an older adult population, where a hip strategy may be used to compensate for distal weakness and where visual perception is diminished.

These findings are limited to a myoelectric controller, where the muscle activity of the user directly relates to the commands sent to the exoskeleton. In these types of controllers, it is possible to magnify erroneous muscle inputs that arise from the mismatch between perceived and true body orientation. Other common control methods for exoskeletons may result in different sensory effects than those presented here. For example, a torque-based method that aims to drive the ankle angle toward the upright body configuration could act against the user when there is a mismatch between vestibular and visual information. In addition, our interpretation of the learning effects associated with the device in conditions where the sensory information is missing or altered is limited to a two-day testing period, with a relatively long delay between day one and day two (18.45 ± 10 days). We acknowledge this as a limitation in our study that resulted from the global health situation, making it difficult to bring participants back at the desired time. It remains to be seen if improvements associated with learning to use the device would be possible with further training. Finally, we cannot fully disentangle which effects from assistance contributed the most to the drop in balance performance. Nonlinear effects from friction, unidirectional torque magnification (plantarflexion), and physiological delays in response [32] could also negatively contribute to postural control.

In this study, we measured the effects of myoelectric ankle joint assistance during postural tasks where the sensory information was systematically perturbed. Our results suggest that joint torque augmentation may not necessarily be beneficial for balancing tasks, particularly in situations when one or more sensory systems are receiving unreliable environmental information, at least in the initial stages of learning to use the device. The general belief in partial-assistive devices is that the user's intended control is adequate for the task, and through joint torque augmentation, we can reduce energetic cost, or enhance stability. However, in a situation where sensory information becomes unreliable, torque magnification will also magnify the erroneous muscle inputs if they occur. This highlights the importance of finding a proper measure of "torque input correctness" before augmenting the user's intent. Future work will focus on distinguishing when the user is commanding a force to the joint that will drive them to instability. In such a scenario, the minimum the device should do is become inactive to avoid further destabilizing effects.

V. Conflicts of Interest

The authors declare that no benefits in any form have been or will be received from a commercial party related directly or indirectly to the subject of this manuscript.

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