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# Human Gait During Level Walking With an Occupational Whole-Body Powered Exoskeleton: Not Yet a Walk in the Park

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**ABSTRACT** With rapid advancements in exoskeleton technologies, a whole-body powered exoskeleton (WB-PEXO) for augmenting human physical capacity (a ''super-operator'') is generating increasing attention as an integral part of Industry 4.0. Our understanding of WB-PEXO use is lagging, however, largely due to the lack of detailed evaluations via human-subjects testing of a WB-PEXO. We examined (independently from the manufacturer of a WB-PEXO) the potential impacts of using a state-of-the-art WB-PEXO prototype (pre-alpha prototype version of the Sarcos Guardian® XO®) on users ($n = 5$) during a common basic activity in the workplace, level walking. With emphasis on the ''human'', impacts of XO use (compared to a no EXO baseline) were assessed in terms of lower limb intersegmental coordination, muscle activity, and postural dynamic stability. A larger variance between participants was observed for intersegmental coordination with XO use, and participants appeared to rely on more hip motions. When using the XO, participants exhibited higher muscle activity levels in the lower limb muscle groups monitored. Further, there was a moderate to high similarity in muscle activity profiles between the XO and no EXO conditions ($R_{XY}(\tau) = 0.70 - 0.92$), yet muscle activity profiles when using the XO were generally time-lagged from those without the XO. We discuss the results within the context of developing a mental model for walking with the XO, and aspects of human-robot interaction such as transparency of the XO and understanding user state and intention. In concluding, we outline several future research topics for occupational WB-PEXO development.

**INDEX TERMS** Gait performance, human-robot interaction, occupational exoskeleton, whole body system.

## I. INTRODUCTION

Use of wearable robotic systems such as exoskeletons – non-powered (passive) or powered (active) mechanical systems that enhance/assist the strength and/or performance of the wearer – has generated great interest in occupational applications. Work-related musculoskeletal disorders (WMSDs) continue to be an important health issue in the workplace of many industrialized countries [1], [2]. The physical augmentation offered by (occupational) exoskeleton use is an innovative solution to control WMSDs, particularly during physically demanding jobs [3], [4]. Lab-based studies have indicated beneficial effects of passive exoskeleton on worker safety and performance, for example, during overhead tasks [5], [6], static trunk bending [7], [8], and repetitive lifting [9], [10]. Further, though limited in quantity and comprehensiveness, several industry pilots have addressed use, acceptance, and effectiveness of passive exoskeletons (e.g., Ford [11], Toyota [12], Boeing [13]). Passive exoskeleton use is thus a very promising intervention approach in industrial settings, and a majority of commercially-available exoskeletons are passive (e.g., exoskeletonreport.com).

In contrast, powered exoskeletons are relatively less mature, being more in the developmental phases. Existing work on powered, occupational exoskeletons has had more emphasis on prototyping, structural design, and control algorithms while targeting a specific body part such as the low back [14], [15], the lower extremity [16], [17], or the upper

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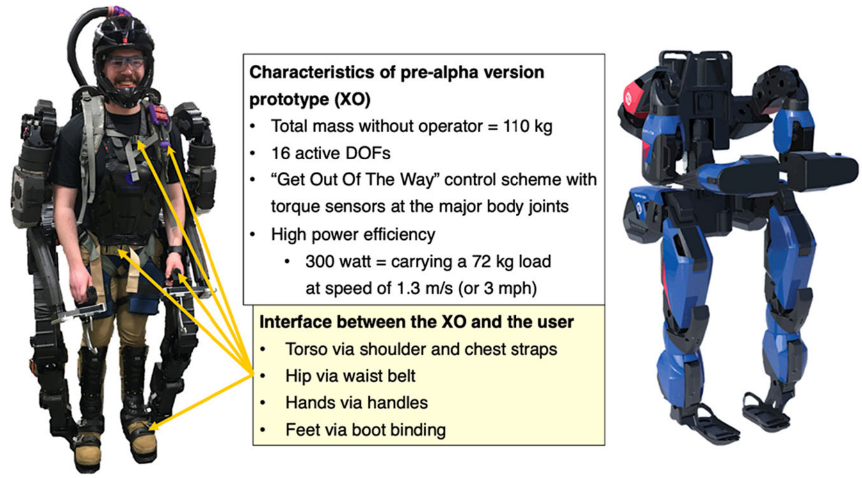


**FIGURE 1.** Early pre-alpha prototype of the occupational whole-body powered exoskeleton (WB-PEXO) tested (Left) and the recent version demoed at the 2020 Consumer Electronics Show (Right), Guardian® XO®, Sarcos Robotics (www.sarcos.com).

extremity [18], [19]. There are a few reports on the effectiveness of a commercially-available powered exoskeleton for the low back [Hybrid Assistive Limb (HAL™) for Care Support, Cyberdyne, Ibaraki, Japan] during snow shoveling [20], repetitive lifting [21], and patient handling [22]. These and related studies suggest that a powered exoskeleton is likely heavier and more complex in design than a passive one, but that it can be more versatile and functional with active control. Powered exoskeletons are also consistent with the idea of a "super-strength operator" in the next industry revolution (Industry 4.0; Romero *et al.* [4]), in that such augmentation technology is likely to be an important part of the future workforce as Industry 4.0 is being realized.

The development of a whole-body powered exoskeleton (WB-PEXO), however, is still a significant challenge. As early as the 1960s, though, the technical concept of WB-PEXOs was promoted for moderate-heavy duty jobs [23], [24]. The first practical implementation of a WB-PEXO was the Hardiman, a hydraulic experimental prototype with 28 joints and two robotic end effectors, developed by General Electric between 1965 and 1971 [25]. Since then, improvements in actuators and control algorithms have enabled the development of modern WB-PEXOs. In the 1990s, a team from Berkeley proposed a novel under-actuated, hybrid material handling system with passive and active joints operating through force-feedback control [26]. Little information, however, is available on the overall performance of this system. The Body Extender is an electrically actuated WB-PEXO, developed as a research platform to investigate user augmentation user during heavy load handling [27]. Sankai [28] reported the (at the time) latest version of the Cyberdyne HAL™-5 WB-PEXO, which uses electromyographic signals to estimate user intention and can support a lifting capacity of up to 70 kg. Several other WB-PEXOs have been developed. For example, in 2010 Panasonic Activelink presented a WB-PEXO designed for heavy-duty jobs [29], and the Daewoo WB-PEXO prototype was described in 2014 had a lifting capacity of 30 kg during tasks relevant to shipbuilding [30].

## II. STATE-OF-THE-ART OF WHOLE-BODY POWERED EXOSKELETONS (WB-PEXOS)

The Sarcos Guardian® XO® ( www.sarcos.com) is an advanced, near market-ready WB-PEXO, and its untethered operation was demonstrated at the 2020 Consumer Electronics Show. This system is lineal to the hydraulic XOS series from Sarcos Robotics. In the early 2000s, the XOS series were hydraulically-actuated WB-PEXOs developed for logistics demonstrations purposes. The XOS 2, demonstrated in 2010, was 50% more energy-efficient than XOS 1, and was a tethered exoskeleton with 24 active degrees-of-freedom (DOFs) that provided a lifting capacity of up to 45 kg. Because of the high power consumption of the XOS series, Sarcos has developed a new WB-PEXO with electric actuators that: 1) can address both the energetic autonomy and heat generation/dissipation issues of previous XOSs; and 2) can enhance human capacity, specifically in industrial settings. With a dramatic reduction in power consumption, Sarcos announced the first battery-powered WB-PEXO, the Guardian® XO®, for occupational applications.

The current study examined an early, pre-alpha version of the XO® (Figure 1), with mass = 110 kg and which included 16 active DOFs spanning the shoulders (flexion/extension and ab/adduction), elbows (flexion/extension), trunk (axial rotation and lateral bending), hips (flexion/extension, ab/adduction, and axial rotation), and knees (flexion/extension). With torque sensors at the major body joints, the XO® controller uses a "Get-Out-Of-The-Way" control scheme to effortlessly follow human movements and to amplify the user's joint torques, by employing an optimal torque control and effective dynamic inertial compensation [31]. It also includes several tunable parameters – including actuation gains (torque amplification) and payload and gravity compensation – each of which can be adjusted for a specific user/operator. User control inputs are obtained using six force sensing units embedded in the hands, feet, upper back and pelvis components of the XO®. The pre-alpha version XO® examined, however, had no task-specific optimization applied to the controller and had underactuated ankle joint. Note that detailed XO® design specification and implementation approaches are proprietary and thus not made available here.

Despite considerable research efforts and progress in occupational WB-PEXO development, no detailed information is available in the current literature on human-subject testing with market-ready WB-PEXOs or their prototypes that have enough fidelity for testing. This lack of evidence is likely due to concerns about intellectual property. However, recent literature on open innovation (e.g., [32], [33]) emphasizes the importance of active collaborations between an organization (or firm) and a variety of external partners to support information sharing and innovation. An example of such a collaborative effort is understanding how using a WB-PEXO affects the human operator in terms of physical and mental demands, usability, task performance, and workplace safety, and modeling human-exoskeleton interactions. Understanding of these aspects can, in turn, help to: 1) promote prevention through design (PtD) in the development of WB-PEXOs that proactively ''design out'' any potential risks to the user; 2) advance research and practices for safe implementation and use of WB-PEXOs, by informing those in the industry about when, how, and the extent to which WB-PEXO use can benefit their workers and workplaces; and 3) prompt regulatory bodies to prepare the necessary rules and regulations (or a plan for these), for future successful applications of such technologies.

We thus conducted a human-in-the-loop study to assess the potential impact of using a state-of-the-art WB-PEXO prototype on users during level walking, as part of a larger ongoing research project that investigates how human operators learn, use, trust/accept and adapt to WB-PEXO technologies. In this first study, level walking was selected since it is a fundamental motor skill, and workers with a WB-PEXO are likely to move about their workplace, regardless of the particular tasks for which the WB-PEXO is eventually going to be used. The impact of the WB-PEXO prototype was examined in terms of the user's control of body motion, muscle activity, and interactions between them and the system. Note that the current study was not intended as a formal user testing (given that an early prototype was examined here) but rather as third-party, human-in-the-loop testing to inform the current state of occupational WB-PEXO development, and support the development of appropriate use-cases, measurement metrics, and other methods. Testing was conducted independently from Sarcos, though company representatives provided technical support to operate the XO® prototype. The current study was exploratory by nature, designed to generate information to facilitate the future development of WB-PEXO technologies.

## III. METHODS

### A. PARTICIPANTS

A convenience sample of five healthy male participants completed the study. Prior to data collection, informed consent was obtained from all participants following procedures approved by the Virginia Tech Institutional Review Board. Respective mean (SD) age, body mass, and stature were 36 (11.9) yrs, 79.0 (8.2) kg, and 183.8 (2.8) cm. None of the participants had self-reported recent (past 12 month) or current musculoskeletal disorders or injuries. Participants were trained to use the XO® prototype in 3-5 sessions over a 2-3 week period, during which they explored a range of tunable parameters and selected their preferred parameter values.

### B. EXPERIMENTAL PROCEDURES, AND INSTRUMENTATION

Participants completed trials of level walking with and without using the XO® prototype. For each trial, participants wore the same type of standard, steel-toed work boots, and they were asked to walk across a linear 10-meter track at a comfortable speed. In both baseline (no XO) and XO conditions, participants first performed several walking trials for familiarization and then completed six walking trials for data collection. When using the XO, participants were fitted comfortably with the XO, and tunable parameters for XO control were adjusted; the fit and comfort were verbally checked by the investigators. Any fine adjustments were made when needed. If an adjustment was made, participants were asked to complete an additional walking trial prior to data collection. A minimum of 1-minute rest was provided between the XO conditions and after familiarization. The presentation order of XO conditions was random across participants.

During each walking trial, whole-body kinematics were captured at 60 Hz using a wearable inertial motion capture system (MVN Awinda, Xsens technologies B.V., Enschede, the Netherlands) with 17 inertial measurement units (IMUs). Note that gait speed was obtained from the MVN system. Muscle activity was monitored using a telemetered surface electromyography (EMG) system (Ultimum™ Noraxon, AZ, USA). After appropriate skin preparation, pairs of pre-gelled, bipolar, Ag/AgCl electrodes with a 2.5 cm inter-electrode spacing were placed bilaterally over four accessible muscle groups following earlier studies [34]: vastus lateralis (VL; role = knee extension), biceps femoris (BF; hip extension and knee flexion), tibialis anterior (TA; ankle dorsiflexion and inversion), and medial gastrocnemius (MG; ankle plantarflexion). To normalize EMG signals, isometric maximal voluntary contractions (MVCs) were performed for each muscle group prior to the gait trials. Participants sat on a chair and performed separate knee and ankle flexion and extension against manual resistance, while the included knee and the ankle joint angles were at ~90°. For a given muscle group, MVC trials were replicated twice, during which non-threatening verbal encouragement was provided. At least 30 seconds of rest were provided between MVC trials. After completing MVC trials, at least 5-min. of rest was given before proceeding with the protocol. Raw EMG signals were sampled at 2 kHz, and these signals were subsequently band-pass filtered (20-450 Hz, 4$^{th}$-order Butterworth, bidirectional). EMG signals were then low-pass filtered (6 Hz cut-off, 2$^{nd}$ order Butterworth, bidirectional) to create linear envelopes. A normalized EMG (nEMG) envelope was obtained for each muscle group using the corresponding maximum value obtained during MVCs (i.e., 0-100%).

### C. GAIT ANALYSIS

Gait performance was assessed based on gait cycles obtained during the level-walking trials. A gait cycle was defined from the time of one right heel strike to a subsequent right heel strike. Heel strikes were first identified using a combination of gait event data from the inertial motion capture system and the method described by Zeni *et al.* [35], then identified heel strikes were visually confirmed.

#### 1) LOWER-LIMB INTERSEGMENTAL COORDINATION

Kinematic coordination between the movements of major leg segments was examined using the planar law of intersegmental coordination [36]. This law suggests that lower-limb elevation angles (EAs) – the angles between the limb segment projected onto the sagittal plane and the vertical axis – covary and form a teardrop-shaped loop on a so-called intersegmental covariation plane. EAs were calculated for the thigh, shank, and foot, and were re-sampled to 100 points for each gait cycle. Normalized gait times were defined as: first heel strike $t = 0\%$, and next heel strike $t = 100\%$. Principal component analysis (PCA) was then applied to normalized EAs to characterize the noted loop, by determining planarity (i.e., the percentage of variance explained by the first 2 PCs), covariance loop width (i.e., percentage variance explained by the second PC), and the orientation of the covariance plane. The latter is the direction cosine between the 3$^{rd}$ principal axis of the loop and the positive semi-axis of the thigh segment, often referred to as $\mu_{3t}$. A planarity value of 100% means an ideal plane (i.e., the 3$^{rd}$ eigenvalue = 0), and values of ≥97-99% are typical for human gait [37]. Planarity and the covariance plane orientation reflect temporal covariances among EAs during gait [38], [39], and these change systematically during different gaits such as marching, crouching, obstacle crossing [37].

#### 2) CROSS-CORRELATION OF LOWER-LIMB MUSCLE ACTIVITY WITH VS. WITHOUT XO USE

Given that no information was available about how the prototype XO interacts with the user during level walking, we examined if participants had similar muscle-recruitment patterns (via nEMG) at comparable gait phases with vs. without XO use. Specifically, normalized cross-correlation (NCC) with time delay ($\tau$) was performed on the pairs of corresponding nEMG profiles (i.e., $X(t)$ and $Y(t)$) in the two XO conditions [40], or $R_{XY}(\tau) = E\,[X(t)Y(t-\tau)]$. Note that $R_{XY}(\tau) \in [-1, 1]$ is a measure of the similarity in shape between two profiles after imposing a delay. To calculate this, for a given XO condition the nEMG values from each muscle group were re-sampled to 100 points for each gait cycle, and these points were then ensemble-averaged to produce nEMG profiles across each gait cycle.

#### 3) LOCAL DYNAMIC STABILITY

To quantify gait stability during specific gait phases, we used phase-dependent local dynamic stability (pLDS; Ihlen *et al.* [41]) and a conventional LDS, since they provide distinct information regarding postural stability [42], [44]. The former was to calculate the time-dependent Lyapunov exponent ($\lambda_{pLDS}$):

$$\lambda_{pLDS} = \frac{d < ln < d_i(t) \gg}{dt} \tag{1}$$

where the outer bracket <> represents the mean of the ensemble of logarithm of divergence curves, and $(< d_i(t) >)$ is the distance between the reference point and the $i^{th}$ nearest neighbor trajectory at time $t$. Before taking the time derivative $(d/dt)$ of the divergence curve over a stride, the logarithm of divergence curves were low-pass filtered (6 Hz, 2nd order Butterworth, bidirectional), similar to the approach by Mahmoudian *et al.* [42]. To compute the divergence curves, a 6D state space was constructed using 3D accelerations and angular velocities of the pelvis IMU of the inertial motion capture system. The state space time series for the first 35 strides were then re-sampled so that each stride was on average 100 samples in length [43], which maintained temporal variability between strides. To examine the stability of the XO, separate pLDS analyses were done using the state space constructed using 3D acceleration and angular velocities of the IMU contained in the "pelvis" part of the XO. Of note, the XO pelvis part was connected securely to the users' pelvis using straps at the waist (see Figure 1). In the case of the conventional LDS, the maximum finite-time Lyapunov exponent ($\lambda_{0.5Stride}$) was estimated as the slope of the resulting divergence curves over 0.5 strides [43], using the same 6D state space as for pLDS.

## IV. RESULTS

### A. LOWER-LIMB INTERSEGMENTAL COORDINATION

Gait speed and stride time (right heel strike to right heel strike) decreased when using the XO. Respective means (SD) for gait speed and stride time were 0.57 (0.12) m/s and 1.27 (0.07) s in the XO condition; and 1.11 (0.09) m/s and 0.54 (0.05) s in the baseline condition. Profiles of foot, shank, and thigh EAs during gait cycle were, however, comparable between the XO and baseline conditions. At the same time, the respective ranges of these EAs were narrower when using the XO (Figure 2). PCA on EA profiles showed that planarity (percentage variance explained by PC1 and PC2) was

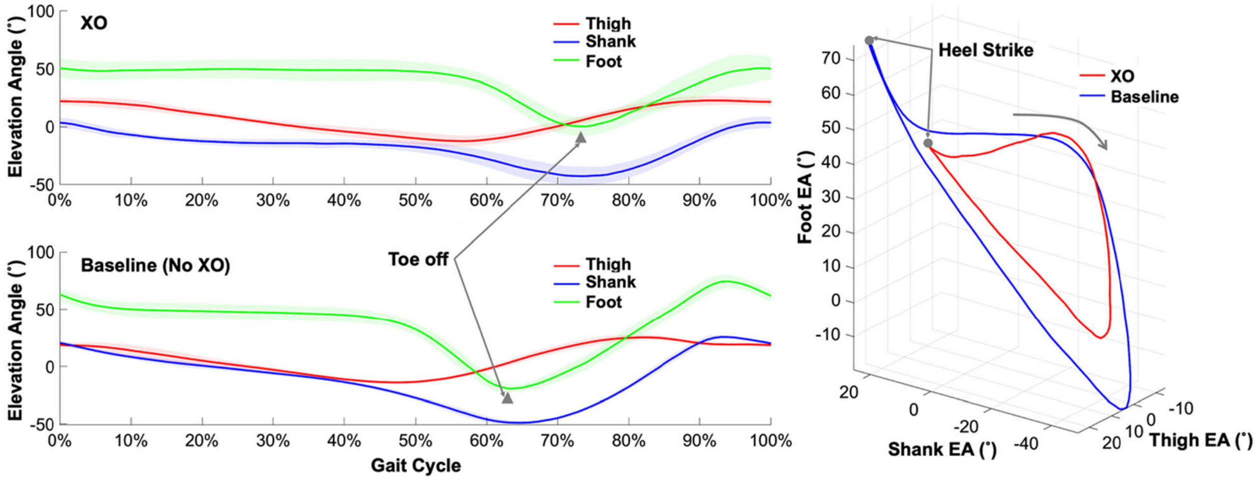


**FIGURE 2.** Ensemble averages of elevation angle (EA) profiles of the foot, shank, and thigh segments during level walking (Left), and their 3D gait (or planar covariance) loops (Right). Note that shaded regions in the left image indicate ±2 standard errors, and that the arrow in the right image indicates progression in time.

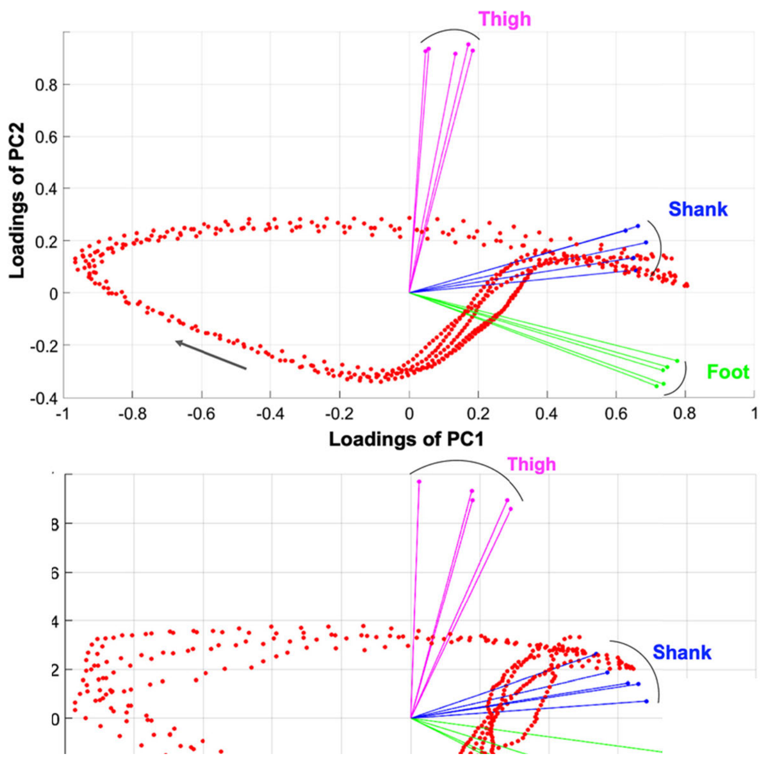


**FIGURE 3.** Biplots of foot, shank, and thigh elevation angles (EAs) for first 2 principal components (PCs) during gait cycles [Top: Baseline (i.e., no XO) condition; Bottom: XO condition]. Note that cusp points in the positive horizontal side of the figures indicate heel strike, and arrows indicate progression in time.

consistently high in both the baseline [99.2 (0.3) %] and XO [98.9 (0.5) %] conditions. Yet, covariance loop width (percentage variance explained by PC2) was larger in the XO condition [26.5 (13.1) %] vs. baseline [13.1 (1.3) %].

A biplot of PCA results on EA profiles (Figure 3) shows how each foot, shank, and thigh EA contributed to PC1 and PC2. In the baseline condition, foot, shank, and thigh EAs were loaded consistently on PC1 and PC2 across participants. Foot and shank EAs strongly influenced PC1, while thigh EAs strongly influenced PC2. Loadings of the EAs suggested high between-subjects variability in the prototype XO condition, in that these loadings were less consistently clustered for each segment compared to the baseline. Magnitudes of $u_{3t}$ (covariance plane orientation), were comparable between the baseline [0.74 (0.02)] and the XO [0.75 (0.06)] conditions.

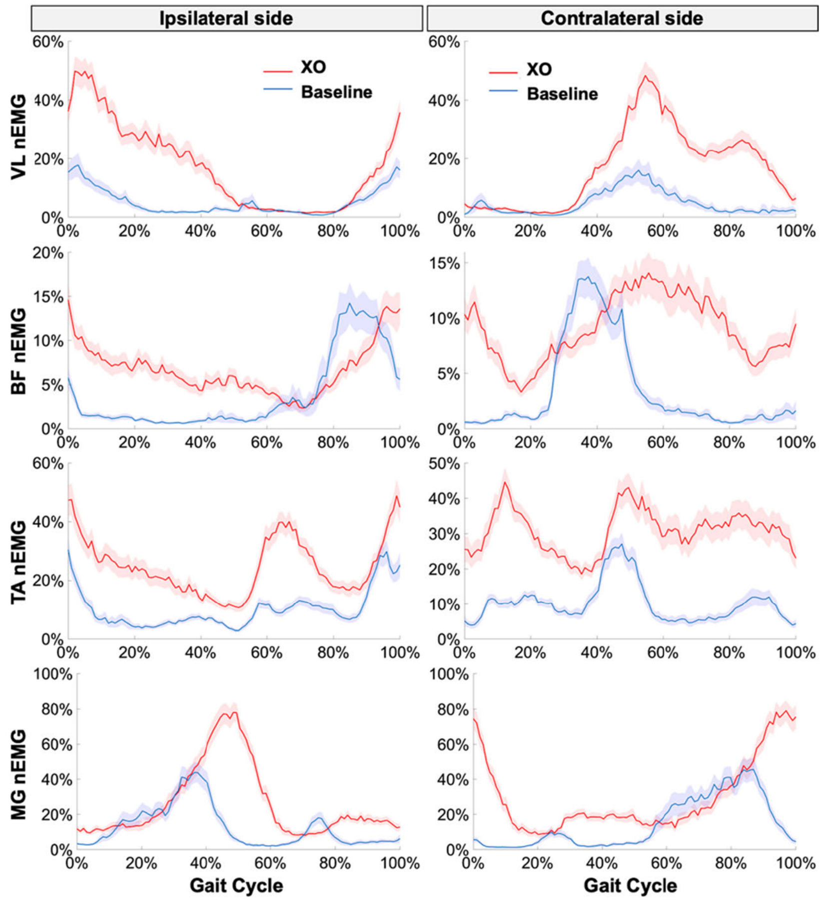


**FIGURE 4.** Normalized electromyography (nEMG) of the vastus lateralis (VL), biceps femoris (BF), tibialis anterior (TA), and medial gastrocnemius (MG) muscle on the ipsilateral (right leg here) and contralateral (left leg here) sides during level walking. Shaded regions indicate ±2 standard errors.

### B. MUSCLE ACTIVITY WITH VS. WITHOUT WB-PEXO USE

Using the prototype XO increased muscle activity levels in both the ipsilateral and contralateral legs during level walking (Figure 4). Further, NCC analyses of nEMG profiles showed that using the prototype XO generally caused phase lags in these profiles, as indicated by negative delay values at maximal $R_{XY}(\tau)$, as summarized in Table 1. There were exceptions, though, for the ipsilateral VL and TA muscles, with respective values of $\tau = 0.0$ and 5.8%. Additionally, BF nEMG profiles were least similar between XO conditions [maximal $R_{XY}(\tau) = 0.7$ for ipsilateral, and 0.73 for contralateral], compared to other nEMG profiles [maximal $R_{XY}(\tau) = 0.82$–0.92 for ipsilateral and 0.82–0.88 for contralateral muscles].

### C. LOCAL DYNAMIC STABILITY DURING LEVEL WALKING

Means of the ensemble of the logarithm of phase-dependent divergence curves for the pelvis are presented in Figure 5. These curves were obtained both with and without the XO, and using both IMU data from participants and from the pelvis component of the XO when relevant. Between XO conditions, participants exhibited a consistent pattern in $\lambda_{pLDS}$ shifts within a gait cycle, and comparable $\lambda_{pLDS}$ values. Interestingly, the phase-dependent curve obtained using the XO pelvis IMU was distinct, in that $\lambda_{pLDS}$ values approached zero after initial heel strike and remained around zero subsequently. Results from using conventional LDS showed that participants had lower $\lambda_{0.5Stride}$ values in the XO condition [0.34 (0.04)] compared to baseline (i.e., no XO) [0.74 (0.21)]. The $\lambda_{0.5Stride}$ values from the XO IMU were higher [1.10 (0.46)] than those values obtained using the participant's pelvis IMU, regardless of whether or not they were using the XO.

## V. DISCUSSION

Overall, the current results indicate that using an occupational WB-PEXO (whole-body, powered exoskeleton; pre-alpha XO® prototype) caused several changes during level walking. The extent of these changes, however, differed in terms of lower-limb segment kinematics and coordination, patterns of leg muscle activity, and local dynamic stability.

**TABLE 1.** Mean (SD) of maximal cross correlation values (maximal $R_{XY}(\tau)$) and corresponding time lags ($\tau$; % in gait cycle) for corresponding normalized EMG profiles between the baseline and XO conditions during level walking. A negative time lag value means that a profile from the XO condition is time lagged from that in the baseline condition.

| Muscle | Ipsilateral side Maximal $R_{XY}(\tau)$ | Ipsilateral side $\tau$ | Contralateral side Maximal $R_{XY}(\tau)$ | Contralateral side $\tau$ |
|---|---|---|---|---|
| **Vastus Lateralis** | 0.82 (0.08) | 0.0 (0.0) | 0.88 (0.05) | -0.8 (6.2) |
| **Biceps Femoris** | 0.70 (0.09) | -1.0 (1.4) | 0.73 (0.08) | -7.2 (22.2) |
| **Tibialis Anterior** | 0.84 (0.08) | 5.8 (13.0) | 0.85 (0.08) | -1.6 (2.3) |
| **Medial Gastrocnemius** | 0.92 (0.05) | -12.6 (4.1) | 0.82 (0.05) | -8.0 (1.2) |

### A. CHANGES IN GAIT KINEMATICS

Both with and without XO use, elevation angles (EAs) of the thigh, shank, and foot formed a loop that was close to a plane, as evidenced by high planarity values [99.2% for the baseline (i.e., no XO) and 98.9% for XO]; this outcome conforms to the planar law of intersegmental coordination [36], [46]. The shape of the gait loop was similar between the XO and baseline conditions (Figure 2). Yet, using the prototype XO shortened the length of the gait loop, as indicated by larger covariance loop widths (i.e., percentage variance explained by PC2 = 26.5% in the XO condition vs. 13.1% in the baseline condition). Given evidence that the gait loop stretches with increasing gait speed [47], [48], the shortening observed here was likely an outcome of reduced gait speed when using the prototype XO. Chow and Stokic [47] further noted that the noted stretching is associated with a larger change of foot EA over the gait cycle (i.e., steeper slope), compared to either shank and thigh EAs. Here, foot EA changed relatively less over the gait cycle when the XO was used (Figure 2), likely explaining the shortening of the gait loop. These distinct outcomes for the foot might have occurred because the ankle joint in the XO prototype was not fully actuated at the time of testing, thereby limiting ankle motion.

Loadings of foot, shank, and thigh EAs on PC1 and PC2 were less consistent across participants when using the XO (Figure 3). Furthermore, foot and thigh loadings on PC1 tended to increase when using the XO, while shank loading on PC1 decreased. Interestingly, these changes in foot, shank, and thigh loadings were also reported with increasing gait speed [47]. Ivanenko *et al.* [49] indicated that an increase of thigh loading on PC1 is associated with the role of thigh motion in regulating gait speed. When using XO, it thus appears that the hip (or thigh motion) played a critical role in regulating gait, even though gait speed was relatively slower compared with the baseline. This difference again may be attributed to the ankle actuator of the XO prototype examined. In this prototype XO, the boot (foot) was connected securely to the rigid foot component of the XO that also contains force sensing units. This interface is one pathway to provide inputs to the XO; thus, foot motion or a perturbation to the foot may cause undesirable inputs to the XO. Accordingly, users might have relied on more hip movements when using the XO.

### B. DIFFICULTY ADAPTING TO WB-PEXO USE?

Muscle activity in the ipsilateral and contralateral lower limbs generally increased with prototype XO use throughout the gait cycle (Figure 4), which suggests an increase in lower-limb joint forces. Earlier studies on powered, lower-limb (e.g., ankle, hip) orthoses/exoskeletons showed that using such a device can increase lower-limb muscle activity, particularly during an initial adaptation/learning period e.g., [50]–[52]. This earlier work may not be directly comparable to the current outcome, though, due to differences in the purpose of the exoskeletons tested (i.e., occupational vs. clinical/rehabilitation). When an exoskeleton is used, however, individuals learn to reduce their joint muscle torques to maintain total joint torque (human plus device) consistent with the biological joint torque [53], [54]. These authors and Shemmell *et al.* [55], in fact, suggested that a priority of motor planning is on maintaining joint kinetic invariance rather than joint kinematics. If so, our observed change in gait kinematics may reflect efforts to maintain joint kinetic invariance.

Our current results further indicate that the participants may have not developed a sufficient mental model or internal representation of walking with the prototype XO. A mental model here is considered as the capability to predict the forces that will be experienced during forthcoming movements and to produce proper motor commands [56], [57]. During the period of learning to walk with a lower-limb exoskeleton, individuals were found to activate their lower-limb muscles almost continuously and, in many cases, at a higher level compared to baseline (i.e., no exoskeleton) [50], [53], [58]. Such muscle recruitment may be used to increase joint and muscle impedance to facilitate increased gait stability e.g., [59], [60]. Consistent with this, we found that local dynamic stability (LDS) over strides ($\lambda_{0.5Stride}$) was lower (i.e., more stable) when using the XO [0.34 (0.04)] than not using it [0.74 (0.21)], although there was no substantial difference in phase-dependent LDS values ($\lambda_{pLDS}$; Figure 5). When using the XO, toe off occurred relatively later in the gait cycle (Figure 2), suggesting an increase in dual stance time. As such, a longer time may be required for the central nervous system to make postural adjustments for initiating a shift from double to single stance [61], perhaps to increase stability [62].

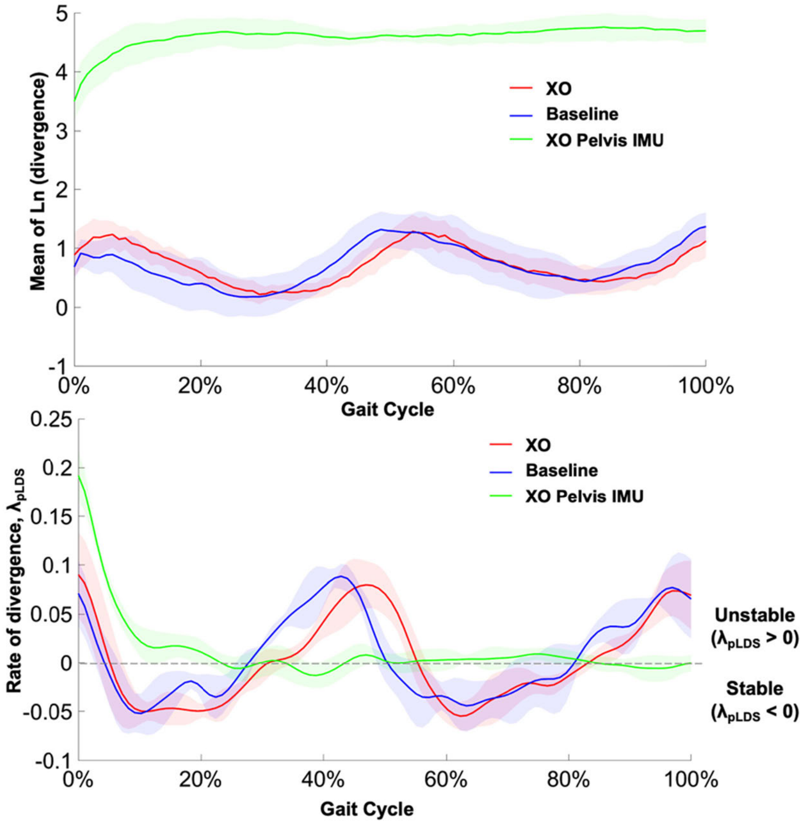


**FIGURE 5. Means of the ensemble of logarithms of phase-dependent divergence curves using pelvis IMU data (Top), and rates of divergence (Bottom). Note that XO Pelvis IMU indicates that divergence curves were obtained using XO pelvis IMU data in the XO condition, and that shaded regions indicate ±2 standard errors.**

In the aforementioned lower-limb exoskeleton studies, which included less than 1-3 days/sessions of practice (≤30 min practice in each), users adapted progressively to activate their muscles in more burst-like manner and at lower or similar levels compared to the baseline. These adaptations suggest some type of predictive representation when using an ankle (or hip) exoskeleton. Though a relatively longer practice period was given here (3-5 sessions), practice may still have not been sufficient in terms of duration and/or effectiveness for operating such a complex and high-powered exoskeleton. That a sufficient mental model was not formed with the XO supports the argument by Kao *et al.* [53], that it takes longer to adapt to walking with a robotic exoskeleton having greater mechanical strength. Indeed, the XO prototype here provided large assistive torques over multiple body joints.

### C. DIFFICULTY CONTROLLING A WB-PEXO?

Another source of the observed increase in lower-limb muscle activity may be related to the "transparency" of the prototype XO – or the capability of having no undesirable forces during human-robot interactions. The prototype XO could have caused high user-exoskeleton interaction forces/torques during the level walking trials we examined. This early XO prototype was rather heavy (110 kg), so it can be expected that any error or delay in estimating the accelerations of XO segments during movements would have caused undesirable inertial forces to be transmitted to the user. If so, the user likely needed to increase muscle activity to carry out intended/planned movements. We did not use force/pressure sensors between the user and the WB-PEXO to assess human-machine interaction forces directly. However, the normalized cross-correlation (NCC) of nEMG profiles between the XO and baseline conditions showed a moderate to high similarity on both the ipsilateral and contralateral sides (Table 1), depending on the specific muscle groups.

The NCC analyses further indicated that there were generally phase lags in nEMG profiles when using the XO (−0.8 – −12.6% of gait cycle), except for the ipsilateral vastus lateralis (0% of gait cycle) and tibialis anterior (5.8% of gait cycle). Thus, using the prototype XO seemed to affect not just lower limb EMG amplitudes, but also recruitment patterns and timing across the gait cycle. These changes may be caused by insufficient transparency of the prototype XO, and suggests that enhancing synchronicity between the XO and the user may need to be examined in future, specifically to ensure sufficient and continuous input of human intention to the XO controller. Further, the high-level controller of the prototype XO may need additional optimization, for estimating the user's state and intention (i.e., deciding when and how to deliver mechanical power).

It should be noted that earlier work has shown that maintaining balance when using a WB-PEXO can be demanding without an active control of balance [27], [30]. The XO prototype examined here did not have such a control layer at the time of testing. Thus, the observed increases in lower-limb muscle activation may be secondary to increased efforts required to maintain balance.

### D. LIMITATIONS

Several limitations were inherent to this exploratory study. First, no data were collected during practice sessions, and we thus lack information on how fast or effectively the users learned/adapted to walking with the prototype XO. In the second phase of our work, not reported here, we collected data over multiple training sessions to characterize the learning process involved when using a WB-PEXO. Second, all participants were males and rather homogeneous in physique, though females were not excluded from the study. Future work should examine a broader population with different sex, age, and physical characteristics. Third, examining correlations between human and XO joint torques would help understand similarities in the timing and magnitude of these kinetic aspects, and help identify the relative contributions of the XO vs. human in level walking (and other tasks). Third, human-exoskeleton interaction forces were not measured. Given that the participants here had rather homogeneous anthropometry, monitoring such interaction forces could help to better assess individual differences in XO learning processes. Finally, other tasks should be considered. Learning how the XO benefits in situations involving high loads (for which the XO is primarily designed), will help determine effective use-cases and promote industrial adoption.

## VI. CONCLUSION AND FUTURE WORK

We reported here results from human-in-the-loop testing of a state-of-the-art, occupational WB-PEXO prototype in the early stages of development, in the context of level walking. The prototype can enable the user to handle large loads without matching physical efforts. Recently, it was demonstrated that users can carry a 50 kg payload with the XO prototype, while feeling that they are handling a 6 kg load [63]. Although WB-PEXO use can substantially augment the physical strength of a user, our results suggest that the current early prototype device may need to be further optimized so that it does not present a challenge to the user during basic, functional activities requiring no (or minimal) external load handling. Specifically, users adopted strategies during level walking that involved thigh motions to regulate gait, though these strategies varied between users. Observed increases in muscle activity levels, along with changes in muscle activation patterns during the gait cycle, suggest that users may have not developed a sufficient mental model of walking with the WB-PEXO.

Completing the current independent, human-in-the-loop testing allowed for several generalizable insights into occupational WB-PEXO development. First, there likely is a tradeoff between the strength capacity and nimbleness of a user with the current generation of WB-PEXOs. Such a tradeoff may be reasonable and practical, though, depending on use cases. A better understanding of this tradeoff needs to be developed with respect to task characteristics (e.g., load handling frequency, walking distance, a range of activities involved), however, to enable optimal tuning of a WB-PEXO for a specific application or optimal task selection for a given WB-PEXO. Second, there is a need to enhance our understanding of the learning processes that occur with WB-PEXO use. Controlling a whole-body system can affect the user at various levels, from perceptions to actions. The extent of such effects will depend on WB-PEXO human interface designs (e.g., force-sensing between a WB-PEXO and its user), controllers (e.g., understanding user intention, intuitiveness in control), and overall human-robot interaction experiences. An improved understanding of the learning processes with WB-PEXO use will enable developing effective training processes and programs for a given WB-PEXO. Third, there is a need for reliable and effective methods to communicate intentions between a WB-PEXO and its user. Ideally, a WB-PEXO should detect, decode, and infer a user's intentions to provide an effective level of augmentation. To effectively develop a mental model for WB-PEXO use, a user also needs to be able to understand the "intentions" of the WB-PEXO through, for example, consistent and expected behaviors of the WB-PEXO and status/state information on user interfaces. Similarly, efforts are needed to increase the transparency of a WB-PEXO, to remove or minimize undesired forces during human-robot interactions. Lastly, we believe that early, third-party, human-in-the-loop testing and information sharing should be encouraged to facilitate the development of an effective, occupational WB-PEXO, and to help industries and regulatory bodies prepare for forthcoming generations of WB-PEXO use.

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## REFERENCES

[1] S. Bevan, "Economic impact of musculoskeletal disorders (MSDs) on work in Europe," *Best Pract. Res. Clin. Rheumatology*, vol. 29, no. 3, pp. 356–373, Jun. 2015.

[2] T. R. Waters, "National efforts to identify research issues related to prevention of work-related musculoskeletal disorders," *J. Electromyogr. Kinesiol.*, vol. 14, no. 1, pp. 7–12, Feb. 2004.

[3] M. P. de Looze, T. Bosch, F. Krause, K. S. Stadler, and L. W. O'Sullivan, ''Exoskeletons for industrial application and their potential effects on physical work load,'' *Ergonomics*, vol. 59, no. 5, pp. 671–681, May 2016.
[4] D. Romero, P. Bernus, O. Noran, J. Stahre, and Å. Fast-Berglund, ''The operator 4.0: Human cyber-physical systems & adaptive automation towards human-automation symbiosis work systems,'' in *Proc. Adv. Prod. Manage. Syst. Initiatives Sustain. World*, 2016, pp. 677–686.
[5] S. Kim, M. A. Nussbaum, M. I. M. Esfahani, M. M. Alemi, S. Alabdulkarim, and E. Rahedi, ''Assessin the influence of a passive, upper extremity exoskeletal vest for tasks requiring arm elevation: Part I—'Expected' effects on discomfort, shoulder muscle activity, and work task performance,'' *Appl. Ergonom.*, vol. 70, pp. 315–322, Jul. 2018.
[6] S. Spada, L. Ghibaudo, C. Carnazzo, M. Di Pardo, D. S. Chander, L. Gastaldi, and M. P. Cavatorta, ''Physical and virtual assessment of a passive exoskeleton,'' in *Proc. 20th Congr. Int. Ergonom. Assoc. (IEA)*, 2019, pp. 247–257.
[7] S. Madinei, M. M. Alemi, S. Kim, D. Srinivasan, and M. A. Nussbaum, ''Biomechanical evaluation of passive back-support exoskeletons in a precision manual assembly task: 'Expected' effects on trunk muscle activity, perceived exertion, and task performance,'' *Hum. Factors J. Hum. Factors Ergonom. Soc.*, vol. 62, no. 3, pp. 441–457, May 2020.
[8] T. Bosch, J. van Eck, K. Knitel, and M. de Looze, ''The effects of a passive exoskeleton on muscle activity, discomfort and endurance time in forward bending work,'' *Appl. Ergonom.*, vol. 54, pp. 212–217, May 2016.
[9] A. S. Koopman, M. Näf, S. J. Baltrusch, I. Kingma, C. Rodriguez-Guerrero, J. Babič, M. P. de Looze, and J. H. van Dieën, ''Biomechanical evaluation of a new passive back support exoskeleton,'' *J. Biomech.*, vol. 105, May 2020, Art. no. 109795.
[10] M. M. Alemi, S. Madinei, S. Kim, D. Srinivasan, and M. A. Nussbaum, ''Effects of two passive back-support exoskeletons on muscle activity, energy expenditure, and subjective assessments during repetitive lifting,'' *Hum. Factors J. Hum. Factors Ergonom. Soc.*, vol. 62, no. 3, pp. 458–474, May 2020.
[11] A. Krok. (Nov. 9, 2017). Ford's Exoskeleton Could Help Factory Workers in a Big Way. Roadshow. Accessed: Feb. 20, 2020. [Online]. Available: https://www.cnet.com/roadshow/news/fords-exoskeleton-could-help-factory-workers-in-a-big-way/
[12] A. M. Selko. (Apr. 11, 2019). SLC: 2019 Toyota's Finding Success With Exoskeletons. EHStoday. Accessed: Feb. 17, 2020. [Online]. Available: https://www.ehstoday.com/safety-leadership-conference/article/21920474/slc-2019-toyotas-finding-success-with-exoskeletons
[13] D. Wren. (Aug. 19, 2018). Boeing Co. Studying Wearable Exoskeletons That Can Reduce Workplace Injuries. Post and Courier. Accessed: Feb. 20, 2020. [Online]. Available: https://www.postandcourier.com/business/boeing-co-studying-wearable-exoskeletons-that-can-reduce-workplace-injuries/article_efc6bbe0-a15d-11e8-a8c4-737ba6ec5b2e.html
[14] S. Toxiri, A. S. Koopman, M. Lazzaroni, J. Ortiz, V. Power, M. P. de Looze, L. O'Sullivan, and D. G. Caldwell, ''Rationale, implementation and evaluation of assistive strategies for an active back-support exoskeleton,'' *Frontiers Robot. AI*, vol. 5, p. 53, May 2018.
[15] T. Zhang and H. Huang, ''A lower-back robotic exoskeleton: Industrial handling augmentation used to provide spinal support,'' *IEEE Robot. Autom. Mag.*, vol. 25, no. 2, pp. 95–106, Jun. 2018.
[16] T. Wang, Y. Zhu, T. Zheng, D. Sui, S. Zhao, and J. Zhao, ''PALExo: A parallel actuated lower limb exoskeleton for high-load carrying,'' *IEEE Access*, vol. 8, pp. 67250–67262, 2020.
[17] B. S. Rupal, S. Rafique, A. Singla, E. Singla, M. Isaksson, and G. S. Virk, ''Lower-limb exoskeletons: Research trends and regulatory guidelines in medical and non-medical applications,'' *Int. J. Adv. Robot. Syst.*, vol. 14, no. 6, 2017, Art. no. 1729881417743554.
[18] H.-D. Lee, B.-K. Lee, W.-S. Kim, J.-S. Han, K.-S. Shin, and C.-S. Han, ''Human–robot cooperation control based on a dynamic model of an upper limb exoskeleton for human power amplification,'' *Mechatronics*, vol. 24, no. 2, pp. 168–176, Mar. 2014.
[19] B. M. Otten, R. Weidner, and A. Argubi-Wollesen, ''Evaluation of a novel active exoskeleton for tasks at or above head level,'' *IEEE Robot. Autom. Lett.*, vol. 3, no. 3, pp. 2408–2415, Jul. 2018.
[20] K. Miura, H. Kadone, M. Koda, T. Abe, H. Kumagai, K. Nagashima, K. Mataki, K. Fujii, H. Noguchi, T. Funayama, H. Kawamoto, Y. Sankai, and M. Yamazaki, ''The hybrid assistive limb (HAL) for care support successfully reduced lumbar load in repetitive lifting movements,'' *J. Clin. Neurosci.*, vol. 53, pp. 276–279, Jul. 2018.
[21] A. von Glinski, E. Yilmaz, S. Mrotzek, E. Marek, B. Jettkant, A. Brinkemper, C. Fisahn, T. A. Schildhauer, and J. Geßmann, ''Effectiveness of an on-body lifting aid (HAL for care support) to reduce lower back muscle activity during repetitive lifting tasks,'' *J. Clin. Neurosci.*, vol. 63, pp. 249–255, May 2019.
[22] K. Miura, H. Kadone, T. Abe, M. Koda, T. Funayama, H. Noguchi, H. Kumagai, K. Nagashima, K. Mataki, Y. Shibao, K. Sato, H. Kawamoto, Y. Sankai, and M. Yamazaki, ''Successful use of the hybrid assistive limb for care support to reduce lumbar load in a simulated patient transfer,'' *Asian Spine J.*, vol. 15, no. 1, pp. 40–45, Feb. 2021, doi: 10.31616/asj.2019.0111.
[23] L. C. Paul, ''Mobile space suit,'' U.S. Patent 3 034 131, May 15, 1962.
[24] N. J. Mizen, ''Powered exoskeletal apparatus for amplifying human strength in response to normal body movements,'' U.S. Patent 3 449 769, Jun. 17, 1969.
[25] B. R. Fick and J. B. Makinson, ''Hardiman I prototype for machine augmentation of human strength and endurance,'' Eng. Psychol. Programs Office Army Mobility Equip. Res. Develop. Center, Fort Belvoir, VA, USA, Tech. Rep. N00014-66-C0051, 1971.
[26] T. J. Snyder and H. Kazerooni, ''A novel material handling system,'' in *Proc. IEEE Int. Conf. Robot. Autom.*, vol. 2, Apr. 1996, pp. 1147–1152.
[27] M. Fontana, R. Vertechy, S. Marcheschi, F. Salsedo, and M. Bergamasco, ''The body extender: A full-body exoskeleton for the transport and handling of heavy loads,'' *IEEE Robot. Autom. Mag.*, vol. 21, no. 4, pp. 34–44, Dec. 2014.
[28] Y. Sankai, ''HAL: Hybrid assistive limb based on cybernics,'' in *Robotics Research* (Springer Tracts in Advanced Robotics), vol. 66, M. Kaneko and Y. Nakamura, Eds. Berlin, Germany: Springer, 2010, doi: 10.1007/978-3-642-14743-2_3.
[29] T. Ishida, T. Kiyama, K. Osuka, G. Shirogauchi, R. Oya, and H. Fujimoto, ''Movement analysis of power-assistive machinery with high strength-amplification,'' in *Proc. SICE Annu. Conf.*, Taipei, Taiwan, 2010, pp. 2022–2025.
[30] G. Chu, J. Hong, D.-H. Jeong, D. Kim, S. Kim, S. Jeong, and J. Choo, ''The experiments of wearable robot for carrying heavy-weight objects of shipbuilding works,'' in *Proc. IEEE Int. Conf. Autom. Sci. Eng. (CASE)*, Taipei, Taiwan, Aug. 2014, pp. 978–983.
[31] S. C. Jacobsen, M. X. Olivier, and B. J. Maclean, ''Control logic for biomimetic joint actuators,'' U.S. Patent 8 731 716, May 20, 2014.
[32] K. Laursen and A. J. Salter, ''The paradox of openness: Appropriability, external search and collaboration,'' *Res. Policy*, vol. 43, no. 5, pp. 867–878, Jun. 2014.
[33] J. West and M. Bogers, ''Leveraging external sources of innovation: A review of research on open innovation,'' *J. Product Innov. Manage.*, vol. 31, no. 4, pp. 814–831, Jul. 2014.
[34] E. Criswell, *Cram's Introduction to Surface Electromyography*. Boston, MA, USA: Jones & Bartlett, 2010.
[35] J. A. Zeni, Jr., J. G. Richards, and J. S. Higginson, ''Two simple methods for determining gait events during treadmill and overground walking using kinematic data,'' *Gait Posture*, vol. 27, no. 4, pp. 710–714, May 2008.
[36] N. A. Borghese, L. Bianchi, and F. Lacquaniti, ''Kinematic determinants of human locomotion,'' *J. Physiol.*, vol. 494, no. 3, pp. 863–879, Aug. 1996.
[37] Y. P. Ivanenko, G. Cappellini, N. Dominici, R. E. Poppele, and F. Lacquaniti, ''Modular control of limb movements during human locomotion,'' *J. Neurosci.*, vol. 27, no. 41, pp. 11149–11161, Oct. 2007.
[38] A. Barliya, L. Omlor, M. A. Giese, and T. Flash, ''An analytical formulation of the law of intersegmental coordination during human locomotion,'' *Exp. Brain Res.*, vol. 193, no. 3, pp. 371–385, Mar. 2009.
[39] L. Bianchi, D. Angelini, and F. Lacquaniti, ''Individual characteristics of human walking mechanics,'' *Pflgers Archiv Eur. J. Physiol.*, vol. 436, no. 3, pp. 343–356, Jun. 1998.
[40] C. Chatfield, *The Analysis of Time Series: An Introduction*, 4th ed. London, U.K.: Chapman & Hall, 1989.
[41] E. A. F. Ihlen, T. Goihl, P. B. Wik, O. Sletvold, J. Helbostad, and B. Vereijken, ''Phase-dependent changes in local dynamic stability of human gait,'' *J. Biomech.*, vol. 45, no. 13, pp. 2208–2214, Aug. 2012.
[42] A. Mahmoudian, S. M. Bruijn, H. R. F. Yakhdani, O. G. Meijer, S. M. P. Verschueren, and J. H. van Dieen, ''Phase-dependent changes in local dynamic stability during walking in elderly with and without knee osteoarthritis,'' *J. Biomech.*, vol. 49, no. 1, pp. 80–86, Jan. 2016.
[43] S. M. Bruijn, J. H. van Dieën, O. G. Meijer, and P. J. Beek, ''Statistical precision and sensitivity of measures of dynamic gait stability,'' *J. Neurosci. Methods*, vol. 178, no. 2, pp. 327–333, Apr. 2009.

[44] P. C. Fino, M. Mancini, C. Curtze, J. G. Nutt, and F. B. Horak, "Gait stability has phase-dependent dual-task costs in Parkinson's disease," *Frontiers Neurol.*, vol. 9, p. 373, May 2018.
[45] M. T. Rosenstein, J. J. Collins, and C. J. De Luca, "A practical method for calculating largest Lyapunov exponents from small data sets," *Phys. D, Nonlinear Phenomena*, vol. 65, nos. 1–2, pp. 117–134, May 1993.
[46] Y. P. Ivanenko, A. d'Avella, R. E. Poppele, and F. Lacquaniti, "On the origin of planar covariation of elevation angles during human locomotion," *J. Neurophysiol.*, vol. 99, no. 4, pp. 1890–1898, Apr. 2008.
[47] J. W. Chow and D. S. Stokic, "Intersegmental coordination scales with gait speed similarly in men and women," *Exp. Brain Res.*, vol. 233, no. 11, pp. 3175–3185, Nov. 2015.
[48] C. Bleyenheuft and C. Detrembleur, "Kinematic covariation in pediatric, adult and elderly subjects: Is gait control influenced by age?" *Clin. Biomech.*, vol. 27, no. 6, pp. 568–572, Jul. 2012.
[49] Y. P. Ivanenko, R. E. Poppele, and F. Lacquaniti, "Five basic muscle activation patterns account for muscle activity during human locomotion," *J. Physiol.*, vol. 556, no. 1, pp. 267–282, Apr. 2004.
[50] K. E. Gordon and D. P. Ferris, "Learning to walk with a robotic ankle exoskeleton," *J. Biomech.*, vol. 40, no. 12, pp. 2636–2644, Jan. 2007.
[51] G. S. Sawicki and D. P. Ferris, "Powered ankle exoskeletons reveal the metabolic cost of plantar flexor mechanical work during walking with longer steps at constant step frequency," *J. Exp. Biol.*, vol. 212, no. 1, pp. 21–31, Jan. 2009.
[52] S. Galle, P. Malcolm, W. Derave, and D. De Clercq, "Adaptation to walking with an exoskeleton that assists ankle extension," *Gait Posture*, vol. 38, no. 3, pp. 495–499, Jul. 2013.
[53] P.-C. Kao, C. L. Lewis, and D. P. Ferris, "Invariant ankle moment patterns when walking with and without a robotic ankle exoskeleton," *J. Biomech.*, vol. 43, no. 2, pp. 203–209, Jan. 2010.
[54] C. L. Lewis and D. P. Ferris, "Invariant hip moment pattern while walking with a robotic hip exoskeleton," *J. Biomech.*, vol. 44, no. 5, pp. 789–793, Mar. 2011.
[55] J. Shemmell, J. Johansson, V. Portra, G. L. Gottlieb, J. S. Thomas, and D. M. Corcos, "Control of interjoint coordination during the swing phase of normal gait at different speeds," *J. NeuroEng. Rehabil.*, vol. 4, no. 1, p. 10, Apr. 2007.
[56] R. Shadmehr and F. Mussa-Ivaldi, "Adaptive representation of dynamics during learning of a motor task," *J. Neurosci.*, vol. 14, no. 5, pp. 3208–3224, May 1994.
[57] M. Kawato, "Internal models for motor control and trajectory planning," *Current Opinion Neurobiol.*, vol. 9, no. 6, pp. 718–727, Dec. 1999.
[58] T. Lenzi, M. C. Carrozza, and S. K. Agrawal, "Powered hip exoskeletons can reduce the User's hip and ankle muscle activations during walking," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 21, no. 6, pp. 938–948, Nov. 2013.
[59] A. J. van Soest, W. P. Haenen, and L. A. Rozendaal, "Stability of bipedal stance: The contribution of cocontraction and spindle feedback," *Biol. Cybern.*, vol. 88, no. 4, pp. 293–301, Apr. 2003.
[60] X. H. Duan, R. H. Allen, and J. Q. Sun, "A stiffness-varying model of human gait," *Med. Eng. Phys.*, vol. 19, no. 6, pp. 518–524, Sep. 1997.
[61] M. C. Do, Y. Breniere, and P. Brenguier, "A biomechanical study of balance recovery during the fall forward," *J. Biomech.*, vol. 15, no. 12, pp. 933–939, Jan. 1982.
[62] J.-Y. You, Y.-L. Chou, C.-J. Lin, and F.-C. Su, "Effect of slip on movement of body center of mass relative to base of support," *Clin. Biomech.*, vol. 16, no. 2, pp. 167–173, Feb. 2001.
[63] E. Ackerman. (Dec. 10, 2019). Sarcos Demonstrates Powered Exosuit That Gives Workers Super Strength. IEEE Spectrum. Accessed: Aug. 10, 2020. [Online]. Available: https://spectrum.ieee.org/automaton/robotics/industrial-robots/sarcos-guardian-xo-powered-exoskeleton

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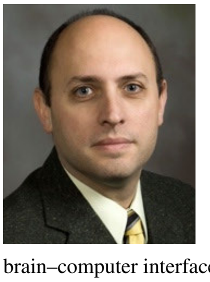

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