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# Kinematic effects of a back-assistance exoskeleton during human locomotion

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ABSTRACT

In the last years, Industry 5.0 has proposed a sustainable and resilient industry model, where the human-centric approach places human needs at the center of the production process. Wearable robots have been designed to assist users, providing support for the entire body or specific regions during task performance. Ergonomic investigations are necessary to test the effects, advantages and possible drawbacks of occupational wearable devices. The present study focuses on the biomechanics of locomotion while wearing the Laevo V2.5 exoskeleton. Experimental tests involved twelve healthy volunteers. Spatio-temporal parameters, human 3D kinematics and exoskeleton 3D kinematics were compared in three settings (without exoskeleton, wearing the exoskeleton without and with passive support). These comparisons aimed to quantify the effects and the possible restrictions on user kinematics due to the interaction with the exoskeleton. Results highlighted a significant reduction in the gait speed (1.14 m/s no-exo, 1.07 m/s exo-no-support, 1.05 m/s exo-with-support) and the stride length (1.29 m no-exo, 1.24 m exo-no-support, 1.23 m exo-with-support) when wearing the exoskeleton. Human angular kinematics showed significant reductions in the range of motion for all joints when wearing the exoskeleton. However, results pointed out no significant differences between the no-support and support configurations, indicating that the primary effect is due to the exoskeleton structure rather than the support provided. Further assessment is essential to determine whether these changes in human kinematics align with ergonomic standards and reflect user adaptation, or if they fulfill acceptable limits, potentially leading to long-term negative effects.

## 1. Introduction

Recent European statistics and surveys have identified muscular disorders (MSDs) as the most prevalent work-related health issue. Among Europeans, 60% reported experiencing MSDs, with muscle pain in the lower back and upper limbs being the most common complaints (De et al., 2019). To reduce injury risk, various strategies have been proposed, including the use of ergonomic equipment, encouraging breaks and task rotation, and promoting standards and training to achieve correct posture (Xu et al., 2021). With the advent of Industry 4.0, the idea of fully automating work tasks was introduced as a potential solution. However, the complete replacement of human workers with advanced robotic technology is not always feasible (Xu et al., 2021). In response, Industry 5.0 complements the existing Industry 4.0 model by promoting a sustainable, human-centric and resilient European industry (Breque et al., 2021). The human-centric approach prioritizes human needs and interests shifting the focus from purely technology-driven progress to a production model centered on societal and human well-being (Nahavandi, 2019). Ensuring a safe and inclusive work environment has therefore become a priority to support the physical and mental health, well-being, and fundamental rights of workers.

In recent years, wearable robots have emerged as assisting devices within this context. Among these, exoskeletons are designed to support either the entire body or specific body regions (i.e. back, shoulders, arms and legs) of the user during working tasks (De Bock et al., 2022). These devices supply mechanical power to one or more human joints to reduce biomechanical loads and efforts during specific movements. Exoskeleton designs vary primarily by the mechanical structures (rigid, soft, hybrid), the body part they support and the mechanism of assistance (passive,

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active, hybrid) (Bosch et al., 2016; Zhu et al., 2021). For occupational exoskeletons that support the human back during lifting and carrying tasks, several commercial solutions and research prototypes have been developed, featuring both active (https://www.germanbionic.com/crayx/, accessed on 2021) and passive (https://www.suitx.com/backx, accessed 2022; http://en.laevo.nl/, accessed 2022) assistance mechanisms. However, extensive evaluations are essential to assess both the effectiveness of exoskeletons and their impact on daily work activities. While the effectiveness of exoskeletons has been evaluated in several studies, few researchers have addressed the second aspect. Therefore, the aim of the present study is to assess the impact of a commercial back-support exoskeleton on walking, an activity associated with almost all work tasks.

Kermavnar and colleagues (Kermavnar et al., 2021) presented a systematic review summarizing recent findings on objective and subjective evaluations of industrial back-support exoskeletons, while methodologies evaluating technical, physical and subjective aspects of various occupational exoskeletons have been deeply discussed in (Hoffmann et al., 2022). Subjective evaluation is a critical measure for assessing the exoskeleton effectiveness. A recent literature review (Elprama et al., 2022) examined empirical studies focusing on the user experience when wearing a passive exoskeleton, highlighting a lack of standardization. The review suggested that exoskeleton design, development, and evaluation should focus on the end-user's perspective rather than only on technical metrics. Therefore, identifying factors that influence user acceptance is essential for designers, researchers, and companies. Hensel (Hensel and Keil, 2019) tested the impact of the passive low-back support exoskeleton Laevo in a four-week field study with automobile manufacturing workers. Users reported reduced physical discomfort in the lower back and improved load distribution, but also noted discomfort from wearing the device, which strongly affected user acceptance. Similar feedback was reported by Schwerha and colleagues (Schwerha et al., 2022), who analyzed different upper-limb and trunk exoskeletons during static and dynamic manufacturing tasks. Despite involving real working tasks and testing social impacts, the absence of objective measures may critically affect the standardization and repeatability of exoskeleton assessments.

To address the limitation of subjective evaluation, several studies have proposed biomechanical validation of exoskeletons in laboratory settings (Huysamen et al., 2018; Alemi et al., 2020; Baltrusch et al., 2019; Kozinc et al., 2020). Huysamen et al. (2018) and Alemi et al. (2020) analyzed repetitive lifting and lowering tasks, while Baltrusch and colleagues (Baltrusch et al., 2019; Kozinc et al., 2020) developed a twelve-task assessment battery to evaluate functional performance and subjective outcomes. Tasks where the exoskeleton assistance is expected to be beneficial (lifting, static forward bending), where it might interfere (wide stance, squatting, trunk rotation) or potentially hinder the performance (stair climbing, sit to stand) were included. Recently, Zelik et al. (2022) introduced Exo-LiFFT, an ergonomic assessment tool that predicts injury risk and evaluates lower-back support biomechanics, focusing on movement repetition and peak back extension moments. Considering in field testing, Marino (2019) objectively evaluated two exoskeletons using wearable inertial sensors to assess step rate among 14 workers while stocking and in tire installation tasks. However, integrating motion capture systems in real work settings is challenging, often due to the complexity of setup and procedures (Crea et al., 2021), making laboratory analysis ideal for an in-depth study of user-exoskeleton interaction.

Experimental studies have evaluated exoskeleton performance through various parameters, including back muscle activation (Madinei et al., 2020; Luger et al., 2021a, 2021b; So et al., 2022; Simon et al., 2021; Hwang et al., 2021; Schmalz et al., 2022), metabolic cost (Schmalz et al., 2022) and heart rate (Luger et al., 2021a). Reduced back muscle activity is a key indicator of effective support, with notable reductions observed in electromyography measurements during lifting (Simon et al., 2021), carrying (So et al., 2022) and functional tasks (Luger et al., 2021b). Several studies, discussed in (Kermavnar et al., 2021), reported significant reductions in erector spinae activity with exoskeleton use, as well as muscle activity changes in the shoulder (Hwang et al., 2021), abdomen, and thighs (Schmalz et al., 2022). Pesenti et al. (2021) proposed a validation framework for industrial low-back exoskeletons, grouping evaluation metrics into muscular, force/torque, metabolic, functional, and subjective domains. In the functional domain, the primary metric is kinematic analysis, focusing on potential range-of-motion restrictions when wearing the exoskeleton.

Research on repetitive lifting tasks has highlighted the effect of exoskeletons on human posture (Madinei et al., 2020; Luger et al., 2021a, 2021b; So et al., 2022; Simon et al., 2021; Hwang et al., 2021; Schmalz et al., 2022). Results reported significant increase of hip, knee, ankle ranges of motion in the sagittal plane when wearing the exoskeleton (Luger et al., 2021b; Simon et al., 2021). These results are in line with the study presented by Ulrey and Fathallah (Ulrey et al., 2013), but other experimental tests reported contrasting results (Näf et al., 2018) or no effect (Baltrusch et al., 2020). Despite conflicting results, it appears that the use of back-supporting exoskeletons results in postural changes, although the long-term beneficial effects on lumbar loads are still unclear. Additionally, Madinei and colleagues (Madinei et al., 2020) found minimal change in back kinematics with Laevo and backX$^{MT}$ exoskeletons. Other studies showed effects on lumbar lordosis and thoracic kyphosis while maintaining ergonomic postures (Luger et al., 2021a). Similarly, the biomechanical assessment of the Muscle Suite Every (Innophys, McKibben artificial muscle) exoskeleton supported the hypothesis that the device could benefit users in lifting and carrying tasks by decreasing the trunk flexion (So et al., 2022). Poliero et al. (2021) considered the XoTrunk under three conditions: without the exoskeleton, wearing the exoskeleton switching between assistance and transparency based on activity recognition, and with the exoskeleton constantly providing assistance. Findings showed increased task completion time, increased trunk range of motion (ROM), and reduced hip and knee ROMs in the sagittal plane, although only sagittal plane kinematics was analyzed. Park and colleagues (Park et al., 2022a, 2022b) explored the passive exoskeleton backX™ in terms of gait performance, kinematics and dynamics during level walking. Experimental tests were conducted on 20 young and healthy participants and trials were repeated across four configurations (no exoskeleton, exoskeleton without assistance, low assistance, and high assistance). Results indicated significant effects on gait parameters and joint ROM measures, including reductions in step length and gait cycle time, an increase in step width, and a reduction in hip ROMs in the sagittal plane, but transverse and frontal plane ROMs were not considered.

Due to the still open questions about user-exoskeleton interaction, limitations in sagittal plane measurements, and mixed findings from previous studies, it is essential to investigate and quantify the 3D biomechanical effects of wearable back-assistance devices during a range of activities beyond the specific tasks for which they were designed. The present study presents a 3D biomechanical investigation of the Laevo V2.5 back-assistance exoskeleton during human locomotion. Experimental tests were conducted across three settings: i) without the exoskeleton, wearing exoskeleton with ii) deactivated and iii) activated assistance. Spatio-temporal parameters, human 3D kinematics and exoskeleton 3D kinematics were compared across the three settings to quantify the effects and potential limitations posed by both the exoskeleton's structure and its assistance functions on human movement.

## 2. Material and methods

Experimental tests were conducted on healthy young volunteers, across three different configurations: without the device (no-exo), with the exoskeleton but without the assistance (exo-no-support) and with the exoskeleton providing assistance (exo-with-support). Participants performed gait trials and both human and exoskeleton kinematics were considered as parameters of interest.

### 2.1. Participants

Twelve healthy young volunteers (4 male and 8 female; age: 24.5 ± 1.1 years; body mass: 61.3 ± 9.4 kg; height: 167.0 ± 4.2 cm) participated in the study. Subjects with any self-reported musculoskeletal disorders or back pains were excluded. The aim of the study, the experimental procedure and the main functions of the wearable device were clearly explained to each participant. All procedures conformed to the Helsinki Declaration and written informed consent was obtained from all participants.

### 2.2. Instrumentation

The experimental tests were conducted indoor at the Polito$^{BIO}$Med Lab of Politecnico di Torino (Torino, Italy). The total volume of the laboratory was 12 m in length and 5.5 m in width. Experimental data recording involved the following instruments.

- 3 Vicon VUE video-cameras for video recording (sample rate of 50 Hz);
- 12 infrared-cameras Vicon Vero (sample rate of 100 Hz) for capturing 3D;
- 39 passive markers (14 mm diameter) positioned on anatomical landmarks of the head, torso, upper limbs, pelvis and lower limbs of the subject for the Plug-in-Gait (https://www.vicon.com/) full body human model;
- 17 passive markers (14 mm diameter) for the customized exoskeleton model (Panero et al., 2021).

Fig. 1 depicts front (A) and back (B) views of the subject outfitted with passive markers, which enable the reconstruction of the full-body Plug-in-Gait model (C).

### 2.3. Passive exoskeleton

The commercial trunk-exoskeleton Laevo V2.5 (Laevo, Netherlands) (http://en.laevo.nl/, accessed 2022; Panero et al., 2021) was selected for this experimental study. The Laevo V2.5 is a passive wearable device, with a total mass of 2.8 kg, designed to support the user's back and reduce effort during specific motion tasks, such as dynamic trunk flexion/extension and maintaining static trunk-flexed posture. Its structure consists of three main parts: i) trunk structure characterized by one butterfly pad (trunk pad) and two rigid bars, designed to support the user's chest, ii) pelvis belt that secure the exoskeleton to the user's pelvis, providing a stable connection point, iii) thigh structure characterized by two rigid bars and thigh pads. The pelvis belt is linked to the thigh structure by two free hinge joints. The trunk and thigh structures are connected by two passive joints that house a cam-spring mechanism (gas spring) which supplies assistance torque. This torque is modulated according to the relative angle between the trunk and thigh parts in the sagittal plane and is applied to the user's body through the trunk and thigh pads. A mechanical latch is embedded in the joint, which can be activated manually to disengage the spring action, thereby deactivating support to the trunk. Additionally, users can adjust the initial angle for spring engagement (from 0° to 35°, in 5° increments) to suit individual anthropometric needs and preferences. The design of the trunk pad features a combination of degrees of freedom, allowing the exoskeleton to adapt to trunk movements in the frontal and transverse planes and enhancing user comfort. Both the trunk and the thigh parts allow the translation of the pads parallel to the longitudinal axis of the human body segments, further accommodating natural movement and comfort.

A customized model of the exoskeleton was developed using Vicon ProCalc software (Panero et al., 2021). Clusters of three passive markers were positioned in correspondence of the trunk pad (TRK cluster) and at each of the two pelvis plates (RPE1, RPE2, RJNT - LPE1, LPE2, LJNT). Additionally, four markers were placed on the two thigh pads (RTH and LTH clusters), for a total of 17 markers. Fig. 2 shows the frontal (A) and back (B) views of the subject wearing the exoskeleton and the passive markers used for the reconstruction of the exoskeleton model. Fig. 2C depicts the customized kinematic model of the exoskeleton.

For each set of markers, a rigid segment was identified and a local three-axis coordinate system was established. In this system, z-axis points upward (blue axis), x-axis points forward (red axis) and y-axis points to the left side (green axis). YXZ Euler angles were defined to assess the relative angles of each distal segment with respect to the adjacent proximal one and the angles of each segment with respect to the global coordinate system of the laboratory.

### 2.4. Protocol

During the donning phase of the exoskeleton, each participant self-selected the initial angle of spring operation. The subject was instructed to wear the exoskeleton and to assume a standing posture with straight back. Starting from 0°, the initial angle of spring operation was

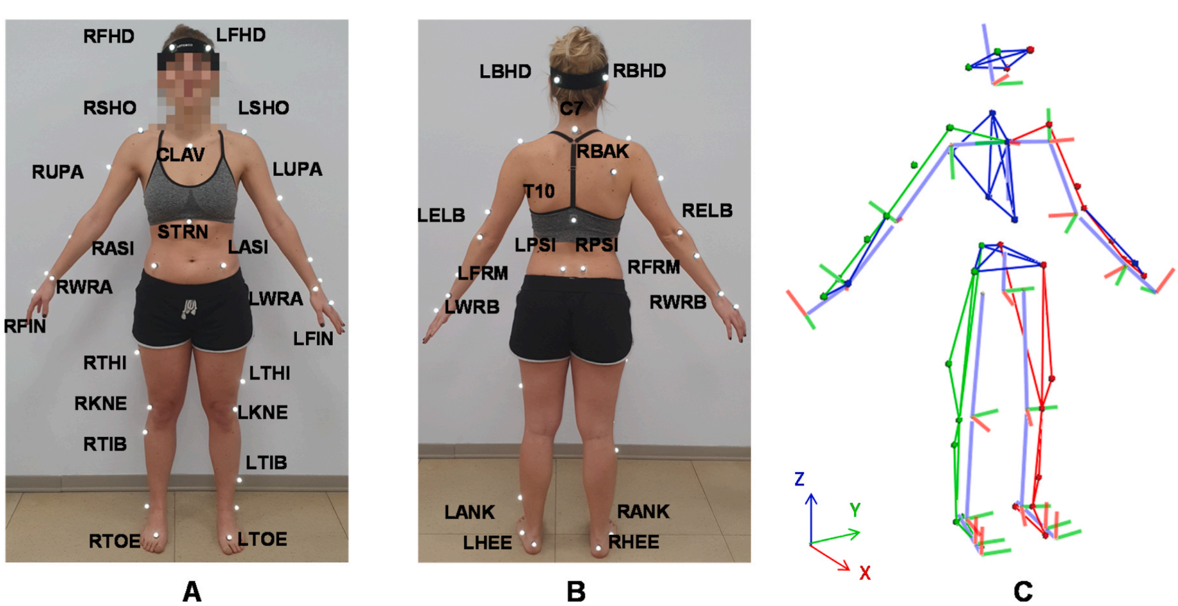


**Fig. 1.** Front (A) and back (B) markers positioning on the human body for the reconstruction of the Plug-in-Gait model (C).

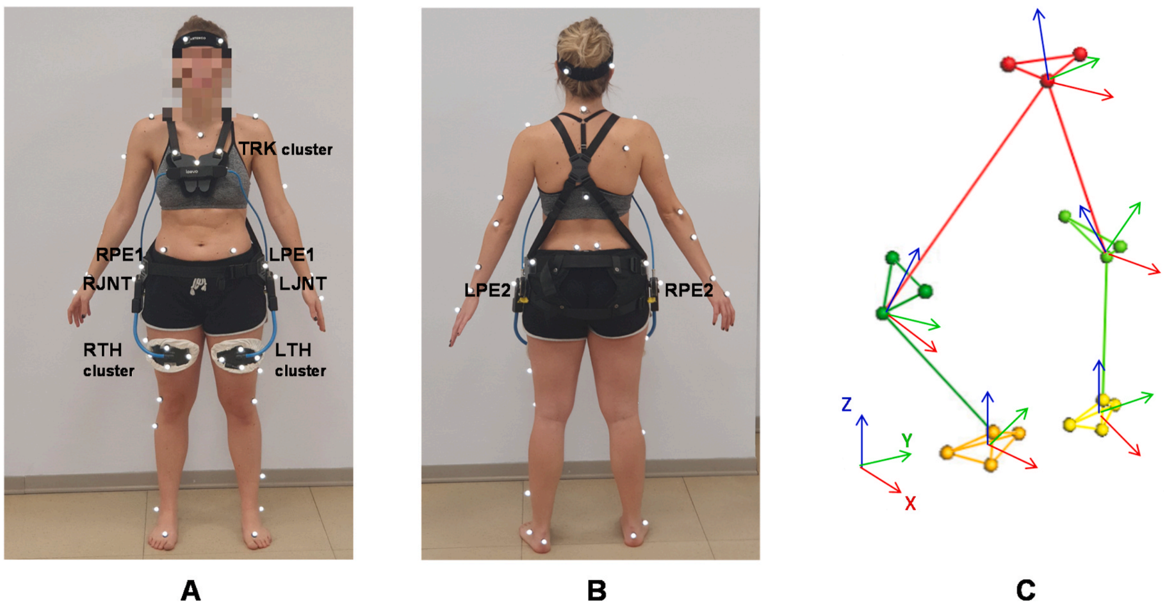


**Fig. 2.** Front (A) and back (B) markers positioning on the exoskeleton rigid parts for the reconstruction of the customized exoskeleton kinematic model (C).

gradually increased until the subject felt comfortable and did not perceive any pressure on the chest. In the current study, participants typically selected an initial angle of either 10° or 15°. To facilitate familiarity with the passive wearable device, an initial training session was conducted. The donning phase and the training session lasted approximately 20 min for each subject. During this time, participants rotated the trunk (flexion/extension, lateral bending, internal/external rotation) and performed different motion tasks, including walking, standing, stooping, squatting and sitting. Subsequently, they were asked to assume a static standing posture for the labeling and scaling of both the human and exoskeleton models.

Next, the main task was to perform gait trials, each consisting of five gait passages. The path length performed by the subject in each passage was 9 m. Participants were asked to walk barefoot back and forth along the path, at a self-selected comfortable speed. For each gait passage, 5 steps for each foot were recorded, resulting in a total of 25 stride cycles across all subjects. Gait events were identified using lateral ankle markers, as proposed by the Plug-in-Gait model. The gait trials were performed i) without the exoskeleton, ii) wearing the exoskeleton without assistance and iii) wearing the exoskeleton with assistance. The order of these trials was randomized.

### 2.5. Signal processing and data analysis

Marker's 3D position were registered and post-processed using Vicon Nexus, employing standard and customized operations for human and exoskeleton kinematics. Vicon's standard algorithms were used to reconstruct marker trajectories and fill in any gaps. The Plug-in-Gait Dynamic pipeline was used to calculate kinematic data of the human body, while gait events were manually identified based on foot marker trajectories and video recordings. Customized pipelines were implemented in Vicon Nexus for the evaluation of the kinematics of the exoskeleton model. Additionally, customized Matlab® routines were implemented to calculate the outcomes of interest. Due to the verified symmetry during gait (limp index 1.00), left and right sides were averaged. The following parameters were estimated for the gait cycle (GC).

- 14 gait spatio-temporal parameters (STPs): Gait speed (m/s), Stride length (m), Step length (m), Step width (m), Stride time (s), Step time (s), Stance time (s), Swing time (s), Single support time (s), Double support time (s), Stance duration (%GC), Swing duration (%GC), Single support phase (%GC), Double support phase (%GC);
- angular kinematics (°) and ROMs (°) of human joints (shoulder, spine, hip, knee, ankle) in frontal, sagittal and transverse planes;
- angular ROMs (°) of human trunk and thighs, and angular ROMs (°) of the exoskeleton trunk and thigh pads in frontal, sagittal and transverse planes with respect to the global coordinate system.

For each subject, gait spatio-temporal parameters, kinematic curves and angular ROMs were evaluated for each gait cycle and across all the different gait trials. The human kinematic curves were normalized with respect to the %GC and the mean curves among cycles were evaluated for each subject. Maximum and minimum kinematic values during the stance and swing subphases have been identified and compared. Considering the exoskeleton, 3D angular ROMs of the trunk and thigh pads were estimated and averaged across gait cycles for each subject. For each biomechanical parameter, an average value was calculated for each subject. Given that no significant differences were found between male and female participants, the subjects were grouped into a single sample for analysis. Then, a descriptive statistics was evaluated for each subject in terms of mean value and standard error of the mean (SEM), defined as the standard deviation divided by the square root of the sample size. A statistical analysis was implemented to compare the biomechanical parameters of interest among subjects in the three exoskeleton settings (no-exo, exo-no-support, exo-with-support). The hypothesis of normality of data distribution was confirmed with the Shapiro-Wilk test ($\alpha > 0.05$). Separate repeated-measures analysis of variance (ANOVA) was performed on each outcome measure related to human movement (STPs, angular ROMs and minimum/maximum kinematic values during stance and swing) and the exoskeleton condition assumed as the within-subject factor. Significant effects were followed by post hoc pairwise comparisons (no exoskeleton – exoskeleton conditions) using Bonferroni corrected t-tests. Post hoc results (mean pairwise difference, 95% confidence interval CI with Bonferroni correction for difference, p-value) were reported only in cases of statistical significance. Statistical significance was set at a level of significance $\alpha = 0.05$ and partial eta-squared ($\eta_p^2$) was used to quantify effect sizes for main/interaction effects and were qualitatively interpreted as 0.01 = small, 0.06 = medium, 0.14 = large (Cohen, 1973). When comparing human and exoskeleton ROMs at trunk and thigh segments, the two exoskeleton settings (exo-no-support, exo-with-support) were considered separately, in each anatomical plane,. Multiple paired t-tests were calculated using Bonferroni correction and Cohen's d values were determined as measures of effect sizes (d = 0.2, small; d = 0.5, medium; d = 0.8, large) (Cohen, 2013).

## 3. Results

Biomechanical results are reported for the three configurations (no-exo, exo-no-support, exo-with-support) in the three subsections: spatio-temporal parameters, human angular kinematics and exoskeleton angular kinematics.

### 3.1. Spatio-temporal parameters

Fig. 3 summarizes the mean and SEM values of the spatio-temporal parameters among subjects. The repeated measure ANOVA with Greenhouse-Geisser correction determined the significant difference for gait speed ($F(1.55,17.09) = 14.08$, *p*-value = 0.001, $\eta_p^2 = 0.56$), stride length ($F(1.87,20.59) = 10.14$, *p*-value = 0.001, $\eta_p^2 = 0.48$), step length ($F(1.88,20.70) = 9.75$, *p*-value = 0.001, $\eta_p^2 = 0.47$), stride time ($F(1.23,13.54) = 6.96$, *p*-value = 0.016, $\eta_p^2 = 0.39$), step time ($F(1.22,13.49) = 6,87$, *p*-value = 0.016, $\eta_p^2 = 0.39$), stance time ($F(1.24,13.61) = 5.83$, *p*-value = 0.025, $\eta_p^2 = 0.35$), and swing time ($F(1.52,16.66) = 4.27$, *p*-value = 0.041, $\eta_p^2 = 0.28$). Post hoc analysis with a Bonferroni adjustment revealed that the gait speed achieved without the exoskeleton shows a significant increase compared to the gait speed obtained when wearing the exoskeleton, both without support (0.074 (95% CI, 0.027 to 0.122) m/s, *p*-value = 0.003, d = 1.23) and with support (0.096 (95% CI, 0.03 to 0.162) m/s, *p*-value = 0.005, d = 1.20).

Considering the spatial parameters, post hoc analysis with a Bonferroni adjustment revealed that the stride length calculated without the exoskeleton is significantly higher compared to the exoskeleton without support (0.053 (95% CI, 0.013 to 0.092) m, *p*-value = 0.009, d = 1.10) and with support (0.060 (95% CI, 0.014 to 0.106) m, *p*-value = 0.01, d = 1.05). Similarly, the step length obtained without the exoskeleton is significantly higher than the value obtained with the exoskeleton without support (0.027 (95% CI, 0.007 to 0.046) m, *p*-value = 0.009, d = 1.13) and with support (0.030 (95% CI, 0.007 to 0.054) m, *p*-value = 0.01, d = 1.04). Considering the temporal parameters, post hoc analysis with a Bonferroni adjustment revealed that the stride time performed without the exoskeleton shows a significant reduction compared to the value obtained when wearing the exoskeleton, both without support (−0.028 (95% CI, −0.055 to −0.001) s, *p*-value = 0.04, d = −0.85) and with support (−0.046 (95% CI, −0.09 to −0.001) s, *p*-value = 0.05, d = −0.79). Similarly, significant decreases were obtained for the step time comparing the no exoskeleton condition with the exoskeleton without support (−0.014 (95% CI, −0.028 to $-1.89 \cdot 10^{-5}$) s, *p*-value = 0.05, d = −0.82) and with support (−0.023 (95% CI, −0.046 to 0) s, *p*-value = 0.05, d = −0.79). Table 1 shows the percentage temporal parameters (swing and stance duration, single and double support phases) in terms of mean value and SEM among subjects, but no statistical differences were obtained among the configurations.

**Table 1**
Mean and SEM of percentage temporal parameters calculated in the three different configurations.

| Percentage temporal parameters (%GC) Mean (SEM) | No exo | Exo no support | Exo with support |
|---|---|---|---|
| Swing duration | 38.72 (0.60) | 38.30 (0.46) | 38.51 (0.50) |
| Stance duration | 61.55 (0.58) | 61.90 (0.44) | 61.75 (0.48) |
| Single support | 38.72 (0.60) | 38.30 (0.46) | 38.51 (0.50) |
| Double support | 22.87 (1.17) | 23.59 (0.90) | 23.24 (0.98) |

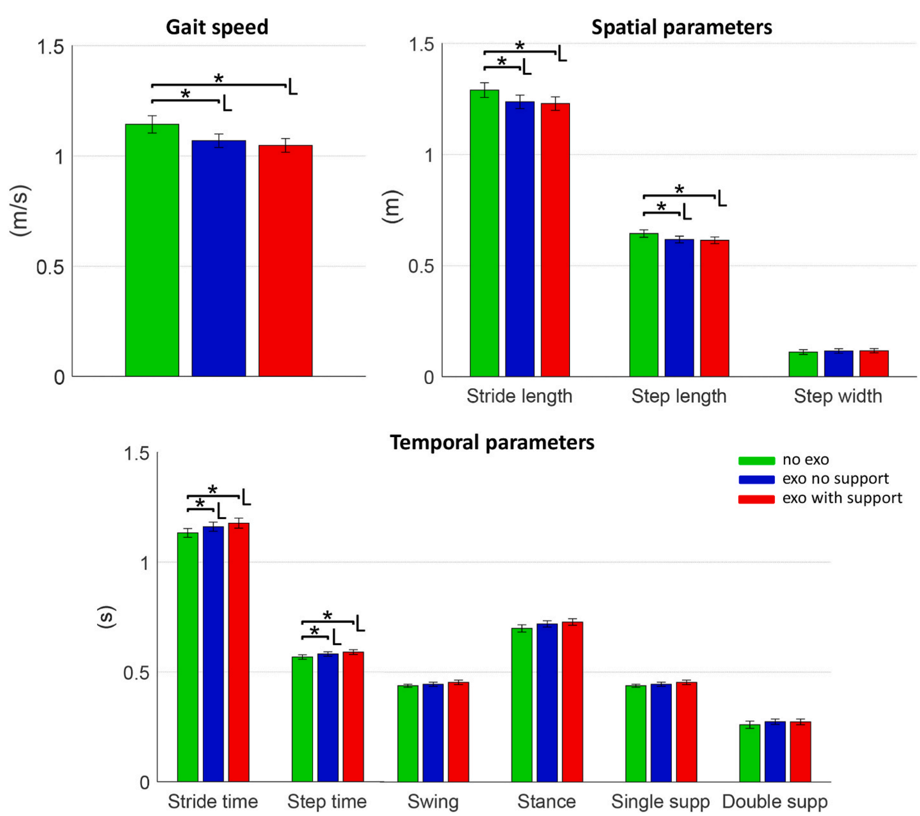


**Fig. 3.** Mean and SEM of spatio-temporal parameters calculated in the three different configurations (* represents a *p*-value <0.05 obtained in the pairwise post-hoc test with Bonferroni correction, L represents large Cohen's d value).

### 3.2. Human angular kinematics

Fig. 4 shows the 3D kinematic curves (mean and SEM values among subjects) of human shoulder and spine angles vs. percentage of gait cycle, evaluated in the three different exoskeleton settings. Moreover, the angular ROMs of each curve are reported as bar graphs with average and SEM values among subjects, reporting the significant difference among configurations obtained with the post-hoc tests. The repeated measure ANOVA with Greenhouse-Geisser correction determined the significant difference for shoulder angular ROMs in the sagittal plane (F(1.51,16.60) = 6.42, *p*-value = 0.013, $\eta_p^2$ = 0.37) and for the spinal angular ROMs in the frontal (F(1.39,15.31) = 8.34, *p*-value = 0.007, $\eta_p^2$ = 0.43), sagittal (F(1.95,21.49) = 25.28, *p*-value<0.001, $\eta_p^2$ = 0.69) and transverse (F(1.48,16.28) = 5.10, *p*-value = 0.03, $\eta_p^2$ = 0.32) planes.

Considering the shoulder angular ROMs, the post-hoc analysis with Bonferroni correction pointed out a significant reduction in the sagittal plane when wearing the exoskeleton with support compared with the no exoskeleton condition (5.55 (95% CI, 1.63 to 9.47) degrees, *p*-value = 0.006, d = 1.15). Considering the spinal angular ROMs, the post-hoc analysis with Bonferroni correction pointed out a significant reduction both in the frontal and sagittal planes. In the frontal plane, the angular ROM registered without the exoskeleton is significantly higher compared to values obtained wearing the exoskeleton without support (0.89 (95% CI, 0.33 to 1.44) degrees, *p*-value = 0.003, d = 1.29) and with support (1.37 (95% CI, 0.25 to 2.48) degrees, *p*-value = 0.01, d = 1.00). In the sagittal plane, the angular ROM registered without the exoskeleton is significantly higher compared to values obtained wearing the exoskeleton without support (1.25 (95% CI, 0.66 to 1.85) degrees, *p*-value = 0.001, d = 1.71) and with support (1.15 (95% CI, 0.62 to 1.69) degrees, *p*-value = 0.001, d = 1.74).

Fig. 5 shows the 3D kinematic curves (mean and SEM values among subjects) of human lower body angles vs. percentage of gait cycle and the angular ROMs, calculated in the three different configurations. Moreover, the angular ROMs of each curve are reported as bar graphs with average and SEM values among subjects, reporting the significant difference among configurations obtained with the post-hoc tests. The repeated measure ANOVA with Greenhouse-Geisser correction determined the significant difference for hip angular ROMs in the frontal plane (F(1.72,18.93) = 16.72, *p*-value<0.001, $\eta_p^2$ = 0.61) and in the sagittal plane (F(1.93,21.18) = 18.20, *p*-value<0.001, $\eta_p^2$ = 0.62). For the knee joint, significant results were pointed out in the frontal plane (F(1.61,17.77) = 22.98, *p*-value<0.001, $\eta_p^2$ = 0.67), in the sagittal plane (F(1.66,18.29) = 6.16, *p*-value = 0.01, $\eta_p^2$ = 0.36) and in the transverse plane (F(1.90,20.92) = 3.56, *p*-value = 0.04, $\eta_p^2$ = 0.24), while for the ankle joint significant results were obtained in the frontal plane (F(1.23,13.52) = 6.52, *p*-value = 0.02, $\eta_p^2$ = 0.37). Considering the hip angular ROMs, the post-hoc analysis with Bonferroni correction pointed out a significant reduction both in the frontal and sagittal planes whear ROM registered without the exoskeleton is significantly higher compared to values obtained wearing the exoskeleton without support (1.58 (95% CI, 0.70 to 2.47) degrees, *p*-value = 0.001, d = 1.45) and with support (2.30 (95% CI, 1.05 to 3.55) degrees, *p*-value = 0.001, d = 1.49). In the sagittal plane, the angular ROM registered without the exoskeleton is significantly higher compared to values obtained wearing the exoskeleton without support (2.84 (95% CI, 1.22 to 4.46) degrees, *p*-value = 0.001, d = 1.42) and with support (2.95 (95% CI, 1.29 to 4.60) degrees, *p*-value = 0.001, d = 1.45).

Considering the knee angular ROMs, the post-hoc analysis with Bonferroni correction pointed out a significant reduction both in frontal and sagittal planes when wearing the exoskeleton. In the frontal plane, the angular ROM registered without the exoskeleton is significantly higher compared to values obtained wearing the exoskeleton without support (2.31 (95% CI, 0.41 to 4.22) degrees, *p*-value = 0.01, d = 0.99) and with support (3.71 (95% CI, 2.33 to 5.12) degrees, *p*-value<0.001, d = 2.17). In the sagittal plane, the angular ROM registered without the exoskeleton is significantly higher compared to values obtained wearing the exoskeleton with support (2.42 (95% CI, 0.75 to 4.07) degrees, *p*-value = 0.005, d = 1.19). Considering the ankle joint, in the frontal plane the post-hoc analysis with Bonferroni correction pointed out a significant decrease for the exoskeleton with support condition compared to the no-exo condition (0.42 (95% CI, 0.000 to 0.84) degrees, *p*-value = 0.05, d = 0.81). The vertical dotted lines reported in Fig. 5 represent the separation between stance and swing phases and allow the analysis of angle trajectories in the different gait subphases. The repeated measure ANOVA with Greenhouse-Geisser correction determined the significant difference for sagittal hip minimum value in stance

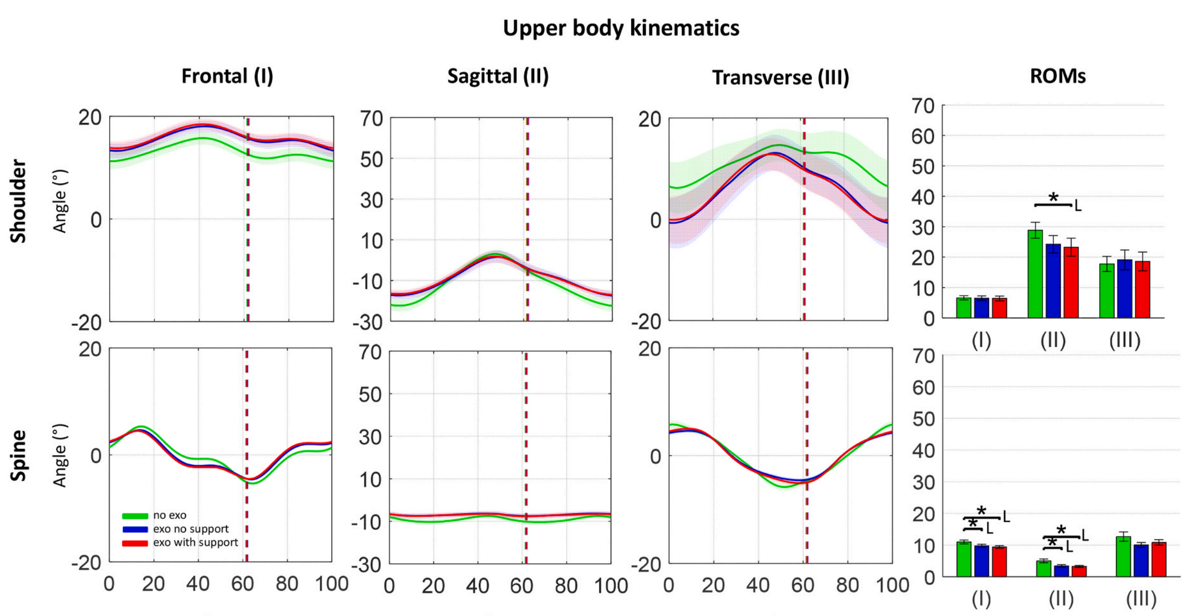


**Fig. 4.** Shoulder and spine relative kinematics in frontal (I), sagittal (II) and transverse (III) planes obtained in the three configurations. Curves are expressed as percentage of the gait cycle (%GC) and vertical dotted lines represent the separation between stance and swing phases. Lines represent the average value among subjects, the shade region the SEM. ROMs of curves are reported with the post hoc pairwise tests results (* represents a *p*-value <0.05, L represents large Cohen's d value). (For interpretation of the references to colour in this figure legend, the reader is referred to the Web version of this article.)

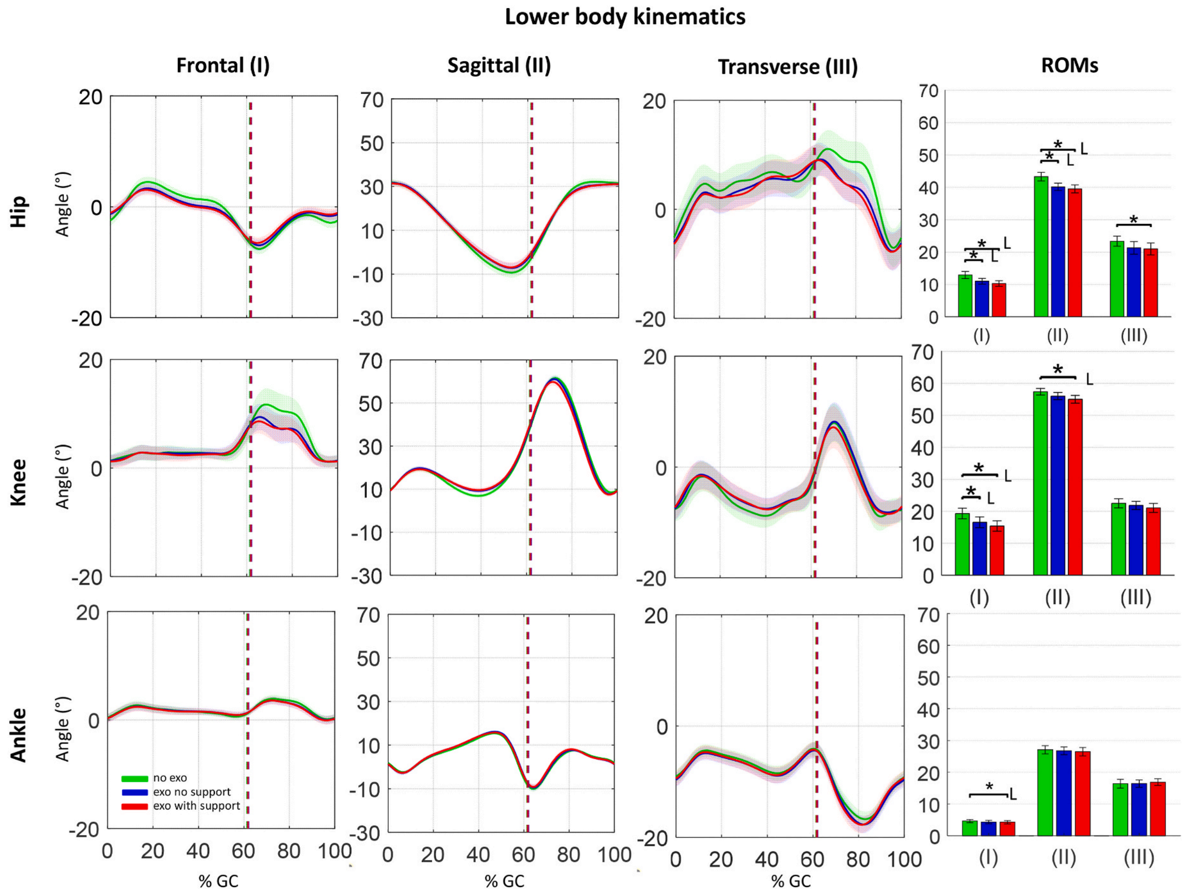


**Fig. 5.** Hip and knee relative kinematics in frontal (I), sagittal (II) and transverse (III) planes obtained in the three configurations. For the foot segment, ankle relative kinematics is reported in the frontal and sagittal planes, while the foot progression angle is reported in the transverse plane. All curves are expressed as percentage of the gait cycle (%GC) and vertical dotted lines represent the separation between stance and swing phases. Lines represent the average value among subjects, the shade region the SEM. ROMs of curves are reported stressing the significance of post-hoc tests (* represents a p-value <0.05, L represents large Cohen's d value). (For interpretation of the references to colour in this figure legend, the reader is referred to the Web version of this article.)

phase ($F(1.49,16.39) = 17.55$, $p$-value$<0.001$, $\eta_p^2 = 0.62$), the significant difference for sagittal knee minimum value in stance phase ($F(1.83,20.13) = 19.48$, $p$-value$<0.001$, $\eta_p^2 = 0.64$), the significant difference for frontal knee maximum value in swing phase ($F(1.62,17.81) = 20.93$, $p$-value$<0.001$, $\eta_p^2 = 0.66$). The post-hoc analysis with Bonferroni correction pointed out significant differences of the no exoskeleton condition with the two exoskeleton ones, while no significant difference were obtained in the comparison between exoskeleton conditions. For the hip joint angle in the sagittal plane during the stance phase, the extension angle is significantly reduced when wearing the exoskeleton without (−2.44 (95% CI, −3.45 to −1.43) degrees, $p$-value $= 0.001$, $d = -1.97$) and with support (−2.55 (95% CI, −4.24 to −0.83) degrees, $p$-value $= 0.004$, $d = -1.22$). For the knee joint angle in the frontal plane during the swing phase, the maximum angle is significantly reduced when wearing the exoskeleton without support (2.53 (95% CI, 0.82 to 4.23) degrees, $p$-value $= 0.005$, $d = 1.21$) and with support (3.61 (95% CI, 1.73 to 5.48) degrees, $p$-value $= 0.001$, $d = 1.57$); in the sagittal plane during the stance phase, the knee extension angle is significantly reduced when wearing the exoskeleton without support (−2.43 (95% CI, −3.70 to −1.16) degrees, $p$-value $= 0.001$, $d = -1.56$) and with support (−2.46 (95% CI, −3.90 to −1.03) degrees, $p$-value $= 0.002$, $d = -1.40$).

### 3.3. Exoskeleton angular kinematics

Tables 2 and 3 show the comparison between the ROMs of human spine and the ROMs of exoskeleton trunk pad rotation, and between the ROMs of human thigh and the ROMs of exoskeleton thigh pad rotation, respectively. Descriptive statistics (mean and SEM values) and paired $t$-test results are reported in tables. In both the exoskeleton configurations, the trunk pad shows a significant greater range of motion compared to the human spine segment. The thigh pad shows a significant lower range of motion compared to the human thigh segment in the frontal and sagittal planes.

**Table 2**
Mean and SEM of human spine and trunk pad angular ROMs calculated with respect to the global coordinate system; paired $t$-test results with Bonferroni correction ([a] adjusted p-value), t-value (t), degrees of freedom (dg) and Cohen'd effect size (d).

| Human spine and trunk pad ROMs<br>Mean (SEM) | Exo no support | | Exo with support | |
|---|---|---|---|---|
| | Human | Exo | Human | Exo |
| Frontal plane | 3.32° (0.47) | 6.20° (0.75) | 3.36° (0.50) | 6.63° (0.48) |
| *t(dg), p-value[a], d* | −2.5(11), 0.36, 0.72 | | −3.71(11), **0.036**, 1.07 | |
| Sagittal plane | 2.86° (0.23) | 5.82° (0.57) | 2.74° (0.26) | 5.88° (0.56) |
| *t(dg), p-value[a], d* | −4.65(11), **0.01**, 1.34 | | −4.51(11), **0.01**, 1.30 | |
| Transverse plane | 8.55° (0.55) | 16.23° (1.10) | 8.53° (0.71) | 16.39° (1.29) |
| *t(dg), p-value[a], d* | −8.77(11), < **0.01**, 2.53 | | −5,91(11), < **0.01**, 2.62 | |

## 4. Discussion

The principal function of the back-assistance exoskeleton is to reduce

**Table 3**
Mean and SEM of human thigh and thigh pad angular ROMs calculated with respect to the global coordinate system; paired *t*-test results with Bonferroni correction ([a] adjusted p-value), t-value (t), degrees of freedom (dg) and Cohen'd effect size (d).

| Human thigh and thigh pad ROMs Mean (SEM) | Exo no support | | Exo with support | |
|---|---|---|---|---|
| | Human | Exo | Human | Exo |
| Frontal plane | 7.38° (0.43) | 5.18° (0.46) | 7.32° (0.37) | 4.93° (0.48) |
| *t(dg), p*-value[a], *d* | 3.44(11), 0.072, 0.99 | | 3.59(11), **0.048**, 1.04 | |
| Sagittal plane | 40.35° (1.20) | 29.73° (1.23) | 39.70° (1.35) | 26.86° (1.12) |
| *t(dg), p*-value[a], *d* | 5.81(11), **<0.01**, 1.68 | | 6.45(11), < **0.01**, 1.86 | |

the biomechanical loads of the human back during specific movements characterized by a trunk-flexed posture. Nevertheless, many work tasks often necessitate a combination of various human activities (e.g. lifting, carrying, walking). For this reason, it is essential to investigate the effects of wearing the device during activities where the assistance may not be required or when the task differs from those for which the exoskeleton was originally designed. Additionally, while the device is engineered to provide assistance in the sagittal plane, it may also induce kinematic effects and restrictions in the frontal and transverse planes. The discussion of the results is reported considering the three distinct groups of biomechanical parameters: spatio-temporal parameters, human angular kinematics and exoskeleton angular kinematics.

### 4.1. Spatio-temporal parameters

The spatio-temporal parameters showed significant differences in the no-exoskeleton condition compared to both exoskeleton conditions (with and without support). Walking speed, step length, and stride length were all significantly higher without the exoskeleton, while step and stride times were shorter, indicating faster movements without the device. However no significant differences were found in percentage-based temporal parameters, such as swing and stance durations, and single and double support phases, across configurations.

The reduction in gait speed observed when wearing the exoskeleton is in line with the preliminary test performed by one healthy subject, as reported in (Panero et al., 2021). In that preliminary analysis, a young healthy female performed several motion tasks without and with the passive Laevo 2.5, aiming to investigate the interaction between human and exoskeleton and to establish a suitable biomechanical procedure for objective evaluation. Despite the presence of the exoskeleton appeared to increase the step width, no significant differences were pointed out.

The reduction in step length observed in the current study aligns with the experimental tests conducted by Park et al. (2022b), where the effects of the passive back-support exoskeleton backX™ were tested. In their study, multiple exoskeleton torque assistance conditions (no supportive torque, low supportive torque, high supportive torque) were compared with a control condition (no exoskeleton) during both over-ground and treadmill walking. Twenty young and healthy subjects participated in the experimental analysis, and both gait performance and stability were investigated. Results highlighted significant reductions in step length when wearing the exoskeleton, particularly under higher torque assistance, and increases in step width across all exoskeleton conditions compared to the control.

However, the increase in stride time and step time observed in the current study contrasts with Park et al. (2022b), who reported a significant reduction in gait cycle time when wearing the exoskeleton, especially under higher support torques. This discrepancy likely arises from differences in the architecture of the passive exoskeletons used in the two studies. Specifically, differences in device weight distribution, joint articulation design, or levels of torque assistance might influence certain spatio-temporal parameters, such as stride time and step time, differently. For instance, while Park et al. (2022b) utilized a back-support exoskeleton designed for torque modulation, the exoskeleton in the current study may have introduced greater resistance or altered the user's natural gait dynamics due to its passive structure and differing mechanical design. These architectural differences appear to influence temporal parameters like stride and step time, while parameters such as step length remain more consistently affected by the general encumbrance of wearing an exoskeleton.

Moreover, Park et al. (2022b) noted high gait variability when wearing the device, whereas stride-to-stride variability was typically lower without the exoskeleton. In contrast, no significant differences in percentage-based temporal parameters were observed in the current study, as reported in Table 1. These parameters provide insight into different gait temporal subphases but STPs results suggest that the slowing of human locomotion is primarily influenced by the structure of the exoskeleton rather than the assistance itself, despite its low total mass and minimal design.

The adaptation of gait parameters could be interpreted as a perceived encumbrance caused by the unfamiliarity of the device, although the temporal distribution of gait subphases in each stride remained unaffected. Further investigations with longer period tasks duration and correlation with subjective evaluations may clarify whether extended use of the device could lead to increased comfort and reduced restrictions during movement.

### 4.2. Human angular kinematics

The present study evaluated angular kinematics to assess potential postural adaptations while wearing the exoskeleton. It found significant decreases in joint ranges of motion (ROM) across the shoulder, spine, hip, knee, and ankle. Specifically, shoulder ROM decreased in the sagittal plane, while spine ROM decreased in both the sagittal and frontal planes. Similarly, hip and knee ROMs declined in both the frontal and sagittal planes, and ankle ROM was reduced in the frontal plane for the supported exoskeleton condition. Additionally, hip extension and knee angles during stance and swing phases were lower with the exoskeleton, indicating consistent effects across configurations.

The current significant effect of wearing the exoskeleton on spinal posture aligns with previous experimental studies on passive exoskeletons (Madinei et al., 2020; Luger et al., 2021a; So et al., 2022), despite the tasks performed by subjects being different (lifting and carrying tasks) and the exclusive observation of the sagittal plane in those studies. The observed reduction in shoulder ROM in the sagittal plane while using the supported exoskeleton suggests that the device's assistance may affect joints not directly involved in support. For the lower limbs, results in the sagittal plane align with previous analyses (Poliero et al., 2021; Park et al., 2022a). During carrying tasks (Poliero et al., 2021), hip ROM decreased by about 23% and knee ROM by 10% when wearing the exoskeleton. Similar to the results presented in this study, in (Poliero et al., 2021) no significant difference was observed between the exoskeleton's assistance and transparency modes. Park and colleagues (Park et al., 2022a) quantified sagittal joint ROMs during walking with and without the exoskeleton, finding significant effects of the exoskeleton on all joint measures. Specifically, sagittal hip ROM significantly decreased in all exoskeleton conditions compared to the control, while sagittal knee and ankle ROMs were lower with high exoskeleton support than in the no-exoskeleton setting.

Current findings, supported by previous studies, indicate that wearing an exoskeleton reduces hip and knee ROMs in the sagittal plane. This study went further by analyzing maximum and minimum values within specific gait subphases (stance/swing). For the hip, extension angles were significantly reduced during the stance phase with the exoskeleton in both configurations, while no significant changes were found in the swing phase. For the knee, the maximum angle in the frontal plane decreased during the swing phase, and knee extension angles were significantly reduced in the sagittal plane during the stance phase. These results are not supported or contradicted by existing literature, as no prior studies have addressed this specific aspect. Given the distinct roles of the lower limbs throughout the gait cycle, further research on lower limb dynamics is suggested to explore the biomechanical effects of the exoskeleton, particularly regarding joint forces, moments, and their interaction with exoskeleton assistance.

All these results demonstrate that the presence of the back-assistance imposes angular restrictions on the lower body not only in the sagittal plane, but also in the frontal plane, despite no assistance being applied in this plane. A 3D multibody analysis of human body could be a key instrument for investigating the postural effects of an external exoskeleton and the obtained results can be used to improve its design.

### *4.3. Exoskeleton angular kinematics*

Due to the different and reduced exoskeleton degrees of freedom compared to the human body joints, misalignments of the device might occur during motion. In both the exoskeleton configurations, the trunk pad shows significant greater range of motion compared to the human spine segment, while the thigh pads show significant lower range of motion compared to the human thigh both in the frontal and sagittal planes. This behavior may be attributed to the design of the pads: the trunk pad features multiple degrees of freedom, while the lower ROMs of the thigh pads are likely due to hinge joint restriction, which allows rotation only in the sagittal plane. Such misalignments and rotation constraints can lead to user discomfort, slippage and undesirable interface forces. The objective evaluation of these parameters might be fundamental for enhancing the exoskeleton's wearability and the adaptability to different anthropometric measures and postures. Also, in this case, there are no previous studies that have conducted these evaluations, making any comparison impossible.

To sum up, experimental tests highlight the biomechanical effects and the postural changes induced by the device during locomotion. While some discrepancies exist with previous studies, current results confirm the kinematic effects of the exoskeleton on the human body. Nevertheless, it is unclear whether this outcome is beneficial in the long-term use and additional investigations are necessary to relate the postural variations to the change in loads in all the human joints. Additionally, based on these laboratory results, establishing a kinematic range of operational relevance is challenging because of the individual variability in body types and movement patterns, as well as the complexity of tasks and limitations in the device design that may restrict motion.

### *4.4. Limitations*

Despite the significant results, the study presents some limitations. Participants involved in the study were young and without muscular disorders; the extension of the current results to a larger population of interest should be done with caution and requires further testing. The outcome measures focused on kinematics, without addressing dynamic measurements. Measuring the interface forces between human segments and exoskeleton pads could help estimate the assistance provided by the exoskeleton during different gait phases and relate it to the human posture. Finally, the experimental tests were conducted in a laboratory setting and for a short duration. Further investigations are necessary to analyze the effects of a prolonged exoskeleton use in real-working tasks.

## 5. Conclusion

In the last decades, according to the human-centric approach promoted by the Industry 5.0, occupational exoskeletons have been introduced as possible solutions to reduce the physical efforts and loads of the user during working tasks. Back-assistance exoskeletons seem to be promising for preventing or reducing the prevalence of low back pain, but some issues are still open concerning their biomechanical effects on the entire body and during different motion tasks. The present study investigated the biomechanical impacts of a passive back-assistance exoskeleton on users during locomotion considering the 3D kinematics of the whole body. Three configurations were compared in terms of spatio-temporal parameters, human kinematics and exoskeleton kinematics to assess the effects of both the exoskeleton structure and its assistance. Users significantly reduced the gait speed and the stride length when wearing the exoskeleton, while they increased the stride time. Kinematic analysis showed significant reductions in ROM for all joints when wearing the exoskeleton. However, no significant differences were found between the two assistance configurations (without and with the support), indicating that the presence of the exoskeleton structure primarily influences the outcomes, rather than the assistance level. Significant ROM differences were highlighted in the comparison of human and exoskeleton kinematics, emphasizing the misalignment of the device during the movement.

Future research with larger population and with longer trial periods are needed to explore the biomechanical effects on the human body related to the user's anthropometry (gender, age, presence of low back pain) and to the prolonged use of the device. Moreover, the investigation of fatigue and discomfort perceived by the user and caused by the additional mass of the exoskeleton should be a focus of attention.

## CRediT authorship contribution statement

**Elisa Panero:** Writing – original draft, Methodology, Investigation, Data curation. **Stefano Pastorelli:** Writing – review & editing, Supervision, Project administration. **Laura Gastaldi:** Writing – review & editing, Supervision, Methodology, Investigation.

## Declaration of competing interest

The authors declare that the disclosed information is correct and that no other situation of real, potential or apparent conflict of interest is known to them. The authors undertake to inform you of any change in these circumstances.


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