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Article

Investigating Kinematics and Electromyography Changes in Manual Handling Tasks with an Active Lumbar Exoskeleton

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Abstract: Companies are becoming increasingly aware of the health of their employees and are now integrating exoskeleton solutions for both prevention and job maintenance. However, the effect of using exoskeletons is still an open question. Therefore, this study aimed to evaluate the impact of an active lumbar exoskeleton and its passive belt on trunk kinematics and muscle activity using instrumented motion analysis. Twenty-three healthy subjects volunteered to perform three handlings of a 5 kg load (free lifting, squat lifting, and load transfer) under three different experimental conditions. The "Control" condition was when the subject did not wear any device, the "Belt" condition was when the subject wore only the passive part of the exoskeleton, and the "Exo" condition was when the subject wore the active exoskeleton. Based on the Rapid Upper Limb Assessment scale, the exoskeleton reduced the time spent in angles that were considered dangerous for the back, according to ergonomic evaluations. Furthermore, for the handling sessions, it was observed that the exoskeleton did not modify muscle activity in the abdominal–lumbar region.

Keywords: lumbar spine; exoskeleton; motion analysis; kinematics; electromyography; handling

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1. Introduction

Musculoskeletal disorders (MSDs) can affect the joints, muscles, and tendons and are often related to professional activity [1]. Work-related factors can contribute to the onset, duration, and worsening of MSDs, which may eventually lead to sick leaves or the need for professional reorientation. Consequently, managing and preventing MSDs has become a major challenge for the companies concerned [2]. In France, MSDs account for 87% of occupational diseases, and back pain accounts for 20% of accidents at work, which can lead to serious aftereffects with the risk of professional exclusion. In addition, it has a very high cost for companies, amounting to nearly two billion euros in 2017 [2]. For several years, studies have been conducted on MSD risk factors, highlighting two types of risk factors in companies: physical factors (excessive repetitive movements, heavy lifting, awkward postures, etc.) and psychosocial factors (lack of social support, high psychosocial work demand, lack of autonomy, etc.) [3]. In this context, the utilization of exoskeletons has been advocated to reduce work-related musculoskeletal stress [4].

More precisely, the back is the area most affected by musculoskeletal stress in various industries, i.e., construction [4], nursing [5], farming [6], and manufacturing [7]. In this context, the utilization of exoskeletons has been advocated to reduce work-related musculoskeletal stress. Additionally, spinal traction devices offer opportunities for preventing low back pain [8]. However, designing lower-back exoskeletons with such spinal traction functions presents challenges in terms of potential side effects on adjacent levels, which could globally limit the efficiency of the exoskeleton [9].

Despite mechanization, automation, and ergonomic interventions aimed at reducing the occurrence and progression of musculoskeletal disorders [10], many industrial tasks require human mobility and the handling of loads [11,12]. The emergence of body-worn assistive devices called exoskeletons now entails the consideration of alternative solutions to relieve workers of musculoskeletal constraints [13]. The main objective of exoskeletons is to prevent injuries while maintaining the mobility of workers [14]. However, the daily use of exoskeletons in companies has raised questions about their direct effect or transparency [15], defined as their neutrality of biomechanical impact on humans [16,17] in terms of ranges of motion and the muscular activity of surrounding muscles [18]. Additionally, owing to the heterogeneity of exoskeleton designs, predicting physical human-exoskeleton interactions accurately to determine the appropriate prescription for use in industrial sectors requires specific investigation [19]. However, evaluating the effectiveness of industrial exoskeletons is a challenging issue [20,21]. Additionally, since an exoskeleton requires classification as a medical device by EU regulation (2017/745) [22], the rigorous evaluation of each active and passive component of the product is required [23]. This is particularly critical in the case of exoskeletons that combine active and passive components because a sophisticated control system is required to optimize performance and demonstrate the benefits of these designs [24].

Biomechanical research methods are commonly used to assess the impact of a device, training, or surgery on the body [25], and motion analysis is defined as a biomarker of the musculoskeletal system [26]. In human motion analysis studies, kinematics provides a quantitative approach to the motion, while muscle activity measurement by electromyography (EMG) is related to the forces that generated the motion [25]. In the context of pathologies or MSDs, motion analysis appears as a powerful decision-making tool [27] to quantify ability reduction or compensation strategy [28] and also could document the time spent at risk postures [29] based on the criteria of the Rapid Upper Limb Assessment (RULA) [30]. Consequently, kinematics and EMG are common to quantify the effect of the exoskeleton [31–33].

The objective of this study is to objectively quantify the biomechanical impact of an active lumbar exoskeleton. By conducting handling tasks in a motion capture laboratory and analysing kinematics and EMG data, we hypothesized that the active exoskeleton would influence the range of motion and modify the muscular activity of the trunk muscles. These two components are crucial for objectively categorizing the prescription for the future use of the active exoskeleton.

2. Materials and Methods

2.1. Exoskeleton

The exoskeleton used in this study was a lumbar ambulatory traction device (Japet.W, Japet Medical Devices©, Loos, France) [14]. It is designed to relieve lumbar strains [34] by applying a vertical traction force. The device consists of two sub-elements, a passive part with the textile belt and the active part with two series elastic actuators (SEA) sets positioned on both sides of the body (Figure 1A). These two parts are never employed in isolation.

The SEA is a compliant mechanism allowing a dynamic movement with an integrated damping for the comfort of the user [32]. An electronic system coupled with a mechanical transmission system allows the application of a lumbar traction force adjusted to the user's desired setting (4 kg, 8 kg, 12 kg, or 16 kg). The SEA is connected to the belts by a ball joint, allowing them to adapt to the movement of the trunk.

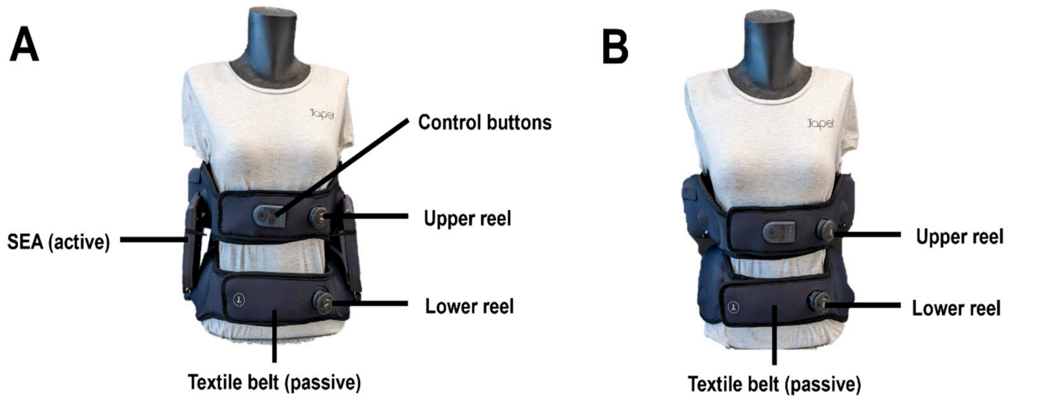

Figure 1. (**A**) Active Japet.W exoskeleton and (**B**) detail of passive belt used in the study.

2.2. Belt

The belt used is the passive part of the exoskeleton (textile belt) once SEAs are removed (Figure 1B). The upper and lower reels were retained to tighten the belt and adjust it to the trunk. This part is similar to a classic lumbar belt.

2.3. Population

Twenty-three healthy volunteers (fourteen women and nine men) participated in this study (age: 20.2 ± 1.2 years, mass: 67.7 ± 11.2 kg, height: 175.4 ± 9.9 cm) after signing a statement of informed consent to the experimental procedure in accordance with the Declaration of Helsinki.

2.4. Instrumented Motion Capture Session and Data Pre-Processing

Three handling task sessions were recorded. The "FREE LIFTING" session consisted of symmetric lifting of a 5 kg load for which the subject was free to choose his lifting technique (Figure 2A). During the "SQUAT LIFTING" session, a 5 kg load was lifted while the subject had to bend their knees and keep their trunk straight (Figure 2B). The "LOAD TRANSFER" session consisted of moving a 5 kg load between two tables that were perpendicular to each other (Figure 2C).

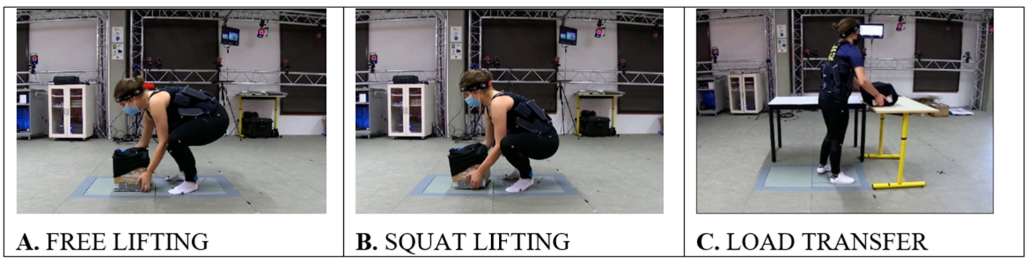

Figure 2. Handling simulation tasks.

Instrumented motion capture session was performed by 36 optoelectronic cameras (Vicon T160, Vicon Motion Systems Ltd., Oxford, UK) sampled to 200 Hz allowed to record 3D position of landmarks (Figure 3). Two body marker sets and two device marker sets were used [35]. The first body marker set was used to analyse the "Control" condition (Figure 3A). The combination of the second body marker set and the first device marker set was used to analyse the "Belt" condition (Figure 3B,i). Finally, the second body marker set combined with the second device marker set allowed the analysis of the "Exo" condition (Figure 3B,ii).

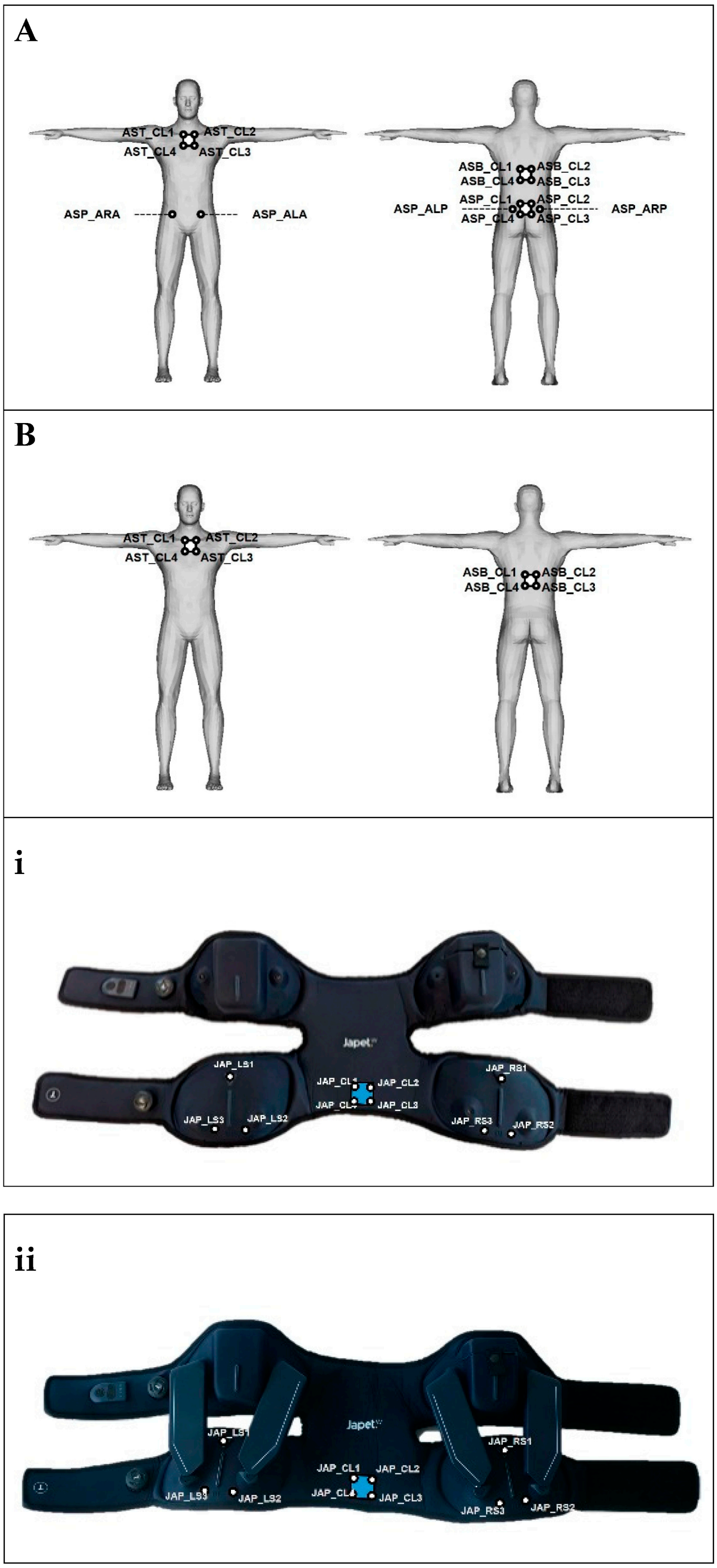

Figure 3. Marker sets for instrumented motion capture sessions. (**A**)—body marker set without device; (**B**)—body marker set with device; (**i**)—device marker set corresponding to the "Belt"; (**ii**)—device marker set corresponding to the "Exo".

All acquisitions were 3D reconstructed and labelled with Nexus Software (Vicon Motion Systems Ltd., Oxford, UK). Three pairs of surface EMG electrodes (PICO, Cometa srl, Milan, Italy) sampled to 1000 Hz were located in the trunk muscles: *Longissimus Dorsi* (LD), *Rectus Abdominis* (RA), and *Obliquus ExternusAbdominis* (OE), according to the SENIAM recommendations [36]. The subjects' skin was rubbed with abrasive paper and then cleaned with alcohol. The EMG sensor for the *Longissimus Dorsi* (LD) was placed 25 mm lateral to the L1 spine. The EMG sensor for the *Rectus Abdominis* (RA) was placed 25 mm above the umbilicus and 40 mm lateral to the midline. The EMG sensor for the *Obliquus Externus Abdominis* (OE) was placed 20 mm above and 20 mm anterior to the iliac crest [37]. Synchronized 3D coordinates of all markers and EMG data were exported from Nexus Software (Vicon Motion Systems Ltd., Oxford, UK) to a .csv file and processed directly on Matlab R2022a (The Mathworks Inc., Natick, MA, USA).

2.5. Data Analysis

2.5.1. Kinematics Data

The calibration phase allowed for the creation of an anatomical coordinate system for each body segment and its relationship with cluster markers [25]. The two body segments of interest in this study were anatomical pelvis segment associated with the reference frame R_pelvis and the thorax associated with the reference frame R_torso. Calculation of R_pelvis was based on the markers cluster {ASP_CL1, ASP_CL2, ASP_CL3, ASP_CL4} for the "Control" condition (Robertson et al. 2014) and {JAP_CL1, JAP_CL2, JAP_CL3, JAP_CL4} for the "Belt" and "Exo" conditions [38]. Calculation of R_torso is defined by the markers cluster [25] {AST_CL1, AST_CL2, AST_CL3, AST_CL4} for the three conditions. After the computation of the rotation matrix associated with the relative orientation of R_torso in the R_pelvis [25], flexion, lateral bending, and axial rotation angles were deduced [39].

To compare subjects performing handling tasks, the percentage of time spent in predefined angular ranges was analysed for the main rotation of the three movements [30]. The angular ranges were based on the RULA method [29] and were defined by {<−10°; [−10° 0°[; [0° 20°[; [20 60°[; ≥60°} for the flexion rotation angle and {<−20°; [−20° −10°[; [−10° 0°[; [0° 10°[; [10° 20°[; ≥20°} for the lateral bending and axial rotation angles. For FREE LIFTING and SQUAT LIFTING, only the flexion rotation angle was analysed [18,40].

2.5.2. EMG

The EMG data were analysed for all conditions for the handling simulation session (FREE LIFTING, SQUAT LIFTING, and LOAD TRANSFER). The root mean square (RMS) values of the EMG data were computed to evaluate the global muscle activity during the session [25]. RMS was calculated from the filtered EMG data thanks to a passband (4th order Butterworth) between 20 Hz and 500 Hz [41]. To normalize the RMS signal, the condition "Control" was taken as a reference, and each signal was divided by the maximum of the "Control" condition (MCC) for all records [42].

2.5.3. Statistical Analysis

For kinematics data, nonparametric Friedman tests were performed with all subjects' percentages of time spent in each RULA area of the handling sessions of the 3 conditions "Control", "Belt", and "Exo". The null hypothesis was no significance difference between the three conditions. The null hypothesis is rejected if the p-value (level of significance) of the Friedman test is less than 0.05. In this case, the alternative hypothesis (significant difference between the three conditions) was supported, and a post hoc test can be performed. As a post hoc test, a Wilcoxon test with set {αi, αj} with α as ROM for the Range of mobility session or the percentage of time spent in each RULA area of handling session, and i ϵ {Control, Belt, Exo} and j ϵ {Control, Belt, Exo}, i $\neq$ j was computed. For statistical analysis, we defined the null hypothesis as no significant difference between the conditions. The null hypothesis is rejected if the p-value of the Wilcoxon test is less than 0.05. In this case, the alternative hypothesis (significant difference between the two conditions) was supported.

For EMG data and all three conditions, statistical analysis was performed on the normalized RMS values, which are a good indicator of the magnitude of the muscle activity [25]. For all muscles, Friedman tests, as nonparametric tests, were performed with the set {RMSControli, RMSBelti, RMSExoi} with i ϵ {LLD, RLD, LRA, RRA, LOE, ROE}. The null hypothesis was no significant difference between the three conditions. The null hypothesis is rejected if the *p*-value (level of significance) of the Friedman test is less than 0.05. In this case, the alternative hypothesis (significant difference between the three conditions) was supported, and a post hoc test can be performed. As a post hoc test, a Wilcoxon test with set {RMSi, RMSj} with i ϵ {Control, Belt, Exo} and j ϵ {Control, Belt, Exo}, i $\neq$ j was computed. We performed the same statistical analysis as previously conducted.

All calculations were performed with MATLAB R2022a software.

3. Results

3.1. Kinematics Analysis

The following figure (Figure 4) represents the percentage of time spent in each RULA area for flexion rotation angle of sessions FREE LIFTING and SQUAT LIFTING under the three conditions (Control, Belt, and Exo).

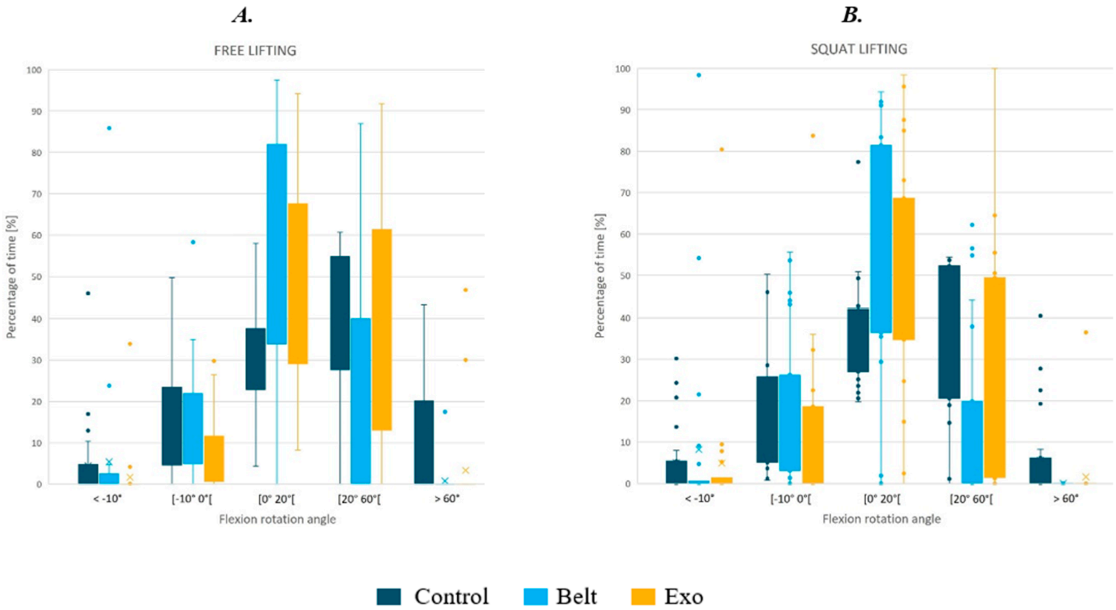

Figure 4. (**A**) represents the time spent in flexion rotation ROM during the FREE LIFTING session. (**B**) represents the time spent in flexion rotation ROM during the SQUAT LIFTING session.

We noticed the difference in the time spent in RULA areas for flexion rotation angle for the FREE LIFTING and the SQUAT LIFTING session according to the conditions (Control, Belt, Exo) confirmed by the Friedman test (Table 1).

Table 1. *p*-values of the Friedman tests of the difference of the time spent in RULA areas for flexion rotation angle for the different conditions for the FREE LIFTING and the SQUAT LIFTING sessions. * indicates that the null hypothesis is rejected (*p*-value < 0.05), and therefore, there is a significant difference between the three conditions.

RULA Area	$<-10°$	$[-10°\ 0°[$	$[0°\ 20°[$	$[20°\ 60°[$	$\geq 60°$
FREE LIFTING	0.0248 *	0.0366 *	0.0046 *	0.0125 *	0.0036 *
SQUAT LIFTING	0.0208 *	0.1767	0.0091 *	0.0026 *	0.0062 *

Post hoc tests corroborate these observations (Tables 2 and 3). Regarding the FREE LIFTING, the Belt tends to decrease the time spent >60° and increase the time spent in the area [0° 20°[compared to the Control group. The Exo tends to decrease the time spent in extension (<0°) and in flexion >60° and increase the time spent in the area [0° 60°[. The same assessment can be made for the SQUAT LIFTING.

Table 2. *p*-values of Wilcoxon tests to compare the different conditions for the FREE LIFTING session. * indicates that the null hypothesis is rejected (p-value < 0.05), and therefore, there is a significant difference between the conditions.

RULA Area	<−10°	[−10° 0°[	[0° 20°[	[20° 60°[	≥60°
Control vs. Belt	0.3894	0.8005	0.0074 *	0.0068 *	0.0046 *
Control vs. Exo	0.0131 *	0.0304 *	0.0095 *	0.9825	0.0276 *
Belt vs. Exo	0.0942	0.0490 *	0.4290	0.0110 *	0.5380

Table 3. *p*-values of Wilcoxon tests to compare the different conditions for the SQUAT LIFTING session. * indicates that the null hypothesis is rejected (p-value < 0.05), and therefore, there is a significant difference between the conditions.

RULA Area	<−10°	[−10° 0°[	[0° 20°[	[20° 60°[	≥60°
Control vs. Belt	0.3894	0.8005	0.0074 *	0.0068 *	0.0046 *
Control vs. Exo	0.0131 *	0.0304 *	0.0095 *	0.9825	0.0276 *
Belt vs. Exo	0.0942	0.0490 *	0.4290	0.0110 *	0.5380

For the LOAD TRANSFER session, the percentage of time spent in each RULA area of flexion, lateral bending, and axial rotation angles was analysed (Figure 5).

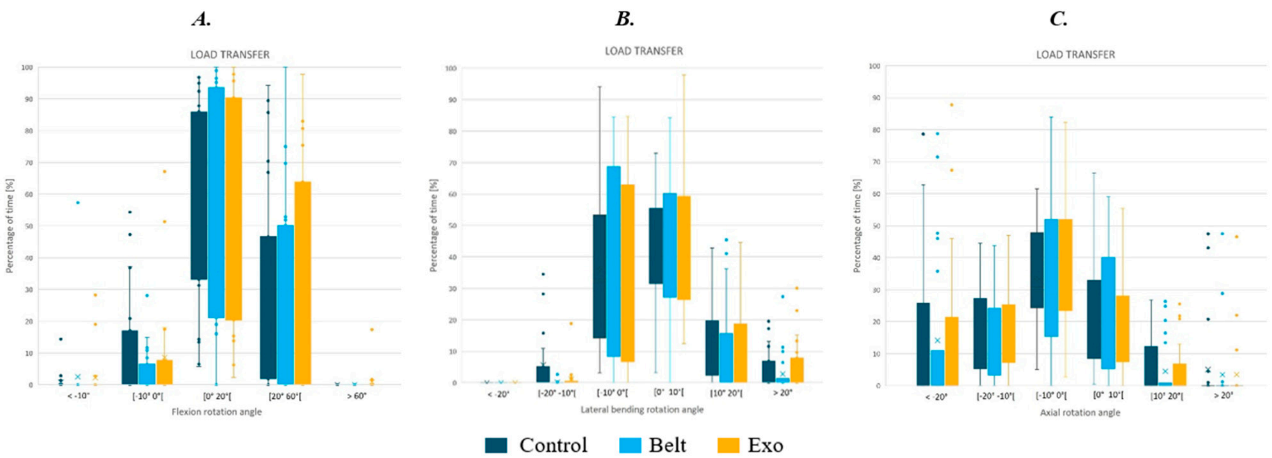

Figure 5. (**A**) is the time spent in flexion rotation angles during the LOAD TRANSFER session. (**B**) is the time spent in lateral bending rotation angles during the LOAD TRANSFER session, and (**C**) is the time spent in axial rotation during the LOAD TRANSFER session.

For flexion rotation angle and axial rotation, we noticed no difference between conditions confirmed by the Friedman test. We observed only a significant difference in the lateral bending rotation angle of the LOAD TRANSFER session for the range [−20° −10[and [10° 20°[with a p-value of the Friedmann test of 3.55×10^{-5} and 0.0655, respectively. On the one hand, the Wilcoxon post hoc test concluded a difference for the two RULA areas between [−20° −10°[and [10° 20°[for Control vs. Belt with a p-value of 9.71×10^{-4} and 0.0272. On the other hand, the Wilcoxon post hoc test concluded only a significant difference for Control vs. Exo for the range [−20° −10°[with a p-value of 0.0218.

3.2. EMG

The following figure (Figure 6) represents the RMS of the EMG of the six muscles recorded under the three conditions (Control, Belt, Exo) for the three handling sessions: 'FREE LIFTING', 'SQUAT LIFTING', and 'LOAD TRANSFER'.

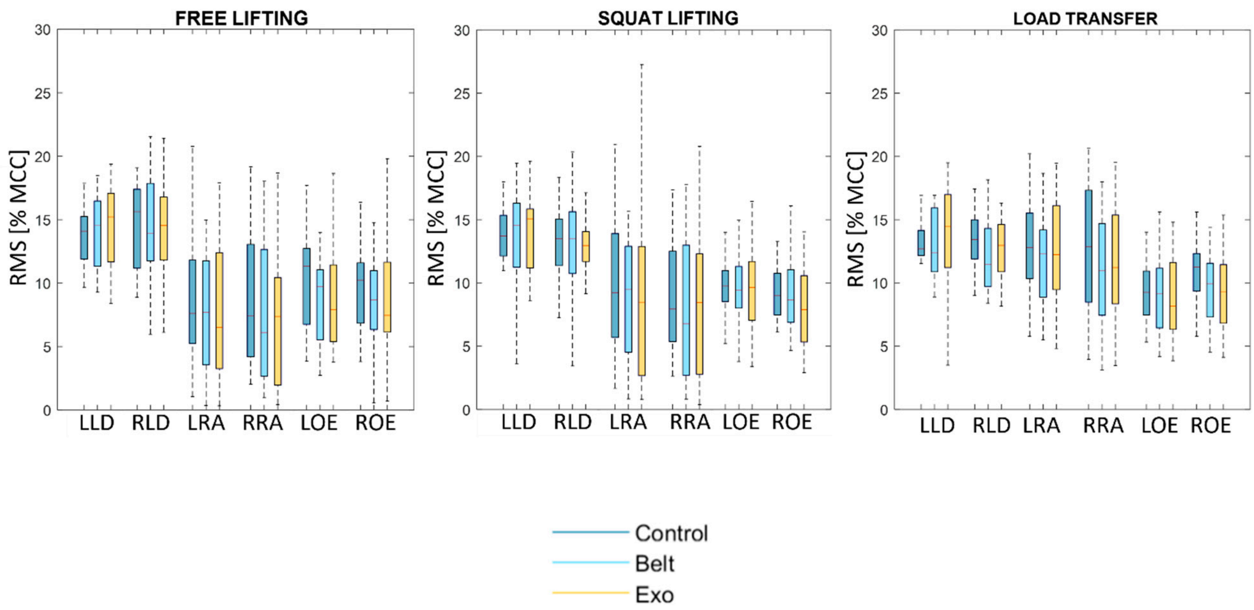

Figure 6. RMS values in percentage of maximum of the "Control" condition (MCC) of EMG for the left *Longissimus dorsi* (LLD), right *Longissimus dorsi* (RLD), left *Rectus Abdominis* (LRA), right rectus abdominis (RRA), left *Obliquus Externus abdominis* (LOE) and right *Obliquus Externus Abdominis* (ROE) during the FREE LIFTING, SQUAT LIFTING, and LOAD TRANSFER sessions.

We noticed little difference in RMS values according to the conditions supported by the Friedman test for all muscles (Table 4).

Table 4. *p*-values of Friedman test on RMS of EMG on right and left *Longissimus dorsi* (LD), right and left *Rectus Abdominis* (RA), and right and left *obliquus externus abdominis* (OE) to compare the different conditions for the Handling simulation session. * indicates that the null hypothesis is rejected (p-value < 0.05), and therefore, there is a significant difference between the conditions.

RMS	LeftLD	RightLD	LeftRA	RightRA	LeftOE	RightOE
FREE LIFTING	0.4013	0.5682	0.0147 *	0.0429 *	0.0363 *	0.0122 *
SQUAT LIFTING	0.8404	0.0175 *	0.3480	0.0657 *	0.1787	0.0351 *
LOAD TRANSFER	0.0175	0.1188	0.0013 *	7.8×10^{-5} *	0.2765	0.0074 *

However, these differences were not confirmed by post hoc tests. Consequently, according to these results we cannot conclude that the difference is statistically significant.

4. Discussion

The goal of this study was to quantify the biomechanical impact through the kinematics and muscle activity of an active lumbar exoskeleton. For this purpose, the exoskeleton was divided into a passive part (textile belt) and an active part (complete exoskeleton with SEA). During handling, our results showed that both the passive and active parts of the exoskeleton reduced the time spent in the risk range of motion, as defined by the RULA method [29]. The passive belt tends to greatly reduce the range of motion of flexion and lateral bending angles. Both devices seem to be transparent on muscle activity. These observations confirm our hypothesis on the kinematics but not on the level of muscle activity by the exoskeleton.

In contrast to the Belt condition, the active exoskeleton tested by the "Exo" condition maintains good trunk mobility while decreasing maximal amplitudes compared to the "Control" condition. Such results have already been observed for lumbar belts [43] and exoskeletons [41]. In fact, common lumbar belts were designed to increase back stability, which results in a reduced range of motion [43]. The current exoskeleton is composed of a passive part (textile belt) and an active part (complete exoskeleton with SEA). Our study demonstrates that the impact of the active part of the exoskeleton, tested during the Exo session, is to increase the range of motion in comparison with the "Belt" condition while preventing the possibility of extreme angles. In handling tasks, such extreme angle positions are known to increase the risk of musculoskeletal disorders [30]. In the 'LOAD TRANSFER' session, which is a complex task, we observed that the trunk moved in all three anatomical planes. By the "Control" condition, we also showed that the range of motion of the trunk is reduced by default and already below the so-called risk range defined by RULA, and therefore, we did not observe any modification of the time spent in the other RULA range of motion areas. This could contribute to the kinematic transparency of exoskeleton design [44] and the acceptance of exoskeletons in daily activities [5,14].

Our results showed that RMS values of EMG remained in the same order of magnitude in all conditions (Control, Belt, and Exo). These observations suggest that the exoskeleton does not interfere with the muscle activity of the operators. The effect of the exoskeleton on muscle activity reduction is still controversial [32,33]. Some studies have shown a decrease in muscle activity in the area of the body supported by the exoskeleton [45,46], while others have shown only minor changes [47] or relatives of the exoskeleton design [32]. The clinical relevance of reductions in muscle activity in relation to the prevention of musculoskeletal diseases remains an open question [47,48]. Some authors advocate for reducing muscle activities to prevent intervertebral disc compression [46], while others endorse muscle activities to prevent injury by enhancing trunk stability [49] or to contribute to the rehabilitation process of low back pain [50]. Furthermore, muscle activity is related to the difficulty of the task [51]. In our case, the load to be handled was 5 kg, requiring a light effort with a low-risk lifting situation [11]. In comparison, in Luger's study [51], where the load handling was 9.6 kg, which is considered a medium-risk lifting situation [11], the exoskeleton resulted in a 10.5% reduction in back muscle activity. Consequently, in our case, we observed only a slight reduction in muscle activity. This observation does not contradict the primary design objective of the exoskeleton used in this study, which aimed to apply an external traction force to the lumbar spine through SEA rather than reducing muscle activities to mitigate intervertebral disc compression [52].

Our study has some limitations. Most biomechanical studies on exoskeletons have been conducted on passive exoskeletons due to the prevalence of passive rather than active exoskeletons in industrial settings [41,53]. Therefore, a direct comparison with the results of passive exoskeleton studies is limited. Our study was conducted in a laboratory environment, and it would be relevant to conduct a study under real working conditions to analyse kinematic and muscular data over a longer period of time (e.g., one day) and under productivity constraints that may require faster movements. In addition, optical motion capture with markers, which is the gold standard for human motion analysis, has data acquisition limitations and limits the range of handling manoeuvres. Moreover, the RULA range of motion areas should be used only as a rough indicator, as it is designed for postural analysis in quasi-static situations and not for rapid, significant changes in posture [54]. Finally, we only investigated asymptomatic subjects, and it is known that the benefits of exoskeletons are less visible in symptomatic individuals [14].

5. Conclusions

This study highlights the effects of the active and passive components of the exoskeleton, which allows the trunk to preserve its range of motion while reducing the time spent in posture considered dangerous for the back according to ergonomic assessments, i.e., the RULA method. In the situation studied, the exoskeleton did not modify muscle activity in the abdominal–lumbar region. Our study contributes to defining the optimal prescriptions for the utilization of active exoskeletons as medical devices. Therefore, further studies should be conducted to evaluate the potential muscle atrophy or overstrain in this area. Future studies conducted under real working conditions, over a longer duration, and with higher load conditions would bolster these findings and offer a deeper understanding of the productivity of operators using exoskeletons. Furthermore, the impact of exoskeletons on the prevention of musculoskeletal disorders (MSD) remains open and requires dedicated investigations.

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Institutional Review Board Statement: All subjects gave their informed consent for inclusion before they participated in the study. The study was conducted in accordance with the Declaration of Helsinki, and the protocol was approved by the Ethics Board of the centre of excellence for human and animal movement biomechanics of UTC (#2022-1).

Informed Consent Statement: Informed consent was obtained from all subjects involved in the study.

Data Availability Statement: The data presented in this study are available on request from the corresponding author. The data are not publicly available due to privacy concerns.

Conflicts of Interest: Mélissa Moulart is an employee of the Japet Medical Devices© company, which designed the Japet.W device. As Mélissa Moulart is Phd student, this study was carried out according to the rules of good practice and with full knowledge of the ethical rules under the supervision of academic partners according to the agreement of the academic ethical committee as defined by the French legislation on scientific integrity (Décret n° 2021-1572 du 3 décembre 2021). The other authors declare no conflicts of interest.

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