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ORIGINAL ARTICLE

Analysis of Active Back-Support Exoskeleton During Manual Load-Lifting Tasks

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Abstract

Purpose Manual material handling (MMH) tasks are the main causes of injuries and work-related musculoskeletal disorders in the industry. For preventing such disorders, several exoskeletons have been introduced to assist workers in performing MMH tasks. This study investigates the effect of an active back-support exoskeleton on muscle activity and analyzes its ergonomic effects.

Methods A custom-made active back-support exoskeleton consisting of two joints (lumber and hip) and three links (trunk, pelvis, and thigh) was used in this study. The ergonomic effect of this exoskeleton on manual load-lifting tasks was investigated by (1) analyzing the muscle activities of the lumbar erector spinae (LES) and upper trapezius using electromyography; (2) conducting the timed up and go (TUG) test; and (3) evaluating the subjective aspect (perceived discomfort). Eighteen healthy subjects participated in the experiment by performing load-lifting tasks and undergoing the TUG test. Thereafter, their perceived discomfort was assessed using the Borg scale.

Results Significant differences were observed with and without the exoskeleton in the (1) root mean squares of the right LES ($p = 0.006$) and left LES ($p < 0.001$), (2) time spent in the TUG test ($p < 0.001$), and (3) perceived exertion level ($p < 0.001$). The active back-support exoskeleton used in this study was effective in reducing muscle activity and risk related to the LES during manual load-lifting; however, problems regarding its usability arose because of its weight.

Conclusion The exoskeleton evaluated in this study can aid in reducing the load on the lumbar spine of workers by decreasing the muscle activity of the LES. From the usability perspective**,** users spent more time performing the tasks and perceived higher exertion levels while wearing the exoskeleton.

Keywords Exoskeleton · Electromyography · Lumbar · Manual load-lifting · Perceived comfort

1 Introduction

Manual material handling (MMH) tasks remain common in most industries despite increasing trends toward automation. These tasks can significantly load a worker's lumbar spine, especially when the worker is in static and forward bending postures, and are the main cause of lower back injuries and occupational disorders [1–3]. The most severely affected part is the lower back region, accounting for 30% of all work-related injuries in Europe from 1999 to 2007 [4]. Work-related musculoskeletal disorders (WMSDs) considerably increase industrial costs and affect the quality of life of workers [5]. Accordingly, these problems are among the foremost reasons that have led industries to pursue full automation. However, total automation is not always feasible [6]. Even if robots are introduced into the workplace, full automation remains limited unless all related processes are planned and designed from the outset. Consequently, manual tasks, such as the loading and unloading of raw materials or intermediate products, continue to be performed by humans and are ubiquitous.

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In recent years, to prevent WMSDs, wearable devices, called exoskeletons, have been introduced. For MMH tasks, in particular, one of the solutions is to utilize exoskeletons that exert external assistive forces to support the worker's body [7]. The back-support exoskeleton can reduce the lumbar compression force by reducing the muscle activity around the spine [5]. Exoskeletons are categorized as "active" and "passive" [6]. Active exoskeletons consistently include one or more actuators (pneumatic actuators, hydraulic actuators, or electric motors) that supply external power to the body by actuating the joints. In contrast, passive exoskeletons are simple and have no external power; they employ materials or springs to store energy to assist in the physical movements of workers. The implementation of passive exoskeletons compared with that of active exoskeletons in industries is generally less complex, whereas the latter exoskeletons are costly and heavy [8]. Nevertheless, active exoskeletons are more versatile and have a wider range of applications. Many back-support exoskeletons have been developed to assist workers. Table 1 summarizes the names of various exoskeletons, their types, tasks performed, and the exoskeletal effects.

Many active exoskeletons have been mainly developed and tested in the laboratory. In contrast, in this study, a custom-made active exoskeleton for assisting in lifting tasks has been developed. It consists of two joints (lumbar and hip) and three links (trunk, pelvis, and thigh). For ease of use, a quick-release strap is attached around the wearer's shoulders, waist, and thighs. For consuming relatively low mechanical power, two series elastic actuators (SEA) with a Bowden cable transmission are used. The exoskeleton devised in this study compared with previously developed active exoskeletons differed in terms of the design concept. For example, the active pelvis orthosis (APO) only considered the hip joint, the hybrid assistive limb (HAL) consisted of pelvis and thigh straps, and the Hyundai waist exoskeleton (H-WEX) used a wire-driven singular actuation.

Previous studies on exoskeletons have mainly focused on musculoskeletal risks. One of the most common methods employed to identify musculoskeletal effort and muscle activity is electromyography (EMG) analysis [9]. Some of these studies have aimed to verify the efficacy of exoskeletons through EMG [10–12] and measure the flexion angle [13]. Most of them have been conducted in the laboratory through experiments because EMG signals are easily affected by the environment; hence, directly verifying field test findings is exigent [7]. As a result, only a few studies [14–18] have examined the use of exoskeletons in the field.

For exoskeletons to successfully fit and be comfortably used in the workplace, they must have sufficient usability [19]. Several studies have considered ergonomic and usability aspects, such as discomfort [16, 20, 21], user acceptance [22], muscular demand [17], and physiological benefits [23] to develop passive exoskeletons. The current work accounts for the perceived exertion required of users due to the weight of the device and the time necessary to move from one working position to another.

The primary objective of this study is to analyze the effect of active back-support exoskeleton using EMG, timed up and go (TUG) test duration, and perceived exertion scores. Postures that are associated with high-risk injuries during load-lifting are first identified. Thereafter, the effects of active exoskeletons according to those postures are investigated. The muscle activities of the lumbar erector spinae (LES) and upper trapezius (UT) with and without the aid of an exoskeleton robot were analyzed using EMG. Furthermore, this work analyzed the usability of exoskeletons considering both objective (TUG test duration) and subjective (perceived exertion) perspectives.

Table 1 Review of previous research [decrease in muscle activity (↓), increase in muscle activity (↑), passive (P), active (A)]

Name	Type	Task	Exoskeleton effects
backX [24]	P	Stooping, lifting, bending or reaching	Thoracic erector spinae↓, lumbar erector spinae↓
Laevo [12]	P	Static bending	Iliocostalis↓, longissimus pars lumborum↓
PAS-Suite [15]	P	Digging	Lumbar erector spinae↓
PLAD [25, 26]	P	Static bending and lifting	Thoracic erector spinae↓, lumbar erector spinae↓, gluteus maximus↓, biceps femoris↓
APO [11]	A	Bending and lifting tasks	Thoracic erector spinae↓, lumbar erector spinae↓, erector spinae iliocostalis↓, biceps femoris↓, rectus femoris↑
HAL [27]	A	Repetitive lifting	Thoracic erector spinae↓, lumbar erector spinae↓
H-WEX [28]	A	Static bending and lifting	Erector spinae↓, gluteus maximus↓, biceps femoris↓
Mk2 [29, 30]	A	Lowering and lifting	Erector spinae↓, biceps femoris↓
Robo-Mate [19]	A	Dynamic lifting and lowering	Lumbar erector spinae↓, biceps femoris↓
WSAD [31]	A	Stooping	Thoracic erector spinae↓, lumbar erector spinae↓, latissimus dors↓i, rectus abdominis↓

2 Target Task Selection

The target tasks that may cause musculoskeletal disorders were selected through field observations and biomechanical analysis. Nine different postures that were expected to exert considered loads on the LES were selected (Fig. 1) by observing various MMH tasks in a factory that manufactured automated pharmacy dispensing systems. Four human factor researchers (average research experience: 4.3 years) participated in the field observations. Among all tasks involving load-lifting and lowering, they selected those that considerably bent the back and legs as well as those deemed to exert a substantial amount of load to the lumbar. Postures similar to lifting and lowering objects but required a different flexion of the knees were classified as separate postures because the burden on the lumbar area differed. The observation-based method is widely used to evaluate musculoskeletal disorders; however, this approach is prone to evaluator subjectivity [32].

Accordingly, this study implements additional biomechanical analyses using a 3D model to improve objectivity in the target task selection. Various mathematical models based on the joints and the body's total mass have been applied to analyze movements [33–36]. Moreover, commercial software for digital human modeling has been developed to calculate movements [37, 38]. The Three-Dimensional Static Strength Prediction Program (3DSSPP) (version 6.0.6), a commercial software developed by the University of Michigan, USA, was used for this purpose. For the anthropometric data, two different datasets were used: the American 50th percentile and the Korean 50th percentile. For the former, 175.1 cm and 83.9 kg were used, whereas 165.0 cm and 63.4 kg were used for the latter. The 3DSSPP analysis showed that three of the postures (i.e., P2, P4, and P8) had high injury risks based on both the American and Korean cohorts. Two postures (P3 and P7) potentially had high injury risks considering the American cohort. Thus, the activities in the bending and sitting positions were found to require further experiments. In the laboratory experiments, an active back-support exoskeleton for these two positions was tested.

The detailed procedures for the field observation and biomechanical analysis are summarized in the Appendix.

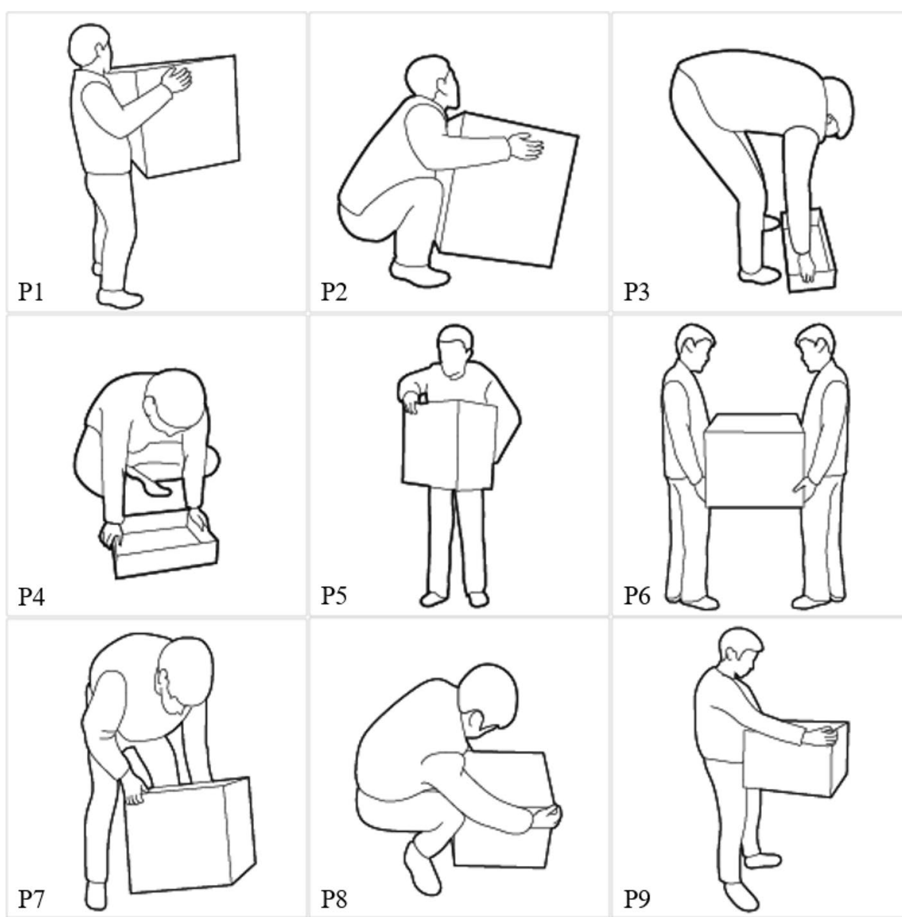

Fig. 1 Nine lifting postures (P1, standing and carrying load with one hand on top; P2, full squatting and starting to lift load; P3, lifting load from shelf located below and stooping with slightly bent knee; P4, knee bent at 90° (similar to P3); P5, lifting and supporting load with force of both arms and shoulders; P6, carrying load in pairs; P7, placing or removing load (lower shelf); P8, sitting with fully bent knee; and P9, standing with load by one arm.)

3 Method

3.1 Participants

Eighteen healthy and physically active Korean university students (11 male and seven female) participated in this study. The mean and standard deviation values of their age, height, and weight are listed in Table 2. This study has been approved by the Institutional Review Board (IRB approval: RERI-IRB-190425-1).

3.2 Apparatus

The exoskeleton [39] in this study is a custom-made prototype for assisting in lifting tasks (Fig. 2) consisting of two joints (lumbar and hip) and three links (trunk, pelvis, and thigh). For ease of use, a quick-release strap is attached around the wearer's shoulders, waist, and thighs. Four degrees of freedom were set for the physiological movement of the lumbar and hip joints (i.e., flexion/extension and abduction/adduction) during lifting and walking. Two SEAs were used for shock tolerance, comfortable interaction with the user, and energy storage and release [40–43]. The actual weight of the exoskeleton was 4.4 kg, excluding the high-level controller and power supply, which were off-board units. The force exerted by the exoskeleton to assist the user depended on the change in the inclination of the trunk and pelvis [43]. As shown in Fig. 3, three states are defined: standing, lowering, and lifting. These states were determined by the angle of the pelvic joint and the inclination of the trunk. In the upright standing state with the knees straight, the motor was deactivated, and the desired output force from the two SEAs was set to zero. In the lifting state, the motor was activated to maintain maximum force, whereas in the lowering state, the assistive force was only generated by the spring and changed linearly with the angle of the pelvic flexion. At maximum bending (corresponding to a

Table 2 Data of participants

Gender	Sample size	Age	Height (cm)	Weight (kg)
Male	11	25.3 (±3.8)	175.0 (±3.5)	72.8 (±15.1)
Female	7	22.3 (±2.2)	163.1 (±5.3)	55.9 (±7.0)
Total	18	24.1(±3.5)	170.4 (±7.2)	66.3 (±14.9)

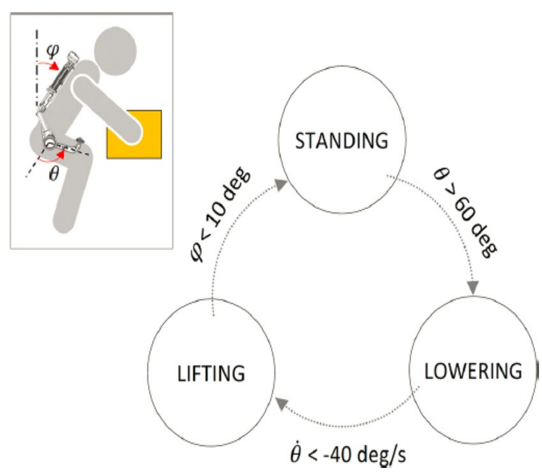

Fig. 3 Three states during task performance (standing, lowering, and lifting)

Fig. 2 Exoskeleton used in study

maximum spring displacement of 75 mm), a 1023-N force was expected to be generated by the spring with increases in the force on the back and each thigh of 68.2 and 78.2 N, respectively. Note that in the TUG test, the standing and lowering states alternated; however, the motor was deactivated, and the spring's assistive force was irregularly exerted. In a previous study [39], one healthy male participant performed load-lifting tasks. The study confirmed the tendency of the exoskeleton to reduce the muscle activation level of the erector spinae and rectus abdominis muscles.

3.3 Experimental Design

Anthropometric data (height and weight), as well as age and gender data, were collected before the experiment was performed; the experimental procedure was also explained to each participant. The EMG sensors were attached to the left and right LES (LLES and RLES, respectively) and UT (LUT and RUT, respectively) of each participant (Fig. 4) [44]. Note that the sensor could not be attached to the rectus abdominal muscles due to the exoskeleton strap. The location where the sensor was attached was carefully swabbed by the experimenter with alcohol. Measures regarding EMG are explained in Sect. 3.5.

To minimize the effect of the test sequence, half of the subjects participated in the experiment wearing the exoskeleton first and then participated without wearing the exoskeleton. The other half participated in the reverse order. The participants underwent the TUG test before performing the load-lifting task. For relaxing their muscles, a break was included in between tasks.

Thereafter, the perceived full-body exertion was assessed using the Borg scale (6–20) [45]. The process was completed for approximately 1 h.

3.4 Tasks

In the experiment, (1) the load-lifting task was performed to analyze the effects of the active exoskeleton, and (2) the TUG test and subjective evaluation were conducted to analyze the usability of exoskeletons. Both tests were performed with and without the exoskeleton. First, the lifting of the 10-kg (two 5-kg dumbbells) load from the ground to the top of the table was repeated three times per minute. Dumbbells were employed to control and generalize the experimental tasks facilely and to distribute the loads to each hand equally. The participants were required to lift the dumbbells from the ground from a sitting posture with knees fully bent; once upright, they were to place the dumbbells on top of the table (Fig. 5). The sitting posture with knees fully bent was included in the lifting task because sitting postures (P2, P4, and P8 in Fig. 1) had high injury risks. The dumbbells were naturally lifted by the subjects according to their shoulder width and raised to approximately the same height. In addition, they were required to raise the load with their hands to a position no higher than the table. The distance between the position of the dumbbells and the subject's feet was 10 in; each position was marked. This task was selected because it involved both sitting and full bending positions. According to the National Institute of Occupational Safety (NIOSH) Lifting Equation, The lifting index was set as 1, indicating a nominal risk to the participants associated with

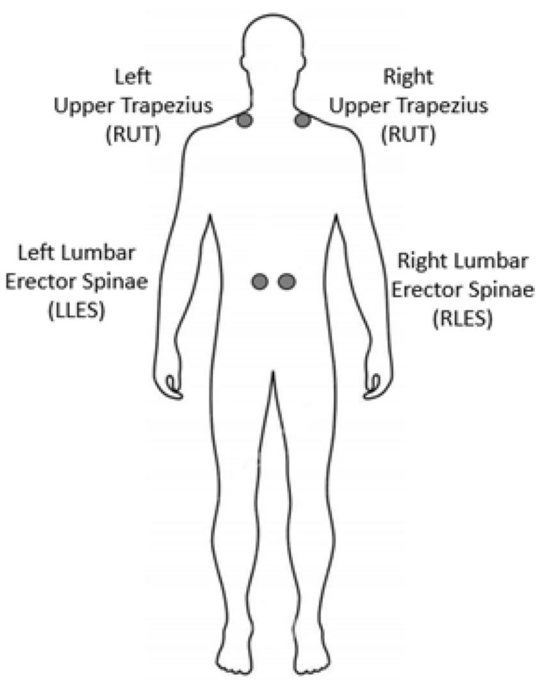

Fig. 4 Position of EMG sensors

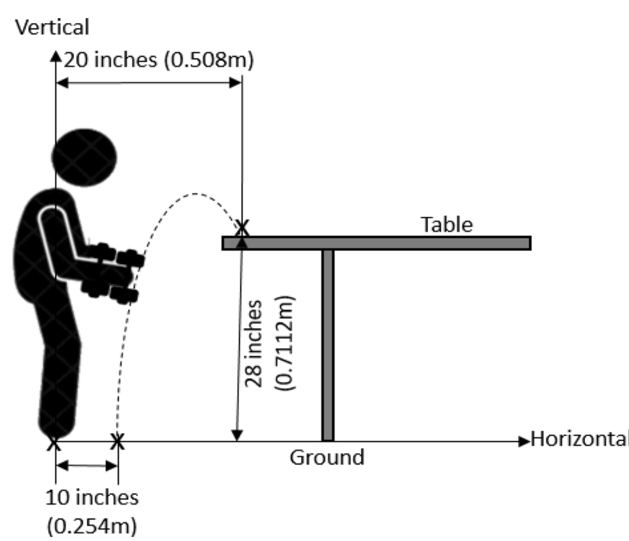

Fig. 5 Example of load-lifting task

the lifting tasks. The NIOSH Lifting Equation defines the acceptable weight for lifting tasks to minimize the potential of workplace back injuries [3]. The heights of the start and endpoints were set as 0 and 28 in (0.7112 m), respectively, whereas the horizontal distance was given as between 10 (0.254 m) and 20 in (0.508 m) (Fig. 5).

After the resting period, the TUG test and subjective evaluation were performed. The TUG test was implemented to assess the mobility of participants with and without an active exoskeleton. The test duration started from a seated to a standing position, walking a distance of 3 m, turning at the return point, walking back to the starting point, and sitting; the participants did not carry any load (Fig. 6). They were directed to walk normally and avoid rushing toward the turning point. The TUG test was so designed because such movements in between tasks always occurred in the workplace. After the TUG test, the subjective evaluation was conducted using the Borg scale (6–20).

3.5 Data Collection and Pre-processing

A four-channel wireless EMG (TeleMyo Desktop Direct Transmission System, Noraxon U.S.A. Inc., Scottsdale, AZ, USA) was utilized to measure the EMG signals. Signals from the LES and UT were acquired from the left and right sides. For the TUG test, the duration for completing the task was measured. After completing the TUG test and lifting task, the perceived full-body exertion, with and without the exoskeleton, was assessed using the Borg scale (6–20).

The maximum voluntary contraction (MVC) was measured at low muscle fatigue. To measure the MVC of the LES, the experimenter pressed down on the subject's upper back, pelvis, and ankles as the upper body prone on the mat was raised. This position is based on the Bierin–Sorensen test; the foregoing is known to be sufficient for obtaining the MVC of the LES [46]. For measuring the MVC of the UT, the subject sat on a chair, and the experimenter pressed down on top of the shoulders as the subject lifted the arm and upper body [47]. Note that for extracting adequate MVC values, each participant performed three MVC trials for which the duration did not exceed 3 s. In performing the MVC tasks, muscle fatigue was minimized by resting for at least 2 min in between trials.

All the UT and LES muscular activations of participants were analyzed using the EMG signals. The pre-processing of EMG was implemented as follows. (1) The EMG signals were filtered at the (20–450)-Hz bandwidth at a sampling rate of 1500 Hz using a finite impulse response filter. (2) The signal from the heart was eliminated by electrocardiogram reduction. (3) The signal was changed to a positive value by rectification. (4) The EMG signals were smoothed using root mean squares (RMSs). The RMSs were based on a 100-ms window and used as the reference parameter for the analysis, which was performed without overlaps. In the case of % MVC, the values for each muscle were also refined by the aforementioned process. Thereafter, the relative MVC value was derived as the final normalization value.

3.6 Statistical Analysis

A paired t-test using the EMG data was implemented with the IBM SPSS statistics 26 to verify the exoskeletal effect. The statistical significance of this effect on the TUG test duration and perceived exertion were compared by repeated measures analysis of variance (RMANOVA) ($p < 0.05$). Note that a parametric test was performed by confirming that the result data was normally distributed. The dependent variables were the normalized RMS (% MVC) for the selected muscles, TUG test duration (s), and perceived exertion (score); the independent variable was the use or non-use of the exoskeleton.

4 Results

4.1 Exoskeletal Effect

4.1.1 EMG Analysis

In the EMG analysis of the load-lifting tasks for the LES, a significant difference in the RMS between RLES ($p = 0.006$) and LLES ($p < 0.001$), with and without the exoskeleton, was

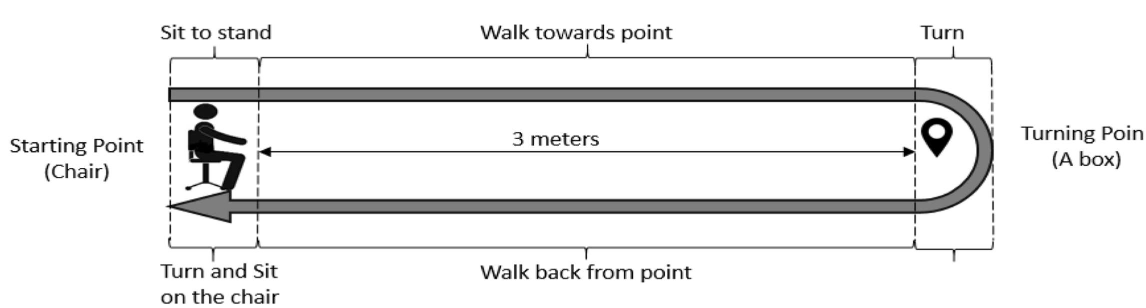

Fig. 6 Timed up and go test design

observed (Fig. 7a). The RLES showed a significant reduction of approximately 11% in the normalized RMS using the exoskeleton, whereas the LLES exhibited a reduction of approximately 16%.

The EMG results of the LUT and RUT for the load-lifting tasks are shown in Fig. 7b; no significant difference between the left ($p = 0.096$) and right ($p = 0.187$) muscles is observed.

4.2 Mobility (TUG Test)

The TUG test was performed over an average of 11.5 s without the exoskeleton and 13.6 s with the exoskeleton. The RMANOVA indicates that a significant difference ($p < 0.001$) between the two groups (i.e., with and without the exoskeleton) (Fig. 7c) exists. This demonstrates that using the exoskeleton reduces the walking speed.

4.3 Perceived Exertion

The Borg scale (6–20) evaluation showed that the average perceived exertion level with the exoskeleton was 12.9 points (Fig. 7d). A 13-point level is interpreted as "somewhat hard" and 11 as "fairly light." The results of the RMANOVA indicated that a significant difference ($p < 0.001$) existed between the two conditions (i.e., with and without the exoskeleton).

5 Discussion

5.1 Exoskeleton Effect

The objective of this study is to investigate whether wearing an exoskeleton affects muscle activity and to analyze exoskeleton usability. The EMG activities for the LLES, RLES, LUT, and RUT were targeted. Further, the TUG test and perceived exertion analysis were implemented.

The EMG analysis for the LES showed that there was a significant reduction in the averaged value of the normalized RMS of RLES (11%) and LLES (16%) when the exoskeleton was used during load-lifting. These significant reductions for the left and right muscles show that the exoskeleton significantly decreases the necessity of using the lumbar vertebrae. Some passive exoskeletons also significantly reduced the muscle activity of the lumbar erector spinae. In the study of Bosch [10], a significant reduction in the muscle activities of the LES (35–38%) was observed

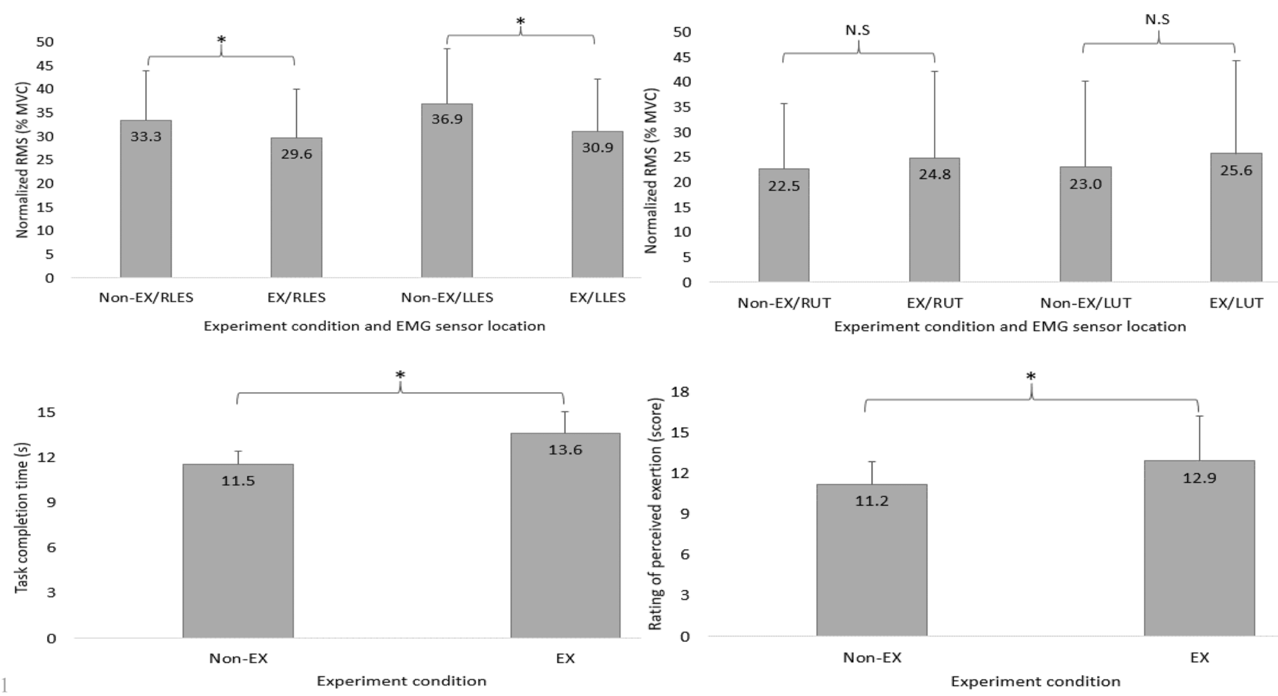

Fig. 7 Results of EMG analysis. **a** Averaged results of LES EMG analysis (upper-left); **b** averaged results of UT EMG analysis (upper-right); **c** timed up and go test results before and after wearing exoskeleton (bottom-left); **d** comparison of Borg scale scores before and after wearing exoskeleton (bottom-right). (*Non-EX* before wearing exoskeleton; *EX* after wearing exoskeleton; *LES* lumbar erector spinae; *EMG* electromyography; *RMS* root mean square; *MVC* maximal voluntary contraction; *RLES* right lumbar erector spinae; *LLES* left lumbar erector spinae; *UT* upper trapezius; *RUT* right upper trapezius; *LUT* left upper trapezius; *Significant difference; *N.S.* not significant)

when the Laevo exoskeleton was worn. In the work of Abdoli [25], a significant decrease (14.4%) was noted when the PLAD exoskeleton was utilized. The LES results of the present work concur with those of active exoskeletons proposed in previous studies. That is, the muscle activity of the LES during a lifting task was reduced using an active exoskeleton. According to Von Glinski [27], a considerable decrease in the muscle activity of the right LES (4.5%) was observed when a repetitive lifting task was performed using the HAL active exoskeleton [48]. Moreover, a significant normalized RMS reduction in the peak muscular activity (28–35%) at the lumbar spine was noted when an active exoskeleton was employed. Koopman [13] identified a 13.3% normalized RMS reduction in the lumbar EMG in the L5/S1 flexion–extension moments. Miura et al. [49] used a visual analog of the lumbar fatigue scale for the lifting task and found a 25% reduction in the subjective lumbar fatigue using HAL. A numerical difference between the results of this study and those of previous studies was found. However, these results confirmed that the exoskeleton significantly decreased the muscle activity in the LES. This reduction in the lower back muscle fatigue can aid in preventing various chronic lower back pain and WMSDs [50–52].

The EMG analysis for the UT showed that there was no significant difference in the reported values with and without the exoskeleton. However, one study [10] observed a significant reduction in the UT muscle activity in forward-bending tasks. The difference between the two studies may be attributed to the different design concepts of each exoskeleton and the variations in the tasks performed. In Bosch's study [10], the participants performed assembly and static holding tasks while bending at the waist, whereas in the current study, the participants performed a ground-to-table load-lifting task (Fig. 5). We can conclude that the load-lifting task does not seem to exert stress on the UT. In Fig. 7, the RMS value increases when the exoskeleton is worn; this may be attributed to the weight of the exoskeleton. According to Cook et al. [53], the EMG LES value increased with increasing weight or load. Further testing is necessary to determine the effect of the exoskeleton on the UT.

The TUG test was performed by considering the movement in between tasks in an actual factory. Previous studies on exoskeletons mainly considered cases in one setting. In Van Ham et al. [41], a walk-and-lift task was performed for the usability test; however, the amount of additional time required to walk with an exoskeleton was not verified. The field observations (Appendix) indicate that the workers not only performed lifting tasks but were also moving as they accomplished several tasks. When the exoskeleton was worn, the TUG duration increased (Fig. 7c); this was assumed to be attributable to the exoskeleton weight. According to Medley and Thompson [54], participants who could complete the TUG test in less than 20 s could also independently perform the necessary movements. Thus, although the walking speed with the exoskeleton compared with that without the exoskeleton slightly decreased, the subjects with the exoskeleton had no mobility problems.

The exertion level with the exoskeleton was analyzed via subjective evaluation using the Borg scale. In one study, the usability of an active exoskeleton was analyzed through a questionnaire with a 5-point scale [28]. Generally, the participants were found to be uncomfortable as they wore the exoskeleton while working. In this study, the perceived exertion between wearing and not wearing the exoskeleton was compared. The Borg scale score was 12.9 points ("somewhat hard") when the exoskeleton was worn and 11.2 points ("fairly light") when it was not worn. In other words, an increased exertion was perceived by the participants while wearing the exoskeleton. Although the EMG results for the LES decreased, the perceived exertion increased with the exoskeleton. It is possible that other parts of the body required greater muscle activity but were not checked in the experiment, or the exoskeleton was psychologically rejected by the participant.

In summary, the active back-support exoskeleton can potentially reduce occupational musculoskeletal risks related to the lower back, especially those associated with the LES in performing MMH tasks; however, its weight limits usability. The proposed exoskeleton can be used for assisting the lower back in lifting tasks by reducing the muscular activity of the lumbar spinae. However, because of the limited usability of exoskeletons, workers may feel reluctant to use them in the workplace. To encourage their use, the production of lighter exoskeletons is necessary. Compared with a passive exoskeleton, an active type is generally heavy; hence, to reduce weight, lightweight materials and actuators must be utilized. Moreover, in the case of actuators, it is important to consider low mechanical power consumption in generating the force. The exoskeleton is a wearable device with a strap; thus, it should be easy to wear, and the strap should comfortably fit the body. Continuous monitoring that does not interfere with the tasks of subjects has to be conducted to find means of increasing the comfort level of users while wearing the exoskeleton.

5.2 Limitations

Three limitations are noted to prevent the generalization of the study. First, the use of dumbbells instead of boxes in the study may subtly differ from the actual field situation. Previous studies generally used boxes [13, 27, 48, 49]. However, specifying the lifting of boxes that fitted the shoulder width of each user was not possible. Moreover, as mentioned in Sect. 3.4 (Tasks), this study specifies that the load must be raised according to the foregoing requirement. Second, there were only 18 participants, which was few to some extent, and gender was not balanced. In particular, further studies on the effects of wearing exoskeletons based on gender must be conducted. In general, during performing load-lifting tasks, men endure greater compression forces than women (approximately 640 N) [55]. Thus, the gender parameter is expected to produce differences in assessing the efficacy of exoskeletons. Third, all participants were in their twenties; hence, generalizing the results for the entire population is inaccurate. The device effect after using it for a long time was also not studied in this study. Accordingly, the conduct of further research considering subjects of various ages is necessary to enhance generalizability.

6 Conclusion

Sitting and bending postures during load-lifting may cause serious injuries to the lower backs of workers. The proposed exoskeleton in this study can aid in reducing the load on the lumbar spinae of workers by reducing the muscle activity of the LES. The muscle activity of the UT did not significantly differ between wearing and not wearing the exoskeleton. The use of the proposed active back-support exoskeleton can reduce the muscle activity of LES during load-lifting tasks. From the usability perspective (TUG test duration and perceived exertion), the users spent more time on the task and perceived higher exertion levels when wearing the exoskeleton; this possibly due to the self-weight of the exoskeleton. For enhancing the role of exoskeletons in the industry, the usability aspect should be investigated further. The findings of this study may serve as a basis for the manufacture of exoskeletons and further research.

Appendix: Biomechanical Analysis for Lumbar Load

Various mathematical models are available for analyzing human movements based on the joints and total mass of the human body [33–36]. Moreover, commercial software for digital human modeling has also been developed to calculate such movements [37, 38]. This study used 3DSSPP (version 6.0.6) to measure the lumbar compression force at the L4/L5 level for various tasks, such as lifting, pressing, pushing, and pulling [37, 56]. The lumbar compression force is calculated from the biomechanical model using the upper body weight, load, upper body flexion angle based on the sagittal plane, and back muscle strength in each static posture.

For the nine postures selected during the field observations, the lumbar compression force was calculated using 3DSSPP. Based on the photographs of the workers assuming the nine postures in the field, the body joint angles were simulated using the 3DSSPP program. For avoiding the subjective evaluation at this time, the body joint angle data were determined by the consensus of the four researchers who conducted the field observations.

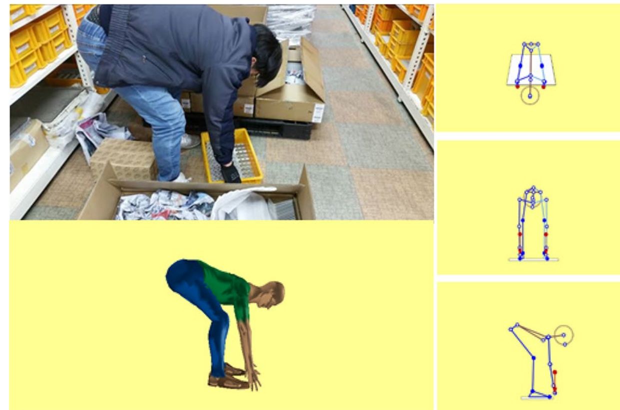

Fig. 8 Examples of Three-Dimensional Static Strength Prediction model (P3)

For the anthropometric data, two different datasets were used: the American 50th percentile and the Korean 50th percentile. For the former, 175.1 cm and 83.9 kg were used, whereas 165.0 cm and 63.4 kg were used for the latter. Figure 8 shows an example of the 3DSSPP model for P3. The human models are analyzed based on each task shown in Fig. 1, and the lumbar compression force at L4/L5 for each static posture is calculated, as summarized in Table 3. In this work, the analysis was performed using a 20-kg load equally distributed between both hands. Considering the NIOSH standards, the criterion for high-risk postures was whether the load on the lumbar spine (L4/L5) exceeded 3400 N, [43, 57]. The biomechanical analysis of the nine tasks confirmed that three (P2, P4, and P8) had high injury risks in both the American and Korean cohorts; two of them (P3 and P7) had high injury risks based on the American cohort.

Table 3 Lumbar compression force during task performance: lumbar compression force (LCF)

Posture	Position	LCF (N) based on American data	LCF (N) based on Korean data
P1	Standing	1833.7	1670
P2	Sitting	4737.5	4015
P3	Bending	3736.1	2981
P4	Sitting	4620.2	3722
P5	Standing	1589	1385
P6	Standing	915.4	804
P7	Bending	3912	3213
P8	Sitting	5328.5	4415
P9	Standing	1767.9	1625

The 3DSSPP analysis showed that the American workers might experience lower back injuries while lifting loads in the bending and sitting positions, whereas Korean workers might sustain injuries while lifting loads in the sitting position. Thus, the tasks in the bending (P3 and P7) and sitting (P2, P4, and P8) positions were found to require further experimental analysis.

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