A Lower-Back Exoskeleton With a Four-Bar Linkage Structure for Providing Extensor Moment and Lumbar Traction Force

Chaerim Moon, Jangho Bae, Jaewon Kwak, and Daehie Hong

***Abstract*— Lower back pain and related injuries are prevalent and serious problems in various industries, and high compression force to the lumbosacral (L5/S1) region has been known as one of the key factors. Previous research on passive lower back exoskeletons focused on reducing lumbar muscle activation by providing an extensor moment. Additionally, lumbar traction forces can reduce the compression force, and is a common treatment method for lower back pain in clinics. In this paper, we propose a novel passive lower back exoskeleton that provides both extensor moment and lumbar traction force. The working principle of the exoskeleton, extending the coil springs during lumbar flexion, and its design criteria regarding the amount of each force element were provided. The kinematic model explained its operation, and the dynamic simulation estimated its performance and validated its satisfaction with the design criteria. The biomechanical model provided a brief insight into the expected exoskeleton's effect on the reduced lower back compression force. Ten subjects performed static holding and dynamic lifting tasks, and the generated force elements in two directions, parallel and perpendicular to the trunk, were evaluated using a force sensor and electromyography sensors, respectively. The experiment demonstrated a pulling force opposite to the direction of intradiscal pressure and reduced erector spinae activation. This implies the effect of wearing the exoskeleton to decrease the intervertebral pressure during static back bending or heavy lifting tasks.**

***Index Terms*— Exoskeletons, four-bar linkage, surface electromyography, lower back pain, lumbar traction force.**

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I. Introduction

LOWER back pain (LBP) has been considered a significant problem in various industries, such as agriculture, construction, and manufacturing [1]–[4]. LBP is disadvantageous to both workers and industries, as workdays may be lost, leading to compensation costs [5], [6]. Even after recovery and return to work, workers report high recurrence rates, ranging from 60% to 78% [7]–[10].

The major risk factors of LBP in the workplace include handling heavy materials, maintaining a static posture, and repetitively bending or twisting the back [11], [12]. Specifically, these types of works possibly overload muscles, tendons, ligaments, and joints or increase disc degeneration with excessive compression [12], [13]. Among potential causes of LBP, NIOSH has determined compression force at the lumbosacral (L5/S1) region as one of the essential criteria of LBP during lifting-related tasks for two reasons: it is where the greatest amount of moment is exerted, and the tissues are vulnerable to force [14].

Various methods have been suggested to reduce the lower back compression force, including wearable types. Back support exoskeletons can be classified as active or passive. Active exoskeletons generate assisting forces with powered actuators, such as electrical motors. Although active exoskeletons provide strong support and varied assisting strategies, their weight, charging-related issues, and need for batteries can limit their use in industries or everyday life [15], [16]. Meanwhile, passive exoskeletons use non-powered mechanical elements to provide assisting forces; hence, passive exoskeletons can benefit certain situations more because of their relatively simple, compact, and lightweight characteristics.

Previous research on passive lower-back support exoskeletons has focused on reduced lumbar muscle activation. For example, the personal lift augmentation device (PLAD) [17]–[20] and the biomechanically assisted garment [21] utilized bands throughout the shoulders, back, and upper legs. These bands act like additional lumbar and upper leg muscles and allow less muscle activation. In addition, the devices reduce the intradiscal pressure with the longer moment arm of the bands compared to the spinal muscles. Other previous exoskeletons, such as Laevo [22]–[24] and the Bending Non-Demand Return (BNDR) [25], [26], possess spring-like elements placed around the hip joints, which provide extensor

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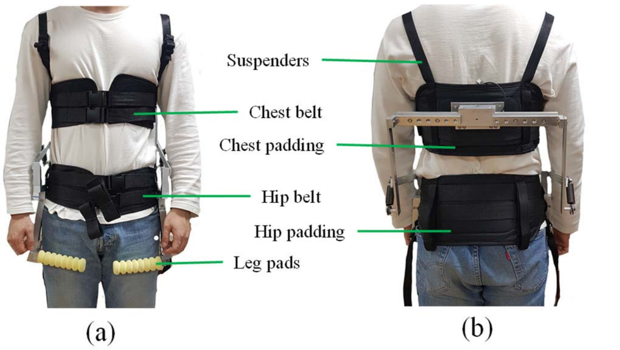

Fig. 1. The (a) front and (b) rear views of the exoskeleton, which is fixed to the human body by the chest and hip belts, and are adjustable to various body sizes. The exoskeleton was prototyped with aluminum, and its weight – except testing-related elements – is approximately 2.5 kg. During the 45-degree lumbar flexion, the extent of protrusion of the exoskeleton from the back is 10 cm and from the side is 3.5 cm.

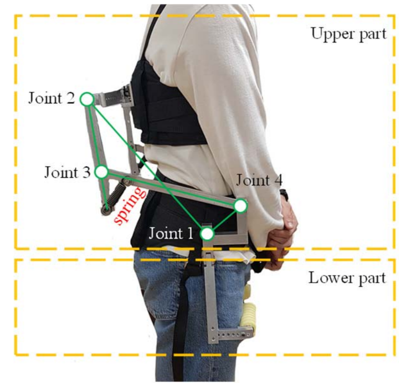

Fig. 2. The side view of the exoskeleton. The upper part includes a crossed four-bar linkage, of which the wearer's upper body is one of the links. It generates a force to the trunk according to lumbar and hip flexion angles. The lower part balances the moment of the system by mildly pushing the thighs.

moments and reduce the moment that the lower back muscles have to counteract on when bending, thereby reducing lower muscle activation and intervertebral pressure.

Although previous research has provided advantages in reducing the lower back compression force, they are limited to focusing on the back muscle force. Their assisting force is determined by the stiffness constant of elastic materials, and the use of a stiffer material will result in stronger support. However, excessively stiff materials may hinder human movement [25] and induce undesirable muscle activation of the trunk flexors (i.e., abdominal muscles) [24], [27]. This may limit lower back compression force reduction, thereby increasing the need of new strategies.

Lumbar traction is a common treatment method for lower back pain patients with proven mechanical effects, including increased intradiscal spaces and reduced prolapsed discs [28]–[31]. Lumbar traction alleviates lumbar disc pressure and relieves lower back pain [31]–[33]. Japet.^{W}, a motorized trunk exoskeleton, generates lumbar traction forces, reducing intradiscal pressure, as reported by numerical and cadaveric studies [34].

This research aims to suggest a novel back-support exoskeleton that reduces the compression force with two force elements: an extension moment generating force and a lumbar traction force. The device includes a four-bar linkage in which coil springs are installed. When a wearer bends its upper body, the spring is extended, generating the aforementioned forces to the lower back.

In this paper, we describe the design of a novel exoskeleton and its kinematic model. Its performance, based on lumbar flexion, was estimated through dynamic simulation. The biomechanical model estimated the effect of the force elements on the reduced lower back compression force. For experimental demonstration, the surface electromyography (EMG) signals of the abdominal and erector spinae muscles were measured while the traction force was measured using a force sensor.

II. Exoskeleton Design

A. Design Overview

We designed a passive lower-back support exoskeleton (Fig. 1) to provide two types of force elements (i.e., a traction force and a moment generating force) to the lumbosacral (L5/S1) region. It consists of upper and lower parts (Fig. 2), the upper part generates the force to the trunk with its crossed four-bar linkage system, and the lower part balances the moment equilibrium by pushing the thighs. It resembles the force distribution when placing one's hands on the knees and stretching the upper body. The exoskeleton works with lumbar or hip flexion. During lumbar flexion, the coil springs extend and generate force at joint 2, which can be distributed into traction and moment-generating forces. The spring force also mildly pushes the thigh, counteracting on a moment at joint 1. The exoskeleton also works with hip flexion in a similar manner.

Several design criteria regarding the amount of assisting force were determined: 80–100 N of traction force and 90–120 N of extension moment-generating force at 60-degree lumbar flexion. The moment-generating force would offload 5–7 kg of holding weight when the hand is 40 cm anterior to the L5/S1 level ($F_{offload} = F_{momentgenerating} * l_{L5S1\_to\_chestharness}/l_{L5S1\_to\_hands}$) [35]. The generated force depends on the spring stiffness and link length. With a selected coil spring (initial length, 75 mm; initial force, 68.6 N; spring constant, 11.7 N/mm), link lengths were determined to satisfy the design criteria. In addition to the four-bar linkage, the length of the link alongside the thigh was determined to limit the pushing force to < 30 N. Kinematic analysis and dynamic simulation were conducted to evaluate its operation and performance.

B. Kinematic Modelling

The crossed four-bar linkage of the exoskeleton (Fig. 3) can be analyzed through the equations of link lengths and joint angles. The following two equations describe the operation of the linkage:

$$l_1 cos\,(\theta_1) + l_2 cos(\theta_2) + l_3 cos\,(\theta_3) - l_4 cos(\theta_4) = 0 \quad (1)$$

$$l_1 sin\,(\theta_1) - l_2 sin(\theta_2) + l_3 sin\,(\theta_3) + l_4 sin(\theta_4) = 0 \quad (2)$$

When all the link lengths are known, and θ_1 and θ_2 are determined by the hip and lumbar flexion angles, two

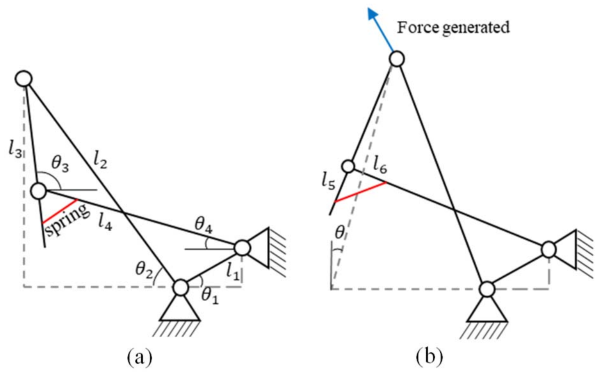

Fig. 3. The scheme of the crossed four-bar linkage of the exoskeleton's upper part, including the link and joint angle numbers. The configuration at (a) the upright posture and (b) the lumbar flexion angle of θ.

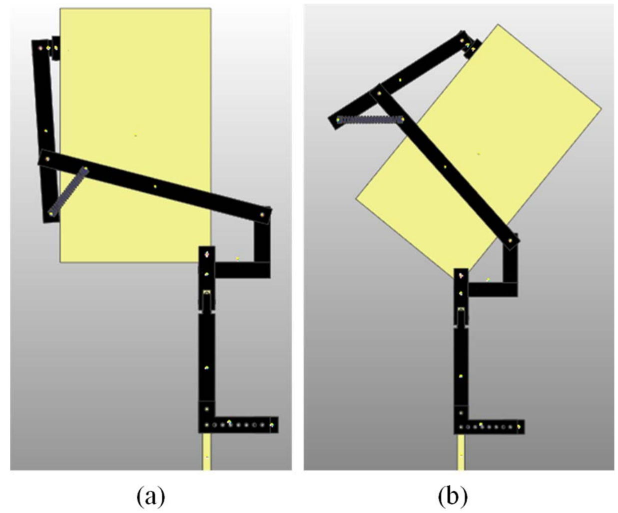

Fig. 4. For dynamic simulation, human body links were assumed as rigid links, and the leg motion was restrained. The exoskeleton and the human body were linked by fixed and revolute joints considering their works. The side view of the (a) upright and (b) 60-degree leaning postures.

unknowns (θ_3 and θ_4) can be solved. When all the angles are determined, the changes in the spring length can also be calculated.

$$\theta_3+\theta_4 = \cos^{-1}\left(\frac{l_3^2+l_4^2-l_1^2-l_2^2-2l_1l_2\cos(\theta_1+\theta_2)}{2l_3l_4}\right) \tag{3}$$

$$\theta_4 = \sin^{-1}\left(\frac{-l_2\cos(\theta_2)-l_1\cos(\theta_1)}{\sqrt{(l_3\sin(\theta_3+\theta_4))^2+(l_3\cos(\theta_3+\theta_4)-l_4)^2}}\right) - \sin^{-1}\left(\frac{(l_3\cos(\theta_3+\theta_4)-l_4)}{\sqrt{(l_3\sin(\theta_3+\theta_4))^2+(l_3\cos(\theta_3+\theta_4)-l_4)^2}}\right) \tag{4}$$

$$l_{spring} = \sqrt{l_5^2+l_6^2-2l_5l_6\cos(\pi-\theta_3-\theta_4)} \tag{5}$$

The kinematic model can be used to predict the exoskeleton configuration and the spring force generated for any lumbar and hip flexion angles.

C. Dynamic Simulation

The dynamic simulation was performed using *RecurDyn V9R3* to estimate the force exerted by the exoskeleton on the trunk (Fig. 4). The aim was to calculate the forces perpendicular and parallel to the wearer's trunk, according to changes in lumbar flexion angles. As it was not specifically intended to observe how the generated force biomechanically affects the human body, the trunk, pelvis, and legs were modeled as a rigid body for simplicity. For body segment length determination, the human model of the 50$^{\text{th}}$ percentile in 20–24-year-old Korean man (height, 1.74 m; weight, 70 kg [36]) and anthropometric data [37] were used. The property of the coil spring was set as the one selected. The simulated dynamic motion was pure trunk flexion from the upright to a 60-degree leaning posture for 10 seconds, without hip and knee flexion.

D. Biomechanical Modelling

In order to estimate the effect of providing parallel and perpendicular force elements to the trunk, we compared the compression forces for three cases: without supporting force, with perpendicular force (the previous exoskeletons), and with parallel and perpendicular force (the suggested exoskeleton). The suggested biomechanical equations [37]–[39] – based on force and moment equilibrium – were used, but supporting forces (i.e., extension moment generating force and traction force) were additionally included. Thus, the spinal muscle force, the abdominal pressure, the weights of the upper body and a load, and the supporting forces were considered. The same lifting motion and anthropometric data from the dynamic simulation were utilized as modelling conditions. The holding load and its distance from L5/S1 were assumed 15 kg and 40 cm, respectively. The results of the dynamic simulation were used as the supporting forces.

III. Experiment

A. Subjects

Ten healthy male subjects volunteered for the in-lab experiment, including static and dynamic lifting under non-wearing and wearing exoskeleton conditions. Their average age, height, and weights were 22.8 ± 1.5 years, 175.5 ± 6.2 cm, and 70.0 ± 6.0 kg, respectively. None of the subjects reported previous back pain or injuries. The study was approved by the Institutional Review Board (IRB) of Korea University (IRB-2021-0139). At the beginning of the experiment, the procedures were delineated to the subjects, and all subjects signed the consent forms.

B. Instrumentation

Surface electromyography (EMG) signals were recorded using a Delsys Trigno™ Wireless System (Delsys Inc., Boston, MA, USA). It differentially amplifies the signals with

Fig. 5. A posture guide was used, which restrains the movement of (a) knees and (b) feet. It was aimed to reduce the effect of leg-related motion during the task and focus on the action of spinal muscles.

a gain of 1000 and filters them with a bandwidth of 20–450 Hz. The sampling frequency of the EMG data was set to 2000 Hz. For EMG measurements, a total of six sites (both sides of the rectus abdominis [RA], erector spinae iliocostalis [ESI] at the L2 level [about 6 cm collateral], and erector spinae longissimus [ESL] at the L1 level [about 3 cm collateral] [40]) which majorly contribute to trunk flexion and extension were observed. RA is related to trunk flexion, while ESI and ESL are related to trunk extension. The spinal bones and iliac crest were palpated. An S-beam loadcell, DBCM-50 (Bongshin Loadcell Co., Ltd, Seoul, Korea) was used to measure the generated traction force. The load cell data were obtained with a sampling frequency of 2000 Hz, along with the EMG data. The lumbar angle was measured using AHRS EBIMU24GV3 (E2BOX, Korea), a wireless inertial measurement unit (IMU). The absolute Euler angles were obtained from the IMU sensor with a sampling time of 0.1 seconds. The IMU sensor was attached at the L3 level, and its data were utilized to observe the lumbar flexion angles during the task and to distinguish the starting point of the dynamic lifting phase. The lumbar flexion angles were obtained from the angle differences between the standing and bending postures. The electrodes and the IMU sensor were secured with 3M Micropore to prevent them from being displaced or removed during the experiment.

C. Experimental Procedures

Before conducting the task, maximum voluntary contraction (MVC) was measured to normalize the subjects' EMG data and compare them among the subjects. For MVC measurements, manual resistance methods [41] were used. The measurement trial was repeated three times, and each trial lasted five seconds [42]. Between each MVC measurement trial, the subjects rested for two minutes, and after the entire session, the subjects rested for at least 10 minutes for muscle recovery.

All the subjects performed the tasks with and without an exoskeleton, with each situation repeated thrice. The task included two phases: (1) static holding of a barbell plate for 5 seconds while leaning back to 45°, and (2) dynamic lifting for 3 seconds from leaning to upright posture. 10 kg and 15 kg were selected for the lifting tasks, considering usual lifting weights from everyday life activities such as house chores. An inclinometer was used to guide the subjects' starting posture, and the phase of the task was controlled using a metronome. The positions of the lower body segments were guided by a posture guide (Fig. 5), which functions similarly to previous studies [35], [43], and the arms were placed in the gravity direction. Minimum thoracic flexion was used to prevent different muscle use patterns [44]. The order of non-wearing and wearing the exoskeleton conditions for each task was randomized to eliminate the muscle fatigue effect. The subjects performed the task with 10 and 15 kg for the non-wearing condition and 15 kg for the wearing condition. The task with 10 kg was aimed at providing a guideline to examine the offloading effect of wearing the exoskeleton. For all sessions, the subjects' postures were constantly supervised and verbally corrected by one of the experimenters.

D. Data Processing

The bandpass filtered (bandwidth of 20–450 Hz) analog data from the Delsys Trigno^{TM} Wireless System were converted to digital data through an NI USB-6221 (National Instruments, Austin, TX, USA). Then, the data were received and saved using a custom software developed in *LabVIEW 2018* (National Instruments, Austin, TX, USA). For data processing, a custom program written in *MATLAB R2020a* (MathWorks Inc., Natick, MA, USA) was utilized. The acquired EMG data were demeaned, full-wave rectified, and low-pass filtered with a cut-off frequency of 2.7 Hz (6th order, Butterworth filter) [45]. For analysis, the EMG data from the static holding and dynamic lifting phases were normalized to each muscle's peak value of the processed MVC data. The loadcell data and angle data were also acquired through *LabVIEW 2018* and processed using *MATLAB R2020a*.

E. Data Analysis

In this study, the independent variable was exoskeleton intervention (2 levels), and the dependent variables were the mean normalized EMG values for the static holding phase and the mean and peak normalized EMG values for the dynamic lifting phase. Because all the tasks were symmetric, the EMG data from the left and right sides of the same muscle were averaged under the same experimental conditions (i.e., phase and wearing conditions). For the static holding phase, the mean EMG value from the same muscle and wearing conditions was calculated over the duration. Subsequently, a paired t-test ($\alpha = 0.05$) was performed to evaluate the effect of wearing the exoskeleton. For the dynamic lifting phase, the mean values were obtained by averaging the data over the phase for the same muscle and condition, and peak values were selected from the highest EMG values during this phase. The wearing condition was compared using the non-wearing conditions through the paired t-test ($\alpha = 0.05$). Before conducting the paired t-test, normality was tested using the Kolmogorov-Smirnov test.

TABLE I
EMG MEASUREMENT EXPERIMENT RESULT

Task phase		Load weight	Wearing condition	RA	ESI	ESL
Static	Mean %EMG (SD)	10 kg	Non-wearing	1.25 (0.75)	9.58 (4.97)	11.40 (4.81)
		15 kg	Non-wearing	1.16 (0.53)	11.67 (5.61)	13.10 (5.08)
		15 kg	Wearing	1.10 (0.55)	9.63 (4.44)	11.46 (4.27)
Dynamic	Peak %EMG (SD)	10 kg	Non-wearing	1.90 (1.17)	15.39 (6.84)	16.27 (7.30)
		15 kg	Non-wearing	1.85 (0.87)	17.63 (10.22)	17.99 (7.67)
		15 kg	Wearing	1.91 (1.15)	14.41 (6.22)	16.19 (5.69)
	Mean %EMG (SD)	10 kg	Non-wearing	1.35 (0.80)	11.41 (5.74)	12.80 (5.45)
		15 kg	Non-wearing	1.24 (0.55)	13.31 (7.45)	14.13 (5.94)
		15 kg	Wearing	1.17 (0.58)	10.51 (4.86)	12.29 (4.55)

The mean and peak %EMG of three observed muscles from the EMG measurement experiment are provided for each condition. EMG, electromyograph; RA, rectus abdominis; ESI, erector spinae iliocostalis; ESL, erector spinae longissimus.

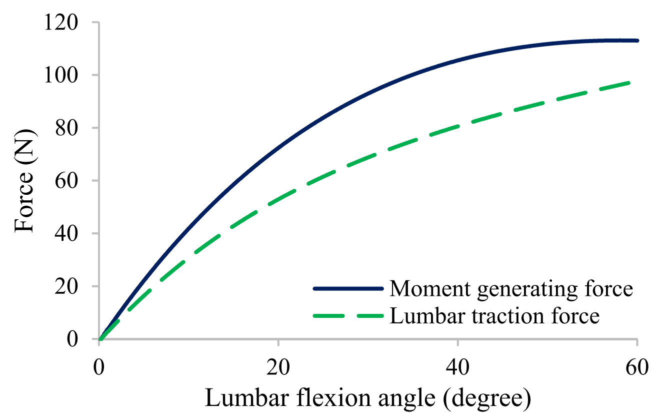

Fig. 6. Through dynamic simulation, the force elements, perpendicular and parallel to the trunk, were calculated, corresponding to the moment-generating and lumbar traction forces, respectively.

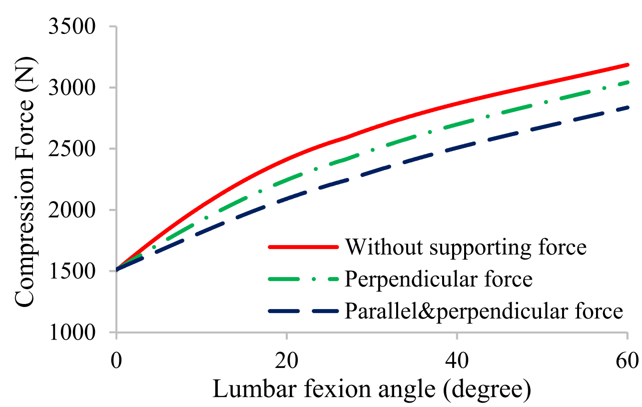

Fig. 7. The calculated lower back compression forces for three cases (i.e., without supporting force, with perpendicular force, and with parallel and perpendicular force).

IV. RESULTS

A. Dynamic Simulation

The force generated by the exoskeleton depended on the lumbar flexion angle (Fig. 6). In the range of 0°–60° of lumbar flexion, both the traction and moment-generating forces increased as the lumbar angle increased. With no lumbar flexion, the lumbar traction and moment-generating forces were almost zero, which was intended not to hinder the wearer's upright posture. At 45° of lumbar flexion, the same angle as the static holding phase of the EMG measurement experiment, the lumbar traction and the moment generating forces were 85.4 N and 109.3 N, respectively. At 60° of lumbar flexion, where the design criteria are set, the forces were 97.7 N and 113.0 N, respectively.

B. Biomechanical Modelling

The compression force at L5/S1 increased with the increased lumbar flexion angle for all three cases (Fig. 7). However, less increment was observed as more supporting force elements were provided. At 45° of lumbar flexion, the compression forces without supporting force, with perpendicular force, and with parallel and perpendicular force were 2951.9 N, 2788.8 N, and 2594.1 N, respectively. At 60° of lumbar flexion, the forces were 3185.6 N, 3041.9 N, and 2836.2 N, respectively. The nominal compression force (i.e., without supporting force) when lifting 10 kg and 5 kg at the posture were 3048.4 N and 2892.8 N, respectively.

C. Experiment Results

When wearing the exoskeleton, the overall mean and peak %EMG of the spinal muscles were significantly reduced, and no significant differences were reported for the abdominal muscles (Table I). In the static holding phase, wearing the exoskeleton significantly reduced the mean %EMG of ESI and ESL by 17.5% ($p < 0.001$) and 12.5% ($p = 0.002$) when compared to the without-15-kg condition, with no significant

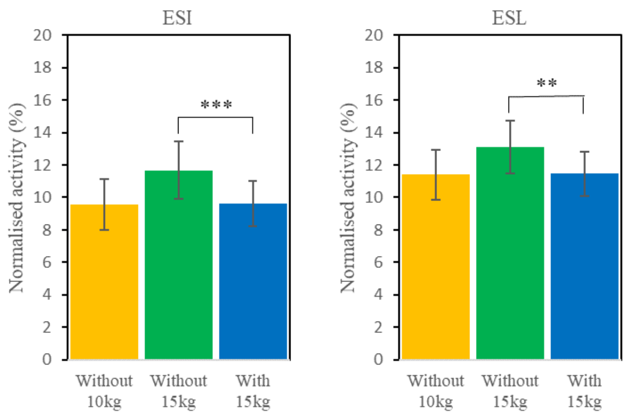

Fig. 8. The mean %EMG of ESI and ESL during the static holding phase was compared by each task case (i.e., wearing condition and load weight). The results demonstrate the offloading effect of wearing the exoskeleton.

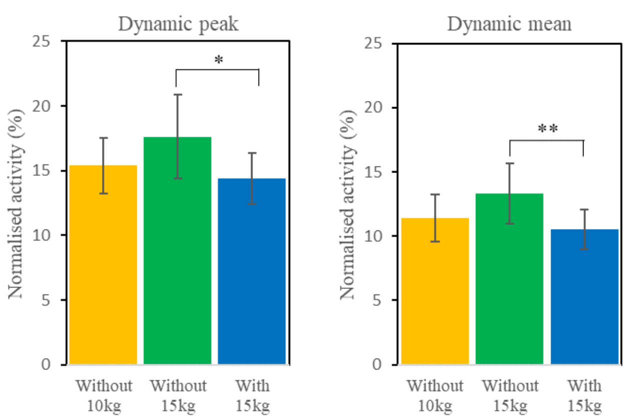

Fig. 9. The peak and mean %EMG of ESI during the dynamic lifting phase were compared by each task case. The results demonstrate the exoskeleton's offloading effect.

difference from the without-10-kg condition ($p = 0.434$ and $p = 0.438$, respectively) (Fig. 8). The reduced mean %EMG was observed in ten and nine subjects for ESI and ESL, respectively. During the dynamic lifting phase, the mean and peak %EMG of ESI were significantly reduced by 21.0% ($p = 0.005$) and by 18.3% ($p = 0.034$) during the with-15-kg condition, compared to the without-15-kg condition. The mean and peak %EMG of ESI decreased in ten and nine out of ten subjects, respectively. The mean %EMG of ESI under the with-15-kg condition was significantly reduced by 7.9% ($p = 0.020$) than under the without-10-kg condition, but with no significant difference for peak %EMG ($p = 0.086$) (Fig. 9). For the same phase, the mean %EMG of the ESL was significantly reduced by 13.0% (0.017), while no significant difference was observed in peak %EMG with $p = 0.074$ for the comparison of the with-15-kg condition and the without-15-kg condition. Nine and seven out of ten subjects, respectively, reported a decrease in mean and peak %EMG of ESL. When comparing the with-15-kg condition to the without-10-kg condition, no significant differences in the mean and peak

TABLE II
TASK PERFORMANCE METRIC

	Without 10 kg	Without 15 kg	With 15 kg
Lumbar angle (degree)	44.1 ± 4.3	45.3 ± 1.5	45.2 ± 3.5
Traction force (N)	-	-	50.7 ± 8.8

The maximum lumbar flexion angles and traction force are provided, which correspond to the angle and force from the static holding phase.

%EMG of ESL were observed ($p = 0.163$ and $p = 0.450$, respectively). No significant changes were reported from the mean %EMG of the RA for both the static and dynamic lifting phases ($p = 0.194$ and $p = 0.184$, respectively). The maximum lumbar flexion angles for each task condition were not significantly different ($p = 0.658$), and the traction force during the static holding phase resulted in 50.7 ± 8.8 N (Table II).

V. DISCUSSION

This study introduces a novel passive lower-back support exoskeleton that provides two types of force elements: lumbar traction force and extension moment generating force. The dynamic simulation validated that the exoskeleton generated the desired range of force for both the perpendicular and parallel directions to the upper body. It also demonstrated an increased supporting force according to the increased lumbar flexion angle. It was intended to compensate for the increased lumbar muscle activation and compression force at L5/S1 with an increased lumbar flexion angle.

The biomechanical modelling demonstrated that lumbar traction force would be an extra option to reduce the compression force, in addition to extension moment. Although the simplified biomechanical model was used, it includes essential force elements and provides sufficient insight into the exoskeleton's supporting effect. While extension moment in the desirable range lowers human muscle burdens and the compression force during lifting, the excessive moment may hinder human movement [25]. By contrast, the direction of the lumbar traction force is less related to lumbar flexion; in this regard, this force might cause minor motion hindrance compared to the extension moment, but further investigation is required to prove it experimentally. The lower back compression force has been considered a crucial criterion of LBP and related injuries in industries [14]. This study suggests additional consideration to support the lower back: lumbar traction force. With its effect on the reduced compression force, it may initiate a new back supporting paradigm.

EMG measurements showed reduced erector spinae longissimus and erector spinae iliocostalis activation when wearing the exoskeleton. This resulted from the exoskeleton's moment-generating force, which was perpendicular to the trunk. The force reduces the moment that the erector spinae must counter during static holding and dynamic lifting. The reduced spinal muscle force reduces the lower back compression force as the muscle force is a major determinant of the compression force [37]. The exoskeleton was designed to offload approximately 5 kg. However, the moment-generating force can be adjusted by changing each link length or replacing the coil spring with a different stiffness, depending on the situation. The extent of reduced muscle activation differed for each subject, and this might be related to their different weights and heights. Specifically, the placement of the exoskeleton joints with respect to the human body was varied for different body sizes. Improving the adjustability of the exoskeleton for different sizes of people will be beneficial for maximizing the effects of the exoskeleton.

The suggested device did not significantly increase abdominal muscle activity during the static holding and dynamic lifting phases. This implies that the spring stiffness was in the desirable range, not leading to additional flexion moment generation. If an exoskeleton creates an excessively high extension moment, additional flexion moments are required for moment equilibrium; therefore, unintended abdominal muscle activation can occur [24], [27]. Several previous studies on passive back exoskeletons have reported increased abdominal muscle activities, especially during stooped lifting, and mentioned its probable advantages in terms of body stability [20], [46]. Indeed, muscle co-contraction may augment trunk stiffness and contribute to enhanced spine stability [47]–[50]; however, this muscle coactivation requires additional physiological energy and increases the risk of muscle fatigue [47]–[49]. In addition, although many simplified trunk models do not consider the effect of trunk muscle co-contraction on the lumbar compression force, they do contribute to it [51], [52]. Whether the additional abdominal muscle contraction is beneficial to the lower back is still disputable, and further studies are required. However, considering the probable negative effects of co-contraction, it might be better to design an exoskeleton that does not require an additional flexion moment, as in this study.

The most distinct feature of the suggested exoskeleton is providing lumbar traction force, which direction is opposite to the lower back compression force. Previous back exoskeletons using elastic bands (i.e., PLAD [17]–[20] and the biomechanically assisted garment [21]) generate elastic force in the direction of the compression force. Those with spring-like elements at the hip joints, such as Laevo [22]–[24] and BNDR [25], [26], may generate the traction force due to the unintended misalignment of joints. However, the misalignment is a random consequence of uncontrollable factors, including the wearer's varied body size. According to our investigation, in order to generate the traction force with a significant amount and an increasing trend with the lumbar flexion angle increases, the length and the direction of the joint offset should be carefully calculated and determined. Thus, it would be uncertain to expect that those exoskeletons have a considerable supporting effect with the traction force. On the other hand, the suggested exoskeleton's joint with a force source (i.e., spring) is intentionally designed to be apart from the hip joint, which works as the key to generating the traction force. Its offset from the hip joint traces the designated path with the changing configuration of the four-bar linkage to create a desirable force profile – lumbar traction force increases with increasing lumbar flexion angle.

Lumbar traction is a prevailing clinical treatment for lower back pain relief with its known mechanical effect on spinal elements [28]–[31]. The amount of applied traction force is clinically determined depending on the duration of force applied, with longer durations indicating smaller forces [29], [53]. As the suggested exoskeleton is targeted for use in industries or everyday life, its expected hours of use can be several hours, implying that it should not be strong enough to cause significant structural changes in the vertebrae [53]. Thus, the exoskeleton was designed to provide a traction force between 80 and 100 N at 60° of back flexion. If necessary, the traction force can be adjusted by using a coil spring of different stiffness or by alternating it with other types of springs.

Although the trend of traction force was identical to the dynamic simulation and the experiment, the force increased with the greater lumbar flexion angle, differences in amount were found for two major reasons. First, the length of the link corresponding to the human upper body was not constant. This was because the human trunk length changed during trunk flexion and extension. In addition, the chest harness slid upward when the exoskeleton generated a traction force. As extended link length lessens spring extension, a weaker traction force than expected could have been generated. Second, the direction of the force measured by the load cell did not concur with the lumbar traction force. The load cell measured the force parallel to the back part of the chest harness because of the lack of methods to measure the force parallel to the human back. However, the chest harness was lifted from the back when the exoskeleton generated a moment-generating force, which was perpendicular to the back. This might have resulted in angle differences between the harness and the human back. Nevertheless, it clearly demonstrated the existence and general trend of the generated lumbar traction force.

This study had several limitations. First, the current prototype includes a rigid linkage, and it allows trunk flexion/extension but limits lateral bending and twisting. Since the primary purpose of the research was to introduce a new supporting mechanism – generating force elements in two directions, the range of motion was relatively considered less. Future research will be followed to enhance its eligibility to the real-world application by adding additional joints and utilizing flexible materials (i.e., carbon fibers). Moreover, the size, weight, and protrusion will be minimized with design optimization. Second, the reduced intradiscal pressure could not be measured directly. The intradiscal pressure can be estimated by an invasive method using a needle-like device with a pressure transducer at the tip [37]. However, this method requires professional medical knowledge and restricts large movements [37]. Thus, we instead measured other available forces, which can be indicators of reduced intradiscal pressure: reduced lumbar muscle activities and the existence of traction force. Future studies which directly measure intervertebral pressure will more clearly reveal the effects of the exoskeleton. Third, the subjects were limited to healthy men without previous back-related injuries. This study aimed to demonstrate the effects of an exoskeleton on the most general group in the manufacturing industry. However, its effect can be varied by subject groups owing to their different back mechanisms. For example, patients with chronic back pain reported a significantly lower flexion-relaxation ratio than healthy subjects [54]. Thus, including various subject groups, such as back pain subjects or female subjects, will be advantageous to validate the effects of the exoskeleton on the general population. Lastly, this study was limited in addressing the short-term effects on selected muscle activation. It is anticipated that muscle fatigue will also be reduced if the exoskeleton is worn during stoop leaning or heavy lifting tasks. In contrast, there might be unpredictable long-term effects due to the unnatural external extension moment and traction force. Therefore, examining the long-term effects is necessary before real-life implementation.

VI. Conclusion

In this study, we developed a passive lower-back support exoskeleton that generates both parallel and perpendicular force elements to the trunk. The exoskeleton is novel as it provides traction force, which is a prevailing clinical method for patients with lower back pain, in addition to the extension moment at the L5/S1 region. Thus, it reduces intradiscal pressure by reducing the activation of the lumbar muscles and pulling the upper body in the opposite direction of the pressure. Its force generation was estimated by dynamic simulation, and it was validated through an experiment with ten subjects. The results imply that lower back pain and related injuries from static back bending or heavy lifting tasks would be reduced or prevented by wearing the suggested exoskeleton. However, the current model requires further development for its real-world application. We are planning on future research to improve the model by enhancing the wearer's range of motion and minimizing its size and weight.

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