A Novel Passive Back-Support Exoskeleton With a Spring-Cable-Differential for Lifting Assistance

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***Abstract*— Lower back injuries are the most common work-related musculoskeletal disorders. As a wearable device, a back-support exoskeleton (BSE) can reduce the risk of lower back injuries and passive BSEs can achieve a low device weight. However, with current passive BSEs, there is a problem that the user must push against the device when lifting the leg to walk, which is perceived as particularly uncomfortable due to the resistance. To solve this problem, we propose a novel passive BSE that can automatically distinguish between lifting and walking. A unique spring-cable-differential acts as a torque generator to drive both hip joints, providing adequate assistive torque during lifting and low resistance during walking. The optimization of parameters can accommodate the asymmetry of human gait. In addition, the assistive torque on both sides of the user is always the same to ensure the balance of forces. By using a cable to transmit the spring force, we placed the torque generator on the person's back to reduce the weight on the legs. To test the effectiveness of the device, we performed a series of simulated lifting tasks and walking trials. When lifting a load of 10 kg in a squatting and stooping position, the device was able to reduce the activation of the erector spinae muscles by up to 41%. No significant change in the activation of the leg and back muscles was detected during walking.**

***Index Terms*— Lower back injuries, back-support exoskeleton, spring-cable-differential, lifting work.**

I. Introduction

LOWER back injuries are the most common work-related musculoskeletal disorder (MSD), and it takes at least one week to recover from lower back injuries [1]. In addition to harming workers' health, lower back injuries also place a tremendous financial burden on workers, their employers, communities, and the economy [2]. In many workplaces, such as warehouses, construction, airports, and hospitals, workers are susceptible to lower back injuries because they must maintain awkward postures for extended periods while lifting heavy objects [3]. The gravity of the objects puts a lot of stress on the workers' lumbar spine (L5/S1 region of the spine), causing fatigue and even damage to the back muscles [4]. To protect workers from lower back injuries during lifting, many studies have been conducted in recent decades on devices that can reduce the load on the lumbar spine [5].

A back-support exoskeleton (BSE) is a wearable device that can reduce lumbar strain during lifting activities. It has been reported that a BSE (an on-body device) is more accepted by workers than off-body devices (such as forklifts and hoists) and has great potential for use in real-world industrial and everyday scenarios [6]. In general, a BSE helps workers lift objects by creating an assistive torque between the trunk and thighs (an extension torque at the hip joints) [7]. According to the different types of torque generation, BSEs can be divided into active, semi-active and passive types. The active BSEs use motors [8], [9], actuators with twisted strings [10], [11] and pneumatic artificial muscles [12], [13] to generate the assistive torque. With the help of the sensing and control system, active BSEs can provide assistance according to the user's needs [14]. However, due to the weight of the actuator, battery, and complex transmission mechanisms, active BSEs are usually heavy, poorly portable, and require regular charging, which severely limits their applications. The semi-active BSEs use low-power servo motors to adapt the behavior of the device, which is a trade-off between the active and passive BSEs. Although lighter than active BSEs, semi-active BSEs still have complex structures including motors, controllers, and batteries, making them less portable than passive BSEs [15], [16].

Passive BSEs use elastic elements to generate the assistive torque and can be lightweight due to the simple structure [5]. To ensure user's comfort and safety, passive BSEs should provide assistance during lifting operations but should not prevent the user from walking. If the user has to overcome the resistance of the device to walk, it will cause great discomfort and even stumbling because the legs are not lifted sufficiently. Therefore, a passive BSE must be able to automatically distinguish between lifting and walking if the mechanism is appropriately designed. However, the current passive BSEs have the problem of meeting the above requirement. For

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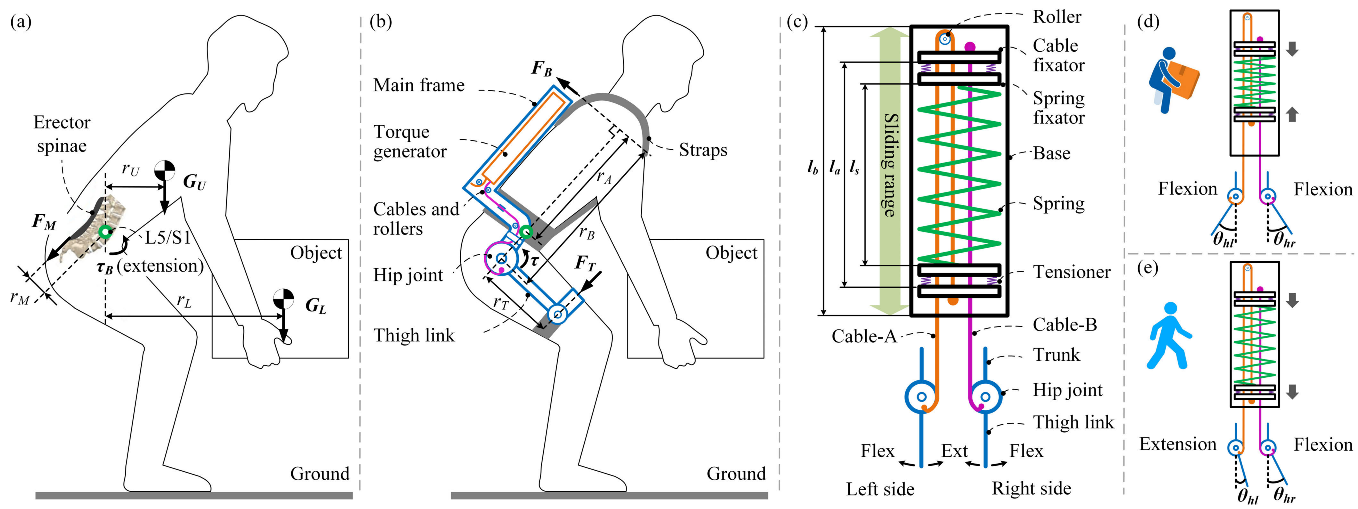

Fig. 1. (a) Force analysis during lifting (the simplified model is built in the sagittal plane under static or quasi-static conditions). (b) Working principle of the proposed BSE. (c) Working principle of the spring-cable-differential (torque generator, the weight of the spring is ignored). (d) The spring is compressed when both hip joints are flexed for lifting (the legs move forward relative to the trunk and the hip joint angle θ_{hl} and θ_{hr} are both positive). (e) The spring slides to one side when one hip joint is extended and the other hip joint is flexed during walking (legs are staggered). The reserved space between the cable fixator and the spring fixator can compensate for the difference in the magnitude of hip extension and flexion during walking.

example, the passive BSEs with elastic bands [17], [18], elastic beams [19], [20] and springs [21], [22] deliver the extension torque to the hip joints as soon as the relative angle between the trunk and thighs decreases (the angle is 180° when standing upright). Therefore, the user must move against the torque of the device when he/she lifts the legs to walk. The passive BSE called Laevo [23], [24] adds one degree of freedom (DOF) of external rotation to the thigh links. If the user pretends to walk, they can manually remove the thigh links from the legs to move freely. However, this process makes it difficult to use the device. In addition, it is difficult for the user to rotate the thigh links when they have lifted the object and want to transport it. The BSE called BackX [25], [26] uses a structure of two gas springs that can rotate with the human torso. The gas springs are compressed only when the torso is bent beyond a certain angle with respect to the vertical, allowing the user can to move the legs freely when the torso is not bent. However, since the assistive torque depends only on the flexion angle of the torso, the device cannot provide adequate assistance when the user lifts loads in a squatting or bending-squatting mixed posture, where the flexion angle of the torso is very small [19]. In addition, the above devices have two independent torque generators on the left and right sides that produce different assistive torque at different angles of the hip joints (with different spring deflections). The unbalanced force may cause user discomfort or device tilt.

To overcome the above challenges, in this paper we introduce a novel passive BSE that can automatically provide adequate assistance for lifting tasks and low resistance for walking. A unique spring-cable-differential was developed as a torque generator, in which two cables pull the ends of a spring in opposite directions and are connected to the left and right hip joints. The spring-cable-differential compresses the spring to generate force when both hip joints are flexed during lifting, and slides the spring to one side (no force generated) when the legs are staggered during walking. The design parameters of the spring-cable-differential are optimized to accommodate the asymmetry of human gait (the difference in the amount of hip flexion and extension). Since the force of two cables is equal to the spring force, the assistance is always balanced with the same assistive torque on both hip joints. In addition, the force generator is located on the human's back to reduce the weight on the legs. To evaluate the "expected" and "unexpected" effects of the device, we conducted a series of muscle activity tests with ten healthy subjects lifting and walking with and without the device.

The paper is organized as follows. In section II, the design of the BSE is presented. Section III describes various experiments performed to validate the proposed device. The results of these experiments are discussed in Section IV. Finally, in Section V, the paper is concluded.

II. Method

A. Working Principle of the BSE

The force analysis of the lower back muscle during lifting operations is shown in Fig. 1 (a). Since the dynamic model of human spine movements is very complex, a simplified model in the sagittal plane under static or quasi-static conditions is made here (friction is also not considered). When the worker lifts the load, the gravity of the upper body and the object (G_U and G_L) causes a large load on the lumbar spine (L5/S1 area). To overcome this load, the worker must generate a large extension moment (τ_B) through the erector spinae. The force of the erector spinae (F_M) must meet the requirement shown in (1). Since the moment arms of G_U and G_L (r_U and r_L) are much larger than that of the F_M (moment arm r_M), the erector spinae must apply a large compressive force and is susceptible to injury due to overstress and overfatigue. In addition, the large compressive force on the lumbar spine poses the risk of spinal injury [27], [28].

$$\tau_B = F_M \times r_M \geq G_U \times r_U + G_L \times r_L. \tag{1}$$

To reduce the load on the lumbar spine, the proposed BSE generates an assistive torque (τ) between the human trunk and thighs. As shown in Fig. 1 (b), the device mainly consists of five parts, namely a main frame, a torque generator, cables and rollers, hip joints (left and right), and thigh links. The main frame and thigh links are attached to the human's back and legs using straps, and are rotatably coupled to the hip joints. During lifting operations, the torque generator drives the hip joints via cables and generates assistive torque. The main frame provides a supporting force to the user's upper body (F_B is the force component perpendicular to the line connecting the center of rotation of the hip joint and L5/S1). A compressive force (F_T is the force component perpendicular to the thigh link) is also applied to the thighs via the thigh links.

With the proposed device, the load on the lumbar spine can be reduced by:

$$\begin{aligned} &\tau_B \geq G_U \times r_U + G_L \times r_L - F_B \times r_A \\ &F_B = \frac{\tau}{r_B} \quad \text{and} \quad F_T = \frac{\tau}{r_T} \end{aligned} \tag{2}$$

The BSE distributes the load on the lumbar spine to the user's shoulders, chest and thighs. By using soft straps with a large contact surface, we reduce the strain on the user's body to avoid discomfort.

The detailed design targets of the BSE can be summarized as follows:

1) Low device weight. According to [29], the weight of an exoskeleton should be around 3 kg to reduce the load on the user's body. Therefore, the target weight of the device is set at 3 kg.
2) Adequate assistive torque. As a passive device, the BSE should provide enough assistive torque to release the back muscles while avoiding excessive bending resistance (to store energy in spring) [18].
3) Sufficient range of motion. In addition to sufficient hip flexion range, the BSE should also provide a certain range of hip abduction/adduction to facilitate other leg movements such as sidekicks [22].
4) Low walking resistance. The BSE should not cause any other resistance during walking than the device weight and the constraints imposed by the straps.

B. The Spring-Cable-Differential

The torque generator is the key component of the BSE, which determines the trend and magnitude of the assistive torque. In this work, we designed a spring-cable-differential as the torque generator. The structure of the spring-cable-differential is shown in Fig. 1 (c) (the weight of the spring is ignored). A spring is connected to the spring fixators and can slide along the base of the torque generator. Cables are connected between the cable fixators and the hip joints. Inside the torque generator, the cables run parallel to the spring. Tensioners (small compression springs) are located between the cable fixators and the spring fixators to pre-load the cables and prevent them from sagging. The principle of differential operation is as follows. During lifting, both hip joints are flexed, the cable fixators are pulled to the middle of the base to compress the spring (large force generated). When legs are staggered in walking, the cable fixators are pulled in the same direction, and the spring moves to one side without compression (no force generated).

We define the initial position of the spring-cable-differential when the user stands upright and the spring is in the center of the base (the structure is symmetrical). The design parameters include the initial length of the spring l_{s0}, the initial distance between the cable fixators l_{a0}, the length of the base l_b, and the spring stiffness k_s. The independent variables that determine the assistive torque are the angle of the left and right hip joints (θ_{hl} and θ_{hr}, the flexion direction is defined as positive). When the user bends to a certain position, the actual distance between the cable fixators (l_a) is calculated as follows:

$$\begin{aligned} l_a &= l_{a0} - 2r\theta_{ha} \\ \theta_{ha} &= \frac{\theta_{hl} + \theta_{hr}}{2} \end{aligned} \tag{3}$$

where r is the radius of the hip joint and θ_{ha} is the average angle of the hip joints. For simplicity, we ignore the thickness of the spring and cable fixators here.

When l_a is greater than l_{s0}, the spring is not compressed. If l_a is smaller than l_{s0}, then the spring is compressed and the actual length of the spring (l_s) is equal to l_a. The deformation of the spring (Δl_s) can be calculated as follows:

$$\begin{aligned} &\Delta l_s = 0, \quad \text{if } l_a > l_{s0}; \\ &\Delta l_s = l_{s0} - l_a, \quad \text{if } l_a \leq l_{s0}. \end{aligned} \tag{4}$$

The assistive torque (τ, the sum of the assistive torque on both sides; friction is ignored) can then be obtained based on Hooke's law as follows:

$$\tau = 2rk_s\Delta l_s. \tag{5}$$

If the legs move the same amount in opposite directions when walking (when $\theta_{hl} + \theta_{hr} = 0$), l_a can always equal l_{a0}. The spring only moves from side to side along the base without compression. However, walking is not a symmetric motion, and the magnitude of θ_{hl} is not equal to that of θ_{hr} during most of the gait cycle [30], [31]. To solve this problem, our design allows l_{a0} to be larger than l_{s0}, which leaves space between the spring fixators and the cable fixators (on the same side). The extra space can compensate for the difference in the magnitude of hip extension and flexion during walking, allowing the user to walk without resistance. The maximum difference in magnitude that can be compensated (represented by $\Delta\theta$) is given by:

$$\begin{aligned} \Delta\theta &= |\theta_{hl} - \theta_{hr}| \\ &= \frac{l_{a0} - l_{s0}}{r}. \end{aligned} \tag{6}$$

A special case is when the user performs one-legged movements. For example, a user lifts the right leg to walk up stairs or climb over obstacles. In this case, θ_{hr} is zero and $\Delta\theta$ is equal to θ_{hl}, the maximum angle at which the leg can be lifted without resistance is $\Delta\theta$.

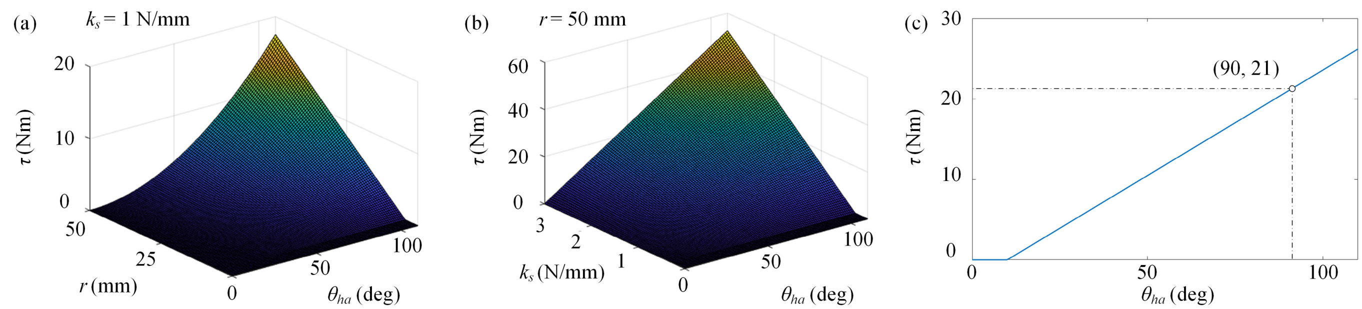

Fig. 2. (a) Relationship between the assistive torque τ (under static or quasi-static conditions) and the radius of the hip joint r during lifting operations (the average angle of the hip joints θ_{ha} is set from 0° to 100° and the spring stiffness is set to 1 N/mm for simplicity). (b) The relationship between the assistive torque τ and the spring stiffness k_s during lifting operations (r is selected as 50 mm). (c) The assistive torque generated by the device with the selected parameters ($r = 50$ mm and $k_s = 1.5$ N/mm).

In lifting applications, the device starts to assist the user when the average angle of the two hip joints (θ_{ha}) is greater than $\Delta\theta/2$. In real applications, we can choose the value of $\Delta\theta$ depending on the specific work scenarios. In workplaces where the worker walks infrequently or only on level ground, $\Delta\theta$ can be set as a small value, and when the user needs to walk up stairs or over obstacles, $\Delta\theta$ can be set larger.

Another requirement of the spring-cable-cable is a sufficient sliding range. The length of the base (l_b) should meet the requirement that:

$$l_b \geq l_{a0} + 2r\theta_{he} \tag{7}$$

where θ_{he} is the maximum extension angle of the hip joint.

Among the design parameters of the spring-cable-differential, $\Delta\theta$ and r are related to the structure design, while l_{s0} and k_s are related to the spring selection. According to (3)-(6), we show that the value of l_{s0} does not affect the output torque. Moreover, based on the gait data presented in [30] and [31], we set $\Delta\theta$ to 20°, which corresponds to the maximum amplitude difference between θ_{hl} and θ_{hr} during walking. Then, we can obtain the relationship between τ and r during lifting by assuming for simplicity that k_s is 1 N/mm (under static or quasi-static conditions, see Fig. 2 (a)). The range of θ_{ha} is set to from 0° to 110° because the possible value of the total angle of the two hip joints during lifting operations is up to 220°, as shown in [32]. As shown in Fig.2 (a), τ increases with r. The value of r is limited to 50 mm to prevent the hip joint from hindering the user's normal movements and to reduce the weight of the device. Therefore, we set r to 50 mm to obtain a large output torque.

The relationship between τ and k_s in lifting operations (when $r = 50$ mm) is shown in Fig. 2 (b). We choose the spring stiffness based on the target assistive torque when θ_{ha} is 90°, which refers to the most common lifting postures [9], [33]. In this work, the target assistive torque is set to 20 Nm (total assistive torque of the device, the same value as in [9]) for the following reasons. First, the BSE helps to relieve the load on the lower back muscles but does not remove the entire load, which can be up to more than 100 Nm [19]. Second, a larger assistive torque (with a harder spring) requires the user to apply more force to bend, which can affect the worker's efficiency. By using a spring with a stiffness of 1.5 N/mm, the device can provide a maximum assistive torque of 21 Nm to entire low-back for common lifting postures, as shown Fig. 2 (c). The selected spring also has a high deflection (195 mm) to provide sufficient stroke for lifting operations (θ_{ha} can be up to 110°). In real applications, the user can achieve a larger or smaller assistive torque by selecting a harder or softer spring. The final design parameters are shown in Table I.

TABLE I
THE DESIGN PARAMETERS OF THE SPRING-CABLE-DIFFERENTIAL

l_{s0}	l_b	l_{a0}	k_s	r
300 mm	450 mm	320 mm	1.5 N/mm	50 mm

[1] The actual l_{a0} is 330 mm when considering the spring fixator thickness (5 mm).

C. The Prototype Design

The overall design of the BSE is shown in Fig. 3 (a). The user can carry the device with the back strap and the thigh strap. Two sliding links (usually secured by screws) are located on the top of the main frame to adjust the height of the device. The contact surface between the human and the device (back, chest, and thigh) is padded to reduce stress on the human body. The exoskeleton's components are made of carbon fiber and aluminums and are designed to be lightweight yet strong. The prototype weighs 3.5 kg, making it easy for users to carry. In order to be suitable for different body shapes, the prototype is manufactured in different sizes, e.g. S, M and L (the 3.5 kg weight is for size M).

The spring-cable-differential is installed on the main frame (which is covered by a back shell), as shown in Fig. 3 (b). We installed the spring-cable-differential in the vertical direction along the human back to make the best use of space. In addition, a long back frame results in a large moment arm of the assistive torque (from the shoulder to the center of the hip joint), reducing the stress on the shoulders. The mechanism is supported by two support rods attached to the main frame (base). The cable fixator and the spring fixator can slide along the support rods with the help of the linear bearing. The spring movement is guided by the cylindrical structure of the spring fixator and a spring guide fixed in the center of the main frame. Small compression springs are mounted on the

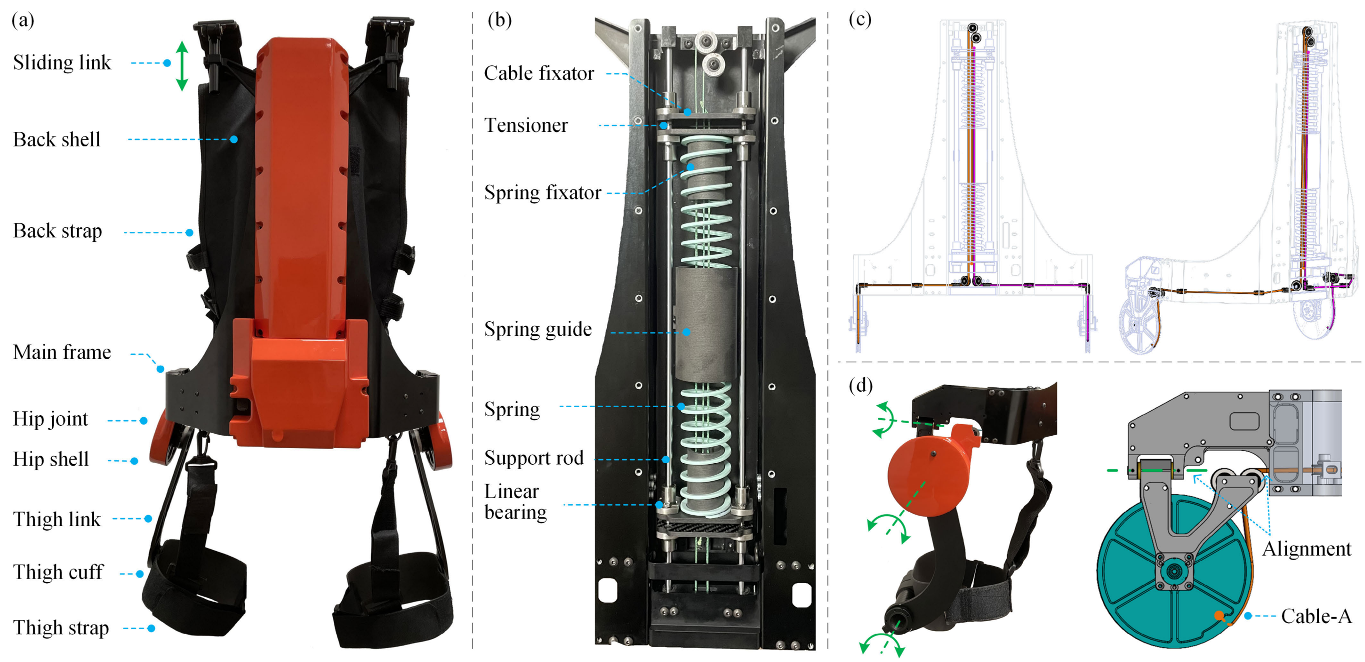

Fig. 3. (a) The overall structure of the SE prototype. (b) The structural details of the spring-cable-differential. (c) The cable route inside the main frame. (d) The DOFs and the structural details of the hip joint (with the hip shell hidden). The centerline of abduction/adduction of the hip is tangential to the edge of the roller through which the cable passes.

support rods (between the cable fixator and the spring fixator) as the tensioner and provide a small amount of preload to the cables. Cable-A and cable-B run parallel through the center of the spring, and the distance between cable-A and cable-B is set to be small to reduce the bending moment on the support rod. By using the rollers attached to the main frame to guide the cables, the force of the spring can be transmitted to the hip joints. The structural details of cables and rollers are shown in Fig. 3 (c).

The degrees of freedom (DOFs) of the hip joint are shown in Fig. 3 (d). There are a total of three DOFs on each side of the BSE, including hip flexion/extension, hip abduction/adduction, and a rotational DOF between the thigh link and the cuff. We placed the centerline of hip abduction/adduction tangential to the edge of the roller where the cable passes, so that hip abduction/adduction does not affect the force transmission of the cable. The DOF between the thigh link and the cuff allows the thigh cuff to rotate with the thigh for comfort. Fig. 4 shows a user wearing the BSE with some typical movements including standing, bending and sidekicks.

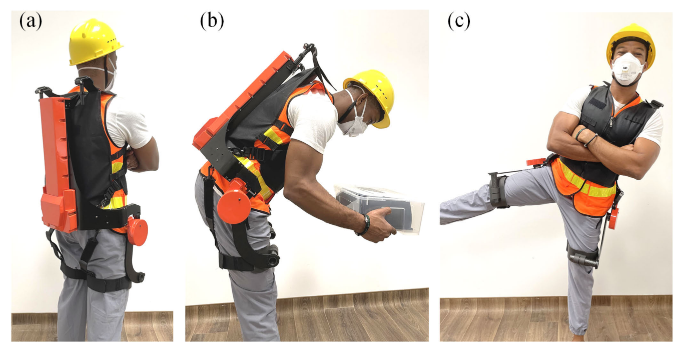

Fig. 4. A user wears the BSE. (a) Standing straight. (b) Bending over with a load. (c) Kicking the leg to one side.

III. Experiments

A. Experimental Design

The aim of these experiments was to investigate the effects of the proposed BSE on muscle activation during simulated industrial lifting tasks. The simulated tasks included (A) squat lifting, (B) stoop lifting from the floor, and (C) walking. The best lifting technique to move something heavy is to squat and use the power of the legs instead of the back to lift the object from the floor. The stoop lifting technique was considered because workers in various industrial environments do not always have the opportunity to bend their knees (e.g., when lifting from or into a mesh box). Finally, walking was evaluated to assess the effort required by the user to wear the exoskeleton when performing tasks other than lifting loads, in this case walking.

A load of 10 kg was chosen for the lifting tasks, which was placed in a plastic box (W × D × H of 40 × 29 × 32 cm) with handles on both sides. The measurement tests were repeated twice, and each test lasted approximately 20 seconds. All subjects performed the tasks alternately with and without the exoskeleton. In A, participants had to lift a box from the floor to the upright position four times with knees bent. In B, participants had to lift a box from the floor to the upright position four times with the knees extended. The position of the subject's feet in relation to the box on the floor for the lifting tasks was determined individually, so that the same foot position could be maintained when switching between conditions A and B. In C, participants walked at walking pace along a 15-meter track. During the walking trials, data were divided into cycles using an accelerometer to indicate heel strike for each epoch of the gait cycle.

B. Participants

Nine male participants (age: 24.6 ± 3.2 yo; height: 176.3 ± 5.5 cm; weight: 72.2 ± 8.5) were recruited to participate in the experiments. None of the participants reported having a previous low back injury at the time of data collection. The participants signed an informed consent prior to study participation. The study was designed in accordance with the Declaration of Helsinki and approved by the local Ethics Committee of the University with protocol number NUS-IRB LH-20-021.

C. Data Analysis

The electrical activity of four muscles was recorded by surface electromyography (EMG) by placing surface electrodes (27 × 37 × 13 mm, Delsys Inc., Boston, MA, USA) on the muscle bellies. The collected EMG signals were differential amplified with a gain of 1000, filtered with a bandwidth of 20-450 Hz, sampled at 2000 Hz, analyzed and stored. The following muscles were measured: erector spinae (ESI), erector spinae longus (ESL), rectus femoris (RF) and biceps femoris (BF). The muscles being monitored were on one side (the right side) of the subjects, since the subjects performed symmetrical tasks and the movements produced by the muscles on both sides were similar. The experiments took place at the Biorobotics Lab at the National University of Singapore. Before performing the experimental tasks, the participants were equipped with the EMG measurement sensors after the skin was cleaned and performed maximal effort to achieve MVC of the back and leg muscles. The peak values of the linear envelope of the EMG signals were defined as the MVC for each muscle and used to normalize the subjects' EMG data and for comparison between subjects. SENIAM methods were used for the MVC measurements [34]. We performed a normality test on the data with and without the BSE through Anderson Darling test. The simple paired t test was used for normally distributed data, and the Wilcoxon signed rank test was used for nonnormally distributed data, with $p < 0.05$ considered statically significant (using MATLAB and GraphPad Prism software version 9.0, San Diego, CA, USA).

D. Experimental Results

The EMG results (in percentage of MVC) of the simulated tasks are shown in Fig. 5. During the squat lifting, significantly lower mean activations of the ESI (−4.21% MVC, $p = 0.014$) and ESL (−10.63% MVC, $p = 0.012$) muscle were found in the condition with the BSE compared to the condition without the BSE. In addition, lower maximal activations of the ESI (−18.69% MVC, $p = 0.03$) and ESL muscles (−31.16% MVC, $p = 0.027$) were found when using the exoskeleton. Significantly lower mean activations of ESI (−4.55% MVC, $p = 0.017$) and ESL muscles (−7.17% MVC, $p = 0.008$) were found in the situation with the BSE when lifting in a stooped position. Similarly, lower maximal activations of ESI (−16.78% MVC, $p = 0.006$) and ESL (−24.47% MVC, $p = 0.021$) were found when using the exoskeleton. In addition, maximum activation of the RF muscle (−8.41% MVC, $p = 0.009$) also decreased with the BSE. No significant differences in BF activation were observed in any of the conditions.

TABLE II
PERCENTAGE REDUCTION IN ACTIVATION OF BACK AND LEG MUSCLES WHEN USING THE BSE WHILE LIFTING AND WALKING

Muscle Group	ESI	ESL	RF	BF
Squat lifting				
Mean	12.82*%	34.72%*	-3.99%	4.91%
Max	29.24%*	41.59%*	-8.76%	17.64%
Stooping lifting				
Mean	14.06%*	25.60%*	15.64%	9.54%
Max	25.62%*	30.31%*	37.61%*	12.38%
Walking				
Mean	0.57%	9.03%	-5.78%	5.34%
Max	0.72%	19.42%	-11.96%	15.89%

[1] ESI: Erector Spinae Iliocostalis; ESL: Erector Spinae Longissimus; RF: Rectus Femoris; BF: Biceps Femoris. The symbol * indicates a significant difference between experimental groups.

In summary, the performance of the BSE in lifting loads and walking is shown in Table II, which lists the percentage reduction in muscle activation that occurred when the exoskeleton was used. We found that, with the exception of the rectus fermoris muscle, all muscles reduced their activation when the exoskeleton was used for both squat and knee bend lifting, with the greatest reductions occurring in the ESI, up to 14% for mean activation and up to 29% for maximal activation, and in the ESL, up to 34% for mean activation and up to 41% for maximal activation. Lastly, as shown in Fig. 5 (c) and Table II, no significant changes occurred in back or leg muscles when walking with the BSE.

IV. DISCUSSION

In this work, we designed a spring-cable-differential to distinguish between lifting and walking. To ensure the user's autonomy during walking, the mechanism provides zero assistive torque when the average angle of the hip joints (θ_{ha}) is smaller than a predetermined threshold ($\Delta\theta/2$). For example, during lifting tests in the stooped posture, no assistive torque was provided to the subjects when the trunk flexion angle was less than 10°. Since the load on the lumbar spine is very low when the user stands almost straight [19], the zero assistive torque (when θ_{ha} is under 10°) had no effect on the assistive performance. Some active BSEs (e.g. APO [9]) specifically set their assistive torque to zero when θ_{ha} is small to prevent the device from pushing the human trunk into a hyperextension posture. There is a trade-off in choosing the value of $\Delta\theta$. Increasing $\Delta\theta$ may allow the user to move freely with more leg movements, but a large $\Delta\theta$ may affect the effect of assistance. In the future, we will conduct more muscle activity tests with different $\Delta\theta$ values to determine the relationship between $\Delta\theta$ and the effect of the assistance, which can support the use of the device under different working conditions. A limitation of the current design is that when $\Delta\theta$ needs to be adjusted,

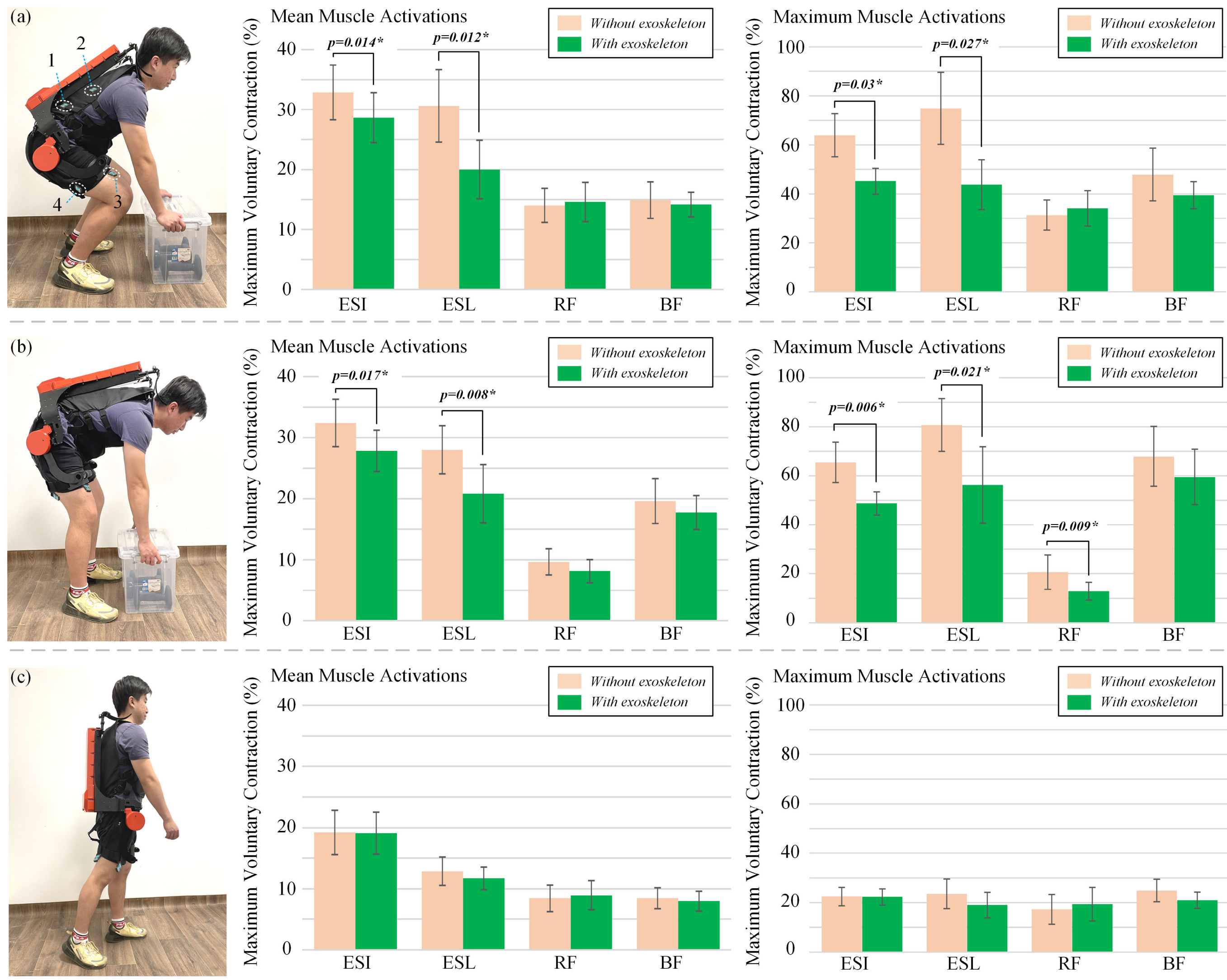

Fig. 5. Mean and maximum activations of 1- erector spinae illiocostalis (ESI), 2- erector spinae longissimus (ESL), 3- rectus femoris (RF), and 4- biceps femoris (BF) during (a) squat lifting, (b) stoop lifting and (c) walking. Error bars indicate standard deviation intervals, and the symbol * indicates a significant difference between experimental groups.

an engineer/technical person is needed to set different initial distances between the cable fixator l_{s0}. Therefore, our future work also includes developing a mechanism for users to manually adjust $\Delta\theta$ by themselves.

The proposed BSE is lighter than the BSE presented in [20] (4.5 kg) and slightly heavier than BackX (3.2 kg) [25]. Although the weight of our prototype (3.5 kg) is close to the design target of 3 kg [29], it is still a burden that cannot be ignored for users to carry. During prolonged operation, the weight of the device can drain the user's energy and lead to potential muscle fatigue. Therefore, further reducing the device weight through lighter materials and simpler designs will be a focus of our next work. In general, the weight of the elastic-band-based BSEs can be low thanks to the lightweight soft materials. For example, the Biomechanically-Assistive Garment in [18] weighs only 2 kg. However, these devices usually generate a large compression force on the human spine because the elastic bands run parallel to the back (with a small moment arm), which is uncomfortable for human shoulders and may even lead to spinal injuries [28]. The BSEs with springs and beams are slightly heavier, but use rigid links to convert the assistive torque into a force supporting the trunk, thus reducing the load on the spine.

When evaluated in the laboratory, muscle activity of the back muscles, i.e., erector spinae, appears to be the most commonly used indicator to determine the effectiveness of the exoskeleton. Other body regions (non-target body regions) showed different effects when an exoskeleton was used, e.g., leg muscle activity either increased [35], decreased [17], or remained unchanged [36]. We included RF and BF because these muscles could be affected by load transfer when wearing the exoskeleton. This study showed that muscle activity of the body region responsible for trunk extension (i.e., the back) was reduced by 12.8% to 41.6%, indicating that the exoskeleton successfully reduced muscle load in the target region during a simulated lifting task (i.e., squat and stoop lifting). These results are in good agreement with those of similar devices [20], [37], [38]. In addition to the expected positive effects in the target body segment, we found significant EMG reductions in non-target body segments during squat lifting, where RF maximal activation also reduced by 37.6%. This result indicates that the exoskeleton may also reduce the load on the leg muscles during squat lifting.

During the development of the exoskeleton, one of the considerations was to ensure that the wearer has sufficient freedom of movement during activities other than lifting loads (e.g. walking). During walking, the leg muscles (i.e., RF and BF) play a key role. The BSE has been shown to require minimal effort from the user when walking with the device, as evidenced by little to no change in activation of these muscles. Our results indicate that, unlike other devices, our BSE is able to realize a low resistance on legs in walking and causing minimal additional muscle strain, which is an important innovation in the development of our BSE.

V. Conclusion

This paper presents a complementary lifting solution for workers to improve their lifting abilities when performing manual lifting tasks. The structural design and prototype of the passive back support exoskeleton were developed and tested. Compared with other passive back exoskeletons designed to support lifting, the proposed exoskeleton has the advantage of providing adequate assistance in lifting but low resistance in walking. Experimental results show that the exoskeleton can reduce muscle strain on the erector spinae muscles.

One limitation of the proposed BSE is that the assistive torque is determined by the spring and cannot be adjusted for different loads in use. In the future, we will add a moment arm adjustment mechanism to facilitate users to adjust the assistive torque according to different loads. For the experimental results, although we observed generally good results during the simulation test when lifting a load in both the squat and stoop positions, the inclusion of training on the use of the BSE during lifting could further improve the utility of the lifting aid. For a large proportion of participants, both lifting and exoskeleton use were new; therefore, they needed to learn both lifting (stooping and squatting) and exoskeleton use. In addition, in a controlled laboratory setting, the obstacles of the normal work environment (e.g., crowded and confined work spaces, long work hours, shapes of load containers) may also influence the effects of exoskeleton use. Therefore, the next phase of research will focus on testing the exoskeletons in a variety of workplaces with skilled workers over an extended period of time. Another aspect of this work that needs improvement is testing the BSE under asymmetric lifting conditions. As an underactuated device that uses one spring to split the torque to two hip joints [39], the proposed BSE has the potential to assist asymmetric lifting tasks that are also common to industrial workers [40], [41]. The effectiveness of the device needs to be verified under asymmetric lifting conditions. In addition, comparing the performance of the BSE in symmetric and asymmetric lifting tasks is meaningful for improving the design of the device.

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