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A Passive Soft Wearable Suit With a Single Elastic Belt and Multiple Pulley System

Sangman Kim, Hanbo Zou, Sangeun Jin, and Junghan Kwon

***Abstract*—Exoskeleton robots promise to enhance safety by supporting workers' back strength during heavy lifting tasks, thereby improving work efficiency and productivity. However, the components of these robots, such as exoskeletal structures, actuators, and batteries, often increase their size and weight, which can reduce wearability and mobility. To tackle this issue, we propose a lightweight, passive wearable suit designed to assist back muscles during lifting tasks. The proposed system features a single elastic belt connected to multiple pulleys, which are attached to the back and lower limb sleeves. These pulleys are attached to the upper and lower limbs, and their relative distances change depending on body movements such as lifting or walking, thereby producing an effect similar to that of the moving pulley system. This innovative design allows the suit to deliver substantial support while efficiently distributing anchoring pressure across the wearer's skin during squatting and stooping positions. Additionally, the movement of belts through the pulleys minimizes the restrictions on gait motion compared to traditional designs. By adjusting the length of the belt, assist mode can be easily turned on and off, and flexibly applied to various body sizes. The supporting force is characterized by modeling and experimental tests. We evaluated the immediate effect of the prototype passively supporting back muscles during lifting tasks and reducing gait restriction during walking tasks.**

***Index Terms*—Soft robot applications, wearable robotics, physically assistive devices.**

I. Introduction

Low back pain is a common symptom in modern society, with around 600 million people globally experiencing disabilities due to it in 2020 [1]. It is more frequent among workers exposed to cumulative lumbar loads, such as those in manual handling jobs or awkward postures in industries like construction, transportation, healthcare, cleaning, and services [2]. Therefore, it is a critical public issue impacting worker health and industrial productivity [3].

During lifting tasks, the lumbar region experiences bending moments due to the weight of the upper body and the load, and it is counteracted by human's back and hip muscles. As the load increases, greater muscle strength is required, and excessive or repetitive use of the muscles can lead to musculoskeletal disorders [4]. Furthermore, the high compressive forces from short moment arms of internal tissues, combined with cumulative moments, contribute to lower back pain and disc disorders [5].

To address this issue, exoskeleton robots can be used to assist the muscle force by applying additional active moment to body joints [6], [7], and are applied in various areas such as rehabilitation [8], walking assistance [9], sports [10], and military applications [11], [12]. However, their components like rigid link structures, heavy actuators, and power sources increase the overall size and weight of the device. The complexity of the mechanical structures and electronic systems makes both manufacturing and repair challenging and costly. Moreover, design limitations in adapting to various body sizes may lead to misalignment with joint rotation axes, hindering natural movement and causing ergonomic challenges [13], [14].

A potential solution to these challenges is the use of soft materials. Soft wearable robots are often designed like clothing, utilizing materials such as fabric, cables, and elastomers. They are also integrated with soft actuators, including cable-driven actuators and pneumatic artificial muscles. Their flexible and lightweight design improves adaptability to the human body, enhancing both comfort and mobility [15], [16], [17].

Another design approach involves using passive components, such as springs and elastic belts, instead of actively powered actuators [18], [19]. These passive elements deform during joint flexion, generating a restoring force that supports muscles in maintaining a bent posture or during extension movements [20]. Additionally, the use of soft, flexible materials like elastic belts enables a design that is further lightweight and offers a suit-like comfort [4], in contrast to rigid structures or mechanical spring-based designs [21], [22], [23].

A commercial soft exosuit, LiftSuit2, has been introduced [24], featuring independently acting elastic bands that are attached only to both thighs. Due to this limited attachment, the design may struggle to reduce unintended tension during non-targeted movements such as walking, and may have difficulty ensuring even tension distribution that maximizes user comfort across various postures.

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This work involved human subjects or animals in its research. Approval of all ethical and experimental procedures and protocols was granted by the Institutional Review Board of Pusan National University (PNU IRB/2024 140 HR), and written informed consent was obtained.

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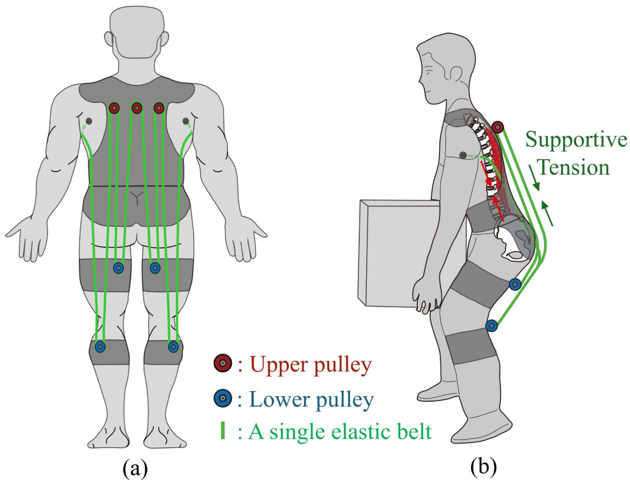


Fig. 1. Overview of the operating mechanism. (a) Routing a single elastic belt through multiple pulleys attached to body parts, (b) Generating supportive force during a weight-lifting task.

Despite these extensive research, several technical challenges remain in improving the design of passive soft wearable suits. First, the relatively low stiffness of elastic belts may provide insufficient support to the body due to their limited restoring force generation. Attaching multiple belts can increase support; however, each belt must be properly adjusted for its initial length, complicating setup and limiting adaptability to various body sizes. Secondly, if a belt is installed too loosely or too tightly, the auxiliary forces acting on each belt during operation may vary, potentially leading to concentrated pressure on certain areas of the body and reducing overall comfort. Lastly, while the design should effectively generate restorative force during lifting tasks, it must avoid interfering with non-targeted movements, such as walking.

To address these challenges, we propose a passive wearable suit design incorporating a single elastic belt and multiple pulley system (Fig. 1(a)). Similar to the moving pulley mechanism, the pulley mechanism can amplify force and evenly distribute anchoring pressure, offering both strong support and improved comfort during lifting tasks (Fig. 1(b)). The single belt is easily adjustable, enabling seamless activation of assistance mode and accommodating a wide range of body sizes. Furthermore, the movable belt mechanism can minimize restrictions on gait motion, while providing targeted assistance during lifting tasks.

II. Design and Fabrication

A. Effects of Multiple Pulley System

1) Assisting Force Amplification: The combination of multiple upper and lower pulleys creates a system in which a single rope is repeatedly arranged in parallel, allowing for a large lifting force with less rope tension. Thanks to this mechanical advantage, the pulley system has been widely and effectively used in industrial applications.

Inspired by this principle, we designed a soft wearable suit incorporating a pulley-based mechanism (Fig. 2(a)). The design includes three upper pulleys positioned on the back and four lower pulleys located on each thigh and shin. An elastic belt is anchored to a front chest buckle, then threaded alternately through the upper and lower pulleys, and finally secured at the opposite chest buckle.

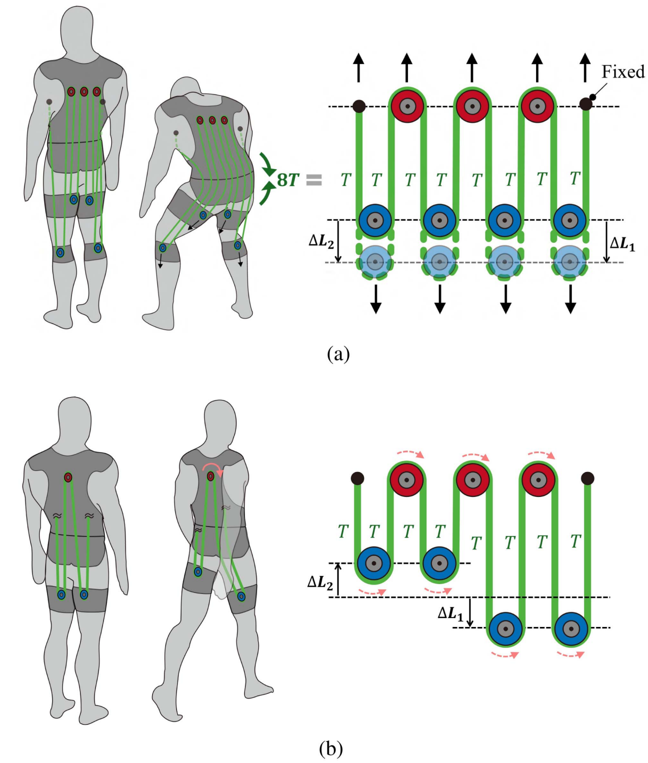


Fig. 2. Conceptual illustration of the effects of the proposed multiple pulley system. (a) Amplifications of assisting force and equalization of belt tension during lifting motion, (b) Reduced restriction of gait motion enabled by the smooth movement of the belt through the rollable pulleys.

When the waist is in flexion, the distance between the upper pulleys on the back and the lower pulleys on the lower-limb sleeves increases, causing the belt to stretch and generate tension, T. With the force generated across eight belt segments, an overall assistive force of $8T$ is produced. This design results in higher effective stiffness with the elastic belt, enabling the generation of substantial assistive force.

2) Equalizing Effect of Belt Tension: As the soft wearable suit offers greater support, the reaction force at the anchoring point also increases, causing discomfort due to the concentrated pressure on the wearer's skin. Therefore, it is crucial to design the suit to distribute the reaction force evenly across a larger area of the body.

The multiple pulley system is advantageous to achieve this. By routing the belt through multiple pulleys, the system ensures that the load is evenly distributed across each belt segment rather than being concentrated on one side. In our design, lower pulley buckles are attached to the sleeves worn around the thighs and shanks, as illustrated in Fig. 2(a). This configuration allows the reaction force to be symmetrically distributed across both thighs and knees, enhancing wearer comfort. The equalizing effects are still valid even if the sleeves worn on the thigh and shank have different heights and asymmetrically worn on both legs.

3) Allowing Movement of the Belt Through Pulleys: A single elastic belt passes through all upper and lower pulleys, allowing it to slide freely and permitting the belt to shift to one side under certain conditions. All belt segments stretch evenly during lifting movements, resulting in minimal shifting. However, during asymmetric lower-limb movements like walking, the belt can shift in a specific direction to reduce unwanted tension that could interfere with natural gait (Fig. 2(b)). A previous study proposed a design in which a belt extending from the back to the pelvis is connected via a pulley to another set of belts attached to thigh sleeves on both sides, effectively minimizing gait interference during walking [15]. Building on this concept, we extend the mechanism by placing multiple pulleys along the back panel and routing them to both the thighs and shanks. Unlike the previous design, which utilizes two separate belt segments, our system simplifies the structure by using a single continuous belt that connects multiple body regions.

Additionally, the single belt design enables simultaneous loosening (off-mode) or tightening (on-mode) of all segments by adjusting the belt length at the front anchoring buckles. This adjustment also makes it easy to accommodate individuals of varying body sizes.

B. Fabrication of a Prototype

Figure 3(a) shows a prototype of the soft wearable suit designed to demonstrate the efficacy of the proposed design. It is made of inexpensive and readily available fabric materials and clothing accessories. Its suit-like design ensures a lightweight structure, with a total weight of 1.5 kg, offering high comfort while reducing interference from the external environment. The prototype mainly consists of an upper-body vest, four lower-limb sleeves, and a single long elastic belt that connects the upper and lower limbs along the back (Fig. 3(b)).

The upper-body vest features three rollable buckles attached to the back panel, serving as fixed pulleys. The circular tube structure of the buckles allows for low-friction slip between the belt and the buckle. The vest is designed with wide shoulder straps, a back panel, and a hip pad, effectively dispersing the reaction force over a large area of the body. Additionally, the one-touch buckle on the chest strap and the Velcro-style waist belt make the suit easy to put on and take off, while also enabling quick and simple adjustments to fit various body sizes.

Four mesh fabric sleeves, fastened with Velcro, are positioned on both thighs and shanks below the knees. Each sleeve features a supporting rigid plate on the back, which serves as an attachment point for a lower pulley buckle. Multiple sleeves cover a wide area of the legs, effectively distributing supporting tension. The triangular-shaped design of the sleeve around the rigid pad attachment helps to prevent excessive pressure from being concentrated on specific areas.

The elastic strap belt material was selected for its light and thin profile with sufficient stiffness, which is commonly used to support the webbing base of sofas. The belt has a length of 3.7 m, a width of 50 mm, and a thickness of 0.2 mm. Similar to the pulley system of a crane, the belt is alternately connected to four moving pulleys and three fixed pulleys. The hooks acting as moving pulleys are secured to the rigid plate on the leg sleeves. This profile allows a single long belt to create eight parallel segments along the back of the body, all of which stretch when squatting to lift a heavy load (Fig. 3(c)). Additionally, to aid in alignment and prevent the belt from shifting sideways during lifting postures, the belt is designed to pass through a channel at the hip pad of the upper-body vest.

Fig. 3. Photos of the proposed wearable suit. (a) Key components of the prototype (b) Front, side, and back view of the prototype being worn, (c) Demonstration of lifting a heavy box with supportive tension by the suit.

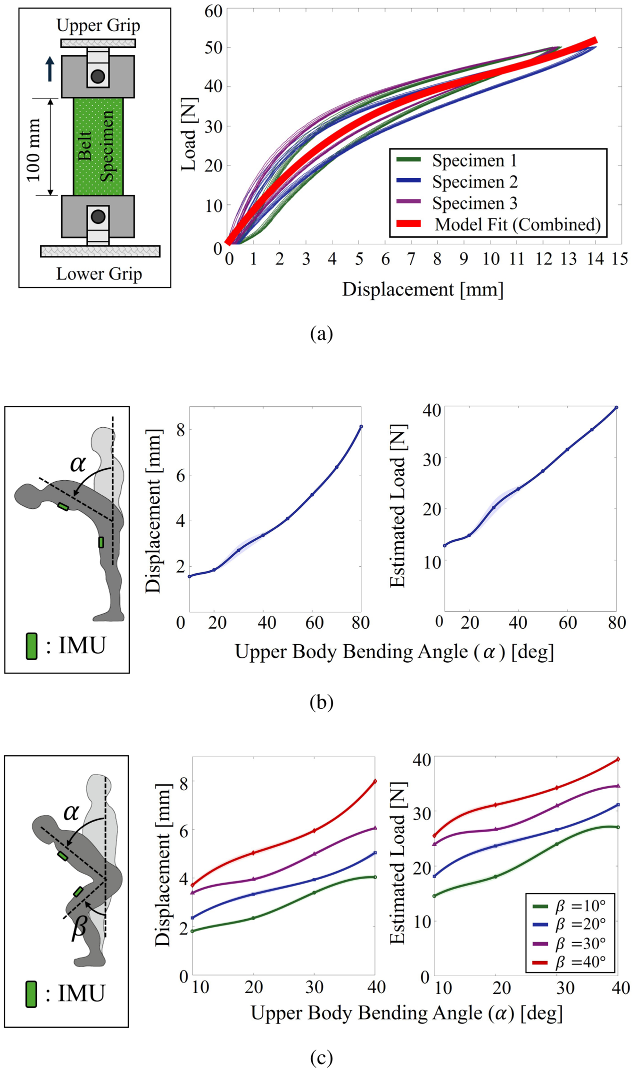


Fig. 4. Supportive forces estimation of the prototype. (a) belt tension and stretch displacement relation of the 100mm belt specimens. The red bold line is the cubic fit from measured data, (b) Measured belt stretches and estimated assisting force at stoop postures, (c) Measured belt stretches and estimated assisting force at squat postures.

Both ends of the elastic belt terminate at adjustable buckles attached to the vest's front chest area. Wearers can adjust the initial length of the belt to fit their body size and, if needed, loosen it to disengage the assistive mode. This front-facing adjustment design is convenient and simple, without any specialized mechanical components like a locking latchet device. The assistive mode can also be disengaged by partially doffing the vest's shoulder straps.

III. Performance Characterization

A. Supportive Force Estimation

To estimate the supportive force provided by the suit during the lifting movement, we first measured the mechanical properties of the elastic belt (Fig. 4(a)). A 100 mm belt specimen was mounted on a motorized test stand (QM100SS, QMESYS). The specimen was then slowly stretched to a tension of 50 N, extending it by approximately 12 to 14 mm, and it returned to its initial length. This extension-contraction cycle was repeated 10 times for each specimen, and the process was conducted with three different specimens to characterize the force-length relationship of the elastic belts, as shown in Fig. 4(a). Due to the nonlinear behavior exhibited by the elastic belt, the data were fitted using a third-order nonlinear spring model as follows:

$$T = 9.292\Delta l - 0.741\Delta l^2 + 0.024\Delta l^3 \tag{1}$$

where T is the tension of the belt specimen, and Δl is the displacement of the belt specimen from its initial length.

Next, the belt stretch resulting from body flexion movements was measured, and the corresponding assistive force was estimated. It was performed on a male participant who was 178 cm tall and weighed 78 kg. The deformation of the belt on the posterior back was measured in relation to the flexion angles of the upper and lower body during stoop and squat postures.

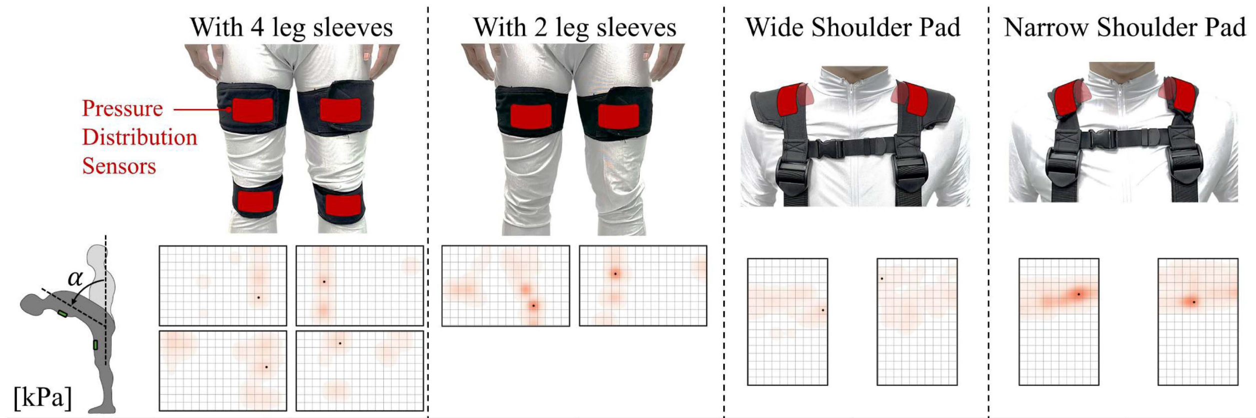
In the stoop posture, the upper body flexion angle varied from 0° to 80°, while the lower body angle was kept constant. In the squat posture, both the upper and lower body flexion angles were varied from 0° to 40°. Measurements were taken at 10° intervals, and all variables were randomly sequenced, with each condition repeated three times. Then, the supportive force generated by a single belt segment was estimated based on the measured data using the force-displacement model for a 100 mm segment in Equation (1).

The experimental results shown in Fig. 4(b) and 4(c) indicated a gradual increase in the magnitude of supportive force as the upper body flexion angle increased. In the stoop posture with the upper body flexion angle $\alpha = 80°$, the restoring force acting on a single belt segment was calculated to be approximately 39.7N. Since the prototype incorporates eight segments through the pulley system, the total assistive force is estimated to be approximately 320 N. In the squat posture with $\alpha = \beta = 40°$, the restoring force acting on a single belt segment was calculated to be approximately 39.4N, and the total assistive force was similarly estimated to be 320N.

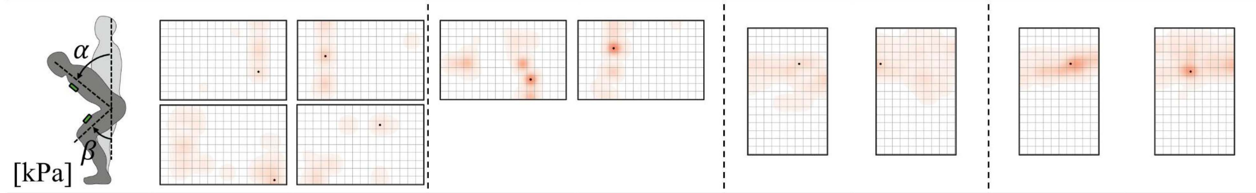
B. Pressure Distribution Effect

Experiments measuring pressure distribution at the anchoring points were performed to verify the effectiveness of distributing reaction forces across the body using a movable pulley mechanism(Fig. 5). We evaluated the effects of splitting the lower body's anchoring points into the thigh and shin regions and applying a wide-area shoulder pad.

Pressure distribution sensors (MS9723, KITRONYX) were placed on the surfaces of the thigh, shank, and shoulder. Additionally, Inertial Measurement Unit(IMU) sensors were attached to the upper body and both thighs to ensure accurate posture during repeated experiments. The experiments were conducted in the Stoop posture (waist flexion angle α of 60°, knee flexion angle β of 0°) and the Squat posture (waist flexion angle α of 45°, knee flexion angle β of 20°), respectively. Each experimental condition was a combination of variables (Stoop/Squat, Shank Sleeve On/Off, Shoulder Pad On/Off). The peak and average



	With 4 leg sleeves		With 2 leg sleeves		Wide Shoulder Pad		Narrow Shoulder Pad	
Peak Pressure	3.98 (1.24)	6.29 (1.08)	16.91 (1.24)	14.98 (0.13)	2.47 (0.40)	2.25 (0.13)	12.38 (0.63)	11.81 (2.47)
	3.38 (0.54)	3.86 (0.72)						
Mean Pressure	1.86 (0.52)	2.18 (0.13)	3.37 (0.61)	2.73 (0.17)	1.01 (0.07)	1.10 (0.12)	3.04 (0.23)	1.64 (0.15)
	1.22 (0.09)	1.61 (0.23)						



	With 4 leg sleeves		With 2 leg sleeves		Wide Shoulder Pad		Narrow Shoulder Pad	
Peak Pressure	3.14 (0.40)	5.71 (0.13)	17.14 (0.13)	16.48 (0.26)	3.61 (0.77)	4.36 (1.03)	11.25 (0.72)	10.17 (2.36)
	2.69 (0.49)	2.47 (0.23)						
Mean Pressure	1.75 (0.03)	1.98 (0.18)	3.36 (0.30)	2.71 (0.17)	1.10 (0.15)	1.13 (0.07)	3.09 (0.15)	1.68 (0.14)
	1.14 (0.12)	1.19 (0.09)						

Fig. 5. Experimental results for evaluating the pressure distribution effect. Pressure maps at anchoring points (shoulder, thighs, and shanks) and their peak and mean pressure values for (a) a stoop posture, and (b) a squat posture.

pressure values for each posture were calculated from the measured data. Each condition was repeated three times, and the mean and standard deviation(SD) were calculated.

The results showed that wearing both thigh and shank sleeves reduced the peak pressure values by an average of 67.2% in the Stoop posture and 73.6% in the Squat posture than the case of wearing only thigh sleeves. Similarly, the mean pressure values were also reduced when wearing both thigh and shank sleeves, with decreases of 32.4% in the Stoop posture and 49.8% in the Squat posture compared to wearing only thigh sleeves. The wide shoulder pad also reduced the peak pressure values of the narrow pad by 80.4% in the Stoop posture and 62.5% in the Squat posture. Additionally, the mean pressure values were reduced by 37.4% in the Stoop posture and 48.6% in the Squat posture compared to the narrow pad.

This reduction in peak and mean pressure values indicates that the proposed design effectively disperses reaction forces generated by assistive forces. Consequently, this alleviates pain at the anchoring points and enhances the comfort of the wearer.

C. Less-Restriction of Walking Movement

In the initial standing posture (Fig. 6(a)), the belt lengths on both legs are equal, and the initial tension is set to zero with the belt segment length of L_0. When the wearer steps forward with the leading foot, the belt on the leading leg requires a geometrically longer path than the one on the trailing leg.

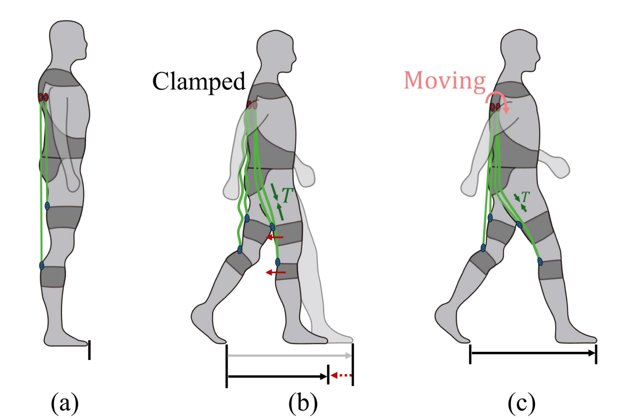


Fig. 6. Movement of the belts during waking. (a) Initial standing posture without belt tension, (b) The movement of the belt through the pulley reduces the generation of unwanted tension. (c) If each belt component is independently anchored, the tension in the leading leg restricts walking motion while the belt components anchored at the back leg are loosened.

If each belt segment is assumed to be clamped at each anchoring point, resistive tension would be generated by the suit on the

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leading leg, while the belt on the trailing leg loosens (Fig. 6(b)). This unwanted tension potentially hinders walking motion. In the Clamped condition shown in Fig. 6(b), the belt segments connected to the leading leg deform to ΔL_1 during the stepping motion. The strain per 100 mm of the elastic belt, $\Delta l_{clamped}$ [mm], can be derived as follows.

$$\Delta l_{clamped} = \frac{\Delta L_1}{L_0} \cdot 100 \tag{2}$$

Then, the tension acting on the belt in the clamped condition, T_C can be estimated by substituting $\Delta l_{clamped}$ into (1).

$$T_C = 9.292\Delta l_{clamped} - 0.741\Delta {l_{clamped}}^2 + 0.024\Delta {l_{clamped}}^3 \tag{3}$$

As a result, a resistive force of $4T_C$ is generated by the four belt segments connected to the leading leg. This force may limit the forward motion of the leading leg during walking, reducing the step length and potentially causing a decline in walking performance, as well as an increase in unnecessary energy consumption.

On the other hand, our proposed design allows the belt to move between the left and right legs (referred to as the Moving condition) during walking. This enables the belt to shift from the trailing leg toward the leading leg (Fig. 6(c)). While the length of the belt for the leading leg changes by ΔL_1, the length of the belt for the trailing leg changes by ΔL_2. Since the same number of belt segments is placed on both legs, the average change in the segment length is calculated as $(\Delta L_1 + \Delta L_2)/2$. Accordingly, the strain per 100mm of the elastic belt in the Moving condition, Δl_{moving}, is given as follows.

$$\Delta l_{moving} = \frac{\left(\frac{\Delta L_1 + \Delta L_2}{2}\right)}{L_0} \cdot 100 \tag{4}$$

By substituting this into (1), the walking resistive force T_M is derived as follows.

$$T_M = 9.292\Delta l_{moving} - 0.741\Delta {l_{moving}}^2 + 0.024\Delta {l_{moving}}^3 \tag{5}$$

During walking, the displacement of the belt connected to the leading leg, ΔL_1, is larger than ΔL_2, resulting in $\Delta l_{clamped} > \Delta l_{moving}$ and, consequently, $T_C > T_M$. Therefore, the proposed design is able to reduce the overall tension generated in the belts, minimizing interference with leg movement during walking, resulting in a decrease in discomfort and less restriction of walking efficiency.

IV. Validation With Human Participants

A. Experimental Setup and Procedure

An human subject test was performed to investigate the biomechanical effects of the proposed passive soft wearable suit during manual lifting and walking tasks.

Four male participants (27.5 ± 0.58 years, 176.5 ± 2.38 cm, 76.25 ± 6.65 kg) volunteered, with no history of low back pain or musculoskeletal disorders. The study protocol was approved by the Institutional Review Board of Pusan National University (PNU IRB/2024 140 HR), and written informed consent was obtained.

Electromyography (EMG) data were collected at 1200 Hz using a surface-EMG system (Bagnoli, Delsys, USA) to measure the activity of the lumbar erector spinae (LES) and gluteus maximus (GM). Two surface electrodes were placed on the right side of LES and GM. EMG data for the two target muscles were collected by using a Roman chair and multipurpose sit-up bench before the experimental tasks to capture the maximum voluntary contraction (MVC). All the raw EMG data were full-wave rectified and filtered using a high-pass filter at 20 Hz, a low-pass filter at 450 Hz, and a notch filter at 60 Hz. Then, a 4th order and 5 Hz cut-off frequency Butterworth filter smoothed the rectified EMG data. The same process was applied to EMG data measured in the MVC trials. The normalized EMG (NEMG, unit in %MVC) for each target muscle was calculated as the ratio of the 50% percentile value of the EMG data from the test sessions (the numerator) to the peak EMG value from MVC trials for the denominator.

In the lifting operation, participants started in a posture for grasping a 10 kg object positioned 30 cm above the ground and performed a symmetric lift in the following order.

1) Lifting with load
2) Lowering with load

Each step lasted 3 seconds, with a 3-second static posture between steps.

Trials were conducted under randomized conditions:

- Suit Usage: With/without the Soft wearable suit or Rigid passive exoskeleton
- Posture: Stoop or squat

Each case was repeated three times, with the order of trials randomized to reduce potential bias and learning effects.

In the walking experiment, participants walked on a treadmill at a constant speed of 4.5 km/h for a distance of 100 m under four conditions, while wearing IMUs placed at L1, S1, and both thighs:

- without a suit
- with a rigid passive exoskeleton device (HARD)
- with a suit with tightly clamped belt segments (SOFT1)
- with a suit featuring freely movable pulleys (SOFT2)

SOFT2 represents the actual proposed design, where the belt can freely slide through pulleys to redistribute tension. In contrast, SOFT1 is an artificially constrained condition, in which clamps were used to prevent belt movement, mimicking a structure with fixed-length belt segments for comparison purposes.

For each trial, conditions were randomly assigned and repeated three times. The angle between both thighs was measured using an IMU to evaluate the restriction imposed by the exoskeleton.

A commercially available back-mounted rigid exoskeleton device (StepUp, FRT Robotics) was also evaluated for comparison with the proposed soft exosuit. The HARD exoskeleton weighs 4.3 kg and generates assistive force in proportion to the trunk flexion angle when the assist mode is on. Unintended tension is applied to the legs, if walking is performed while the assist mode remains on.

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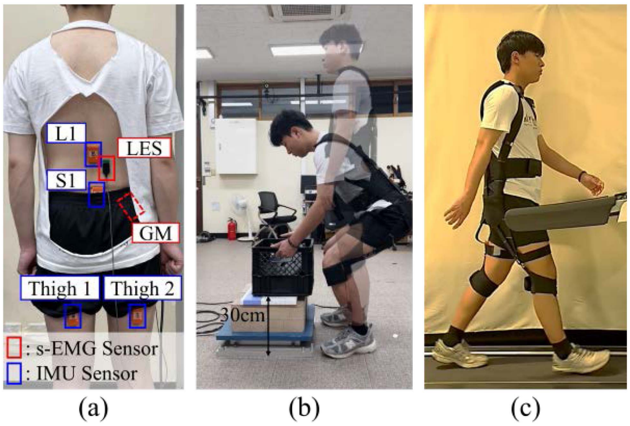

Fig. 7. Experimental setup for human subject tests. (a) s-EMG sensor and IMU sensor placement, (b) lifting task, (c) walking task.

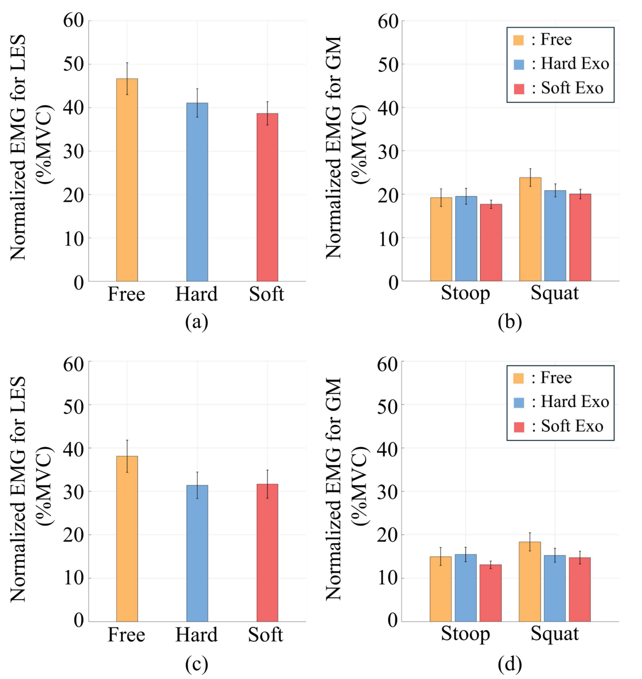

Fig. 8. Results of the lifting tasks. (a) LES main effects of lifting task, (b) GM interaction diagram for lifting task, (c) LES main effects of lowering task, (d) GM interaction diagram for lowering task.

B. Muscle Activations During Lifting Task

Fig. 8 a through Fig. 8 d illustrate the effects of assistive device type (FREE, HARD exoskeleton, SOFT exoskeleton) and posture (Squat, Stoop) on normalized electromyographic (EMG) activity (%MVC) of the lumbar erector spinae (LES) and gluteus maximus (GM) muscles during both lifting and lowering tasks. With respect to LES activation, Fig. 8(a) and Fig. 8(c) consistently demonstrate reduced muscle activity under both exoskeleton conditions. During the lifting task (Fig. 8(a)), LES activation decreased from 43.84%MVC (FREE) to 39.54%MVC (HARD) and 38.05%MVC (SOFT) in the squat posture, and from 49.47% MVC (FREE) to 42.87%MVC (HARD) and 42.57%MVC (SOFT) in the stoop posture. A similar trend was observed in the lowering task (Fig. 8(c)), where LES

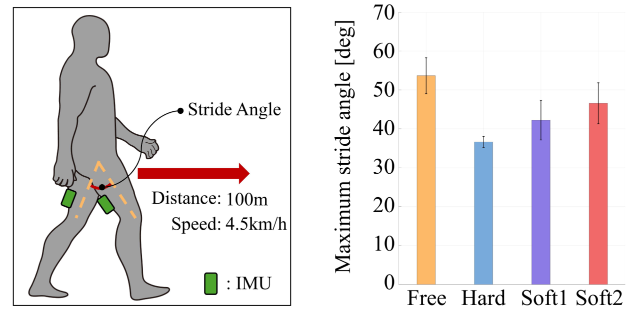

Fig. 9. Results of walking tasks.

activity declined from 39.68%MVC (FREE) to 34.84%MVC (HARD) and 37.30%MVC (SOFT) in the squat posture, and from 36.50%MVC (FREE) to 28.60%MVC (HARD) and 29.87%MVC (SOFT) in the stoop posture. These findings suggest that both exoskeletons significantly reduce low back muscle loading, with the HARD exoskeleton exhibiting superior performance during lowering, and the SOFT exoskeleton performing better during lifting.

Regarding GM activation, Fig. 8(b) and Fig. 8(d) reveal more pronounced posture-dependent effects. During the lifting task (Fig. 8(b)), GM activation peaked under the FREE condition in the squat posture (23.39%MVC), followed by the HARD (20.88%MVC) and SOFT (20.06%MVC) exoskeletons. In the stoop posture, GM activation was reduced across all conditions, with the SOFT exoskeleton showing the lowest value (17.71%MVC). A similar trend was observed during the lowering task (Fig. 8(d)): in the squat posture, the FREE condition again elicited the highest activation (18.39%MVC), while the SOFT (14.75%MVC) and HARD (15.29%MVC) exoskeletons significantly reduced GM activity. During stoop lowering, the SOFT condition again resulted in the lowest GM activation (13.10%MVC), outperforming both the FREE (15.00%MVC) and HARD (15.47%MVC) conditions.

In summary, both exoskeletons demonstrated clear physiological benefits by reducing muscular effort in the LES and GM. Although the HARD exoskeleton provides greater structural support, it may also limit the user's range of motion and wearing comfort. In contrast, the SOFT exoskeleton, while offering slightly less structural assistance, may provide a more comfortable and adaptable user experience during dynamic tasks due to its enhanced flexibility and compliance.

C. Effects on Walking Movement

The results of the walking experiment revealed a decrease in maximum stride angle when wearing exoskeletons compared to the FREE condition, indicating that the three exoskeleton conditions limited walking efficiency. The stride angle reduction was 17.04° in HARD, 11.44° in SOFT1, and 7.10° in SOFT2 (Fig. 9). Nonetheless, the decrease in maximum stride angle in the SOFT2 conditions was significantly less than in the other two exoskeleton conditions, indicating that walking efficiency when using the SOFT2 was significantly better than that of the HARD and SOFT1 conditions.

These findings indicate that the movement of the belts plays an important role in reducing unwanted tension generation, thereby reducing walking restrictions.

V. Discussion and Conclusion

In this study, we proposed a design integrating a movable pulley mechanism into the soft wearable suit. In comparison to previous studies, our design can provide enhanced supporting force during lifting tasks while evenly distributing reaction tension over a larger area of the wearer's skin due to the load amplification and equalization effects of the pulley system. Furthermore, the design supports lifting while minimizing interference with walking, making it particularly effective for tasks requiring repeated transitions between the two. During the rest state, the assistance mode can also be easily switched by loosening or tightening the initial length of the belt. These benefits arise from the unique mechanical advantages of the embedded pulley system in the proposed suit design. Overall, the approach significantly improves the performance of the soft wearable suits, while also being lightweight, easy to fabricate, and cost-effective. The effects of wearing the proposed suit are also investigated through experimental characterization and human subject tests.

Assistive force modeling can be further improved by examining the effects of tilted belt routing and friction, as these factors may involve complex interactions between the components. Additionally, our current tests included only four subjects to assess the immediate effects of the suit. Therefore, future studies should involve a larger sample size and investigate longer-term effects.

References

[1] F. Fatoye, T. Gebrye, C. G. Ryan, U. Useh, and C. Mbada, "Global and regional estimates of clinical and economic burden of low back pain in high-income countries: A systematic review and meta-analysis," *Front. Public Health*, vol. 11, 2023, Art. no. 1098100.

[2] A. Petit and Y. Roquelaure, "Low back pain, intervertebral disc and occupational diseases," *Int. J. Occup. Saf. Ergonom.*, vol. 21, no. 1, pp. 15–19, 2015.

[3] J. Hartvigsen et al., "What low back pain is and why we need to pay attention," *Lancet*, vol. 391, no. 10137, pp. 2356–2367, 2018.

[4] E. P. Lamers, A. J. Yang, and K. E. Zelik, "Feasibility of a biomechanically-assistive garment to reduce low back loading during leaning and lifting," *IEEE Trans. Biomed. Eng.*, vol. 65, no. 8, pp. 1674–1680, Aug. 2018.

[5] P. Coenen, I. Kingma, C. R. Boot, P. M. Bongers, and J. H. van Dieën, "Cumulative mechanical low-back load at work is a determinant of low-back pain," *Occup. Environ. Med.*, vol. 71, no. 5, pp. 332–337, 2014.

[6] K. Huysamen, M. de Looze, T. Bosch, J. Ortiz, S. Toxiri, and L. W. O'Sullivan, "Assessment of an active industrial exoskeleton to aid dynamic lifting and lowering manual handling tasks," *Appl. Ergonom.*, vol. 68, pp. 125–131, 2018.

[7] T. Zhang and H. Huang, "A lower-back robotic exoskeleton: Industrial handling augmentation used to provide spinal support," *IEEE Robot. Automat. Mag.*, vol. 25, no. 2, pp. 95–106, Jun. 2018.

[8] G. Colombo, M. Joerg, R. Schreier, and V. Dietz, "Treadmill training of paraplegic patients using a robotic orthosis," *Rehabil. Res. Develop.*, vol. 37, no. 6, pp. 693–700, 2000.

[9] A. Kapsalyamov, P. K. Jamwal, S. Hussain, and M. H. Ghayesh, "State of the art lower limb robotic exoskeletons for elderly assistance," *IEEE Access*, vol. 7, pp. 95075–95086, 2019.

[10] R. Nasiri, A. Ahmadi, and M. N. Ahmadabadi, "Reducing the energy cost of human running using an unpowered exoskeleton," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 26, no. 10, pp. 2026–2032, Oct. 2018.

[11] J. K. Proud et al., "Exoskeleton application to military manual handling tasks," *Hum. Factors*, vol. 64, no. 3, pp. 527–554, 2022.

[12] A. Zoss, H. Kazerooni, and A. Chu, "Biomechanical design of the Berkeley lower extremity exoskeleton (BLEEX)," *IEEE/ASME Trans. Mechatron.*, vol. 11, no. 2, pp. 128–138, Apr. 2006.

[13] M. B. Näf, K. Junius, M. Rossini, C. Rodriguez-Guerrero, B. Vanderborght, and D. Lefeber, "Misalignment compensation for full human-exoskeleton kinematic compatibility: State of the art and evaluation," *Appl. Mechan. Rev.*, vol. 70, 2019, Art. no. 050802.

[14] N. Jarrasse and G. Morel, "Connecting a human limb to an exoskeleton," *IEEE Trans. Robot.*, vol. 28, no. 3, pp. 697–709, Jun. 2012.

[15] J. Chung et al., "Lightweight active back exosuit reduces muscular effort during an hour-long order picking task," *Commun. Eng.*, vol. 3, no. 1, 2024, Art. no. 35.

[16] C. Thakur, K. Ogawa, T. Tsuji, and Y. Kurita, "Soft wearable augmented walking suit with pneumatic gel muscles and stance phase detection system to assist gait," *IEEE Robot. Automat. Lett.*, vol. 3, no. 4, pp. 4257–4264, Oct. 2018.

[17] X. Yang et al., "Spine-inspired continuum soft exoskeleton for stoop lifting assistance," *IEEE Robot. Automat. Lett.*, vol. 4, no. 4, pp. 4547–4554, Oct. 2019.

[18] A. Ali, V. Fontanari, W. Schmoelz, and S. K. Agrawal, "Systematic review of back-support exoskeletons and soft robotic suits," *Front. Bioeng. Biotechnol.*, vol. 9, 2021, Art. no. 765257.

[19] J. In Kim, J. Choi, J. Kim, J. Song, J. Park, and Y. -L. Park, "Bilateral back extensor exosuit for multidimensional assistance and prevention of spinal injuries," *Sci. Robot.*, vol. 9, no. 92, 2024, Art. no. eadk6717.

[20] J. Ahn, H. Jung, J. Moon, C. Kwon, and J. Ahn, "A comprehensive assessment of a passive back support exoskeleton for load handling assistance," *Sci. Reports*, vol. 15, no. 1, 2025, Art. no. 3926.

[21] S. J. Baltrusch, H. Houdijk, J. H. van Dieën, and J. T. C. M. de Kruif, "Passive trunk exoskeleton acceptability and effects on self-efficacy in employees with low-back pain: A mixed method approach," *J. Occup. Rehabil.*, vol. 31, no. 1, pp. 129–141, 2021.

[22] A. S. Koopman, I. Kingma, G. S. Faber, M. P. de Looze, and J. H. van Dieën, "Effects of a passive exoskeleton on the mechanical loading of the low back in static holding tasks," *J. Biomech.*, vol. 83, pp. 97–103, 2019.

[23] E. P. Lamers and K. E. Zelik, "Design, modeling, and demonstration of a new dual-mode back-assist exosuit with extension mechanism," *Wearable Technol.*, vol. 2, e1, pp. 1–26, 2021.

[24] A. Cardoso et al., "Assessing the short-term effects of dual back-support exoskeleton within logistics operations," *Safety*, vol. 10, no. 3, 2024, Art. no. 56.