EXOSUIT

Body-powered variable impedance: An approach to augmenting humans with a passive device by reshaping lifting posture

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The movement patterns appropriate for exercise and manual labor do not always correspond to what people instinctively choose for better comfort. Without expert guidance, people can even increase the risk of injury by choosing a comfortable posture rather than the appropriate one, notably when lifting objects. Even in situations where squatting is accepted as a desirable lifting strategy, people tend to choose the more comfortable strategy of stooping or semisquatting. The common approach to correcting lifting posture, immobilizing vulnerable joints via fixation, is insufficient for preventing back injuries sustained from repetitive lifting. Instead, when lifting small but heavy objects, the entire kinetic chain should cooperate to achieve a series of squat-lifting patterns. Inspired by the observation that force fields affect the coordination of voluntary human motion, we devised a passive exosuit embedded with a body-powered variable-impedance mechanism. The exosuit adds impedance to the human joints according to how far the wearer's movement is from the squat-lifting trajectories so that it hinders stooping but facilitates squatting. In an experiment that entailed lifting a small 10-kg box, 10 first-time users changed their voluntary lifting motion closer to squatting on average. Simulation results based on recorded kinematic and kinetic data showed that this postural change reduced the compression force, shear force, and moment on the lumbosacral joint. Our work demonstrates the potential of using an exosuit to help people move in a desirable manner without requiring a complicated, bulky mechanical system.

INTRODUCTION

Manual work, sports, exercise, and everyday activities involve motor tasks that can lead to injury if not performed in a proper way. Tasks that involve lifting have a particularly high rate of causing musculoskeletal injury (*1*, *2*). Contrary to the common belief that taking a stooping posture at lifting causes spinal injuries, the relationship between posture and injury has not been clarified (*3*–*5*). However, at least, researchers in the field of biomechanics have found that the magnitude of back compression force (BCF) during the lift process can vary depending on the various settings, such as wearer's posture and object size and shape (*6*–*8*). Kingma *et al.* (*9*) compared different lifting techniques in various object conditions and showed that squatting reduces BCF more effectively than stooping when lifting an object small enough to pass between the legs. However, when the object is large, squatting may cause higher BCF than other lifting techniques, such as stooping, kneeling, and weight lifters' technique (*9*, *10*). The U.S. National Institute of Occupational Safety and Health (NIOSH) recommends that manual workers should use context-appropriate lifting techniques to avoid injury (*11*): (i) If the object is small, a worker should keep the back straight and keep the target object close to the body by widening the knees laterally and flexing them as much as possible (i.e., squat to lift); (ii) if the object is large, a worker should place one knee on the floor and lift the object using the kneeling technique. In environments where small or specially shaped objects are handled close to the body—such as some industrial settings (*12*), gyms (*13*), and rehabilitation centers (*14*)—squat technique is accepted as a desirable lifting strategy.

However, people tend to choose comfortable lifting behaviors, such as stooping and semisquatting, when there is no specific instruction or training. It has been consistently argued that human movement patterns are planned by an innate optimization algorithm that takes into account certain cost functions, although any universal cost has never been clearly revealed; the cost may include such dynamic variables as muscle activation level (*15*, *16*), energy expenditure (*17*–*19*), torque change (*20*), and jerk (*21*). Perceptual costs and constraints also play a role (*22*, *23*). In motor tasks, including lifting, choosing a desirable movement often counters the instinct of humans, which prefers comfortable movements. In some industrial settings, economic costs are expended on adjusting the layout of the worksite or training the target group to control their lifting behavior (*24*). However, the low efficacy of the training, pointed out in some studies (*25*–*27*), calls for interventions that can directly and immediately induce people to lift items with more desired form.

To date, motion correction strategies have been mostly focused on static methods such as fixing or hindering the use of vulnerable joints (*28*). For example, some workers wear back belts that compress the trunk in radial direction to stabilize the spine. However, the usefulness of the back belt in correcting workers' lifting behavior or preventing injuries has been questioned by several studies (*28*–*30*). Today's evolved version of the back belt corresponds to a back-assistive wearable robot. These devices attach actuators (*31*–*33*) or springs (*34*–*37*) with a large moment arm to the torso to reduce muscle stress and spine compression during lifting. However, modifying users' lifting motion pattern has deviated from their main concerns. To ensure that people use a particular desirable motion pattern for a movement, simply interfering or assisting certain single joint is not sufficient. Instead, proper coordination of all joints in the involved kinetic chain is required.

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Our focus in this study is on finding a way to induce people to voluntarily adopt a specific posture to lift objects by artificially changing the force field surrounding the body. The properly designed force field can intervene in the dynamic motion pattern generated by all joints in the kinetic chain to induce any desired movement. Several studies have found that artificially changing the force field of a person's workspace changes their voluntary movement patterns (*38*–*40*). Robotic systems have used this principle to provide movement training for patients with neurological disorders. These robots use impedance-based control to create a force field in which a restoring force is applied when the user's movements deviate from the desired trajectory (*41*–*45*). However, many of these robots are large and heavy because they require motors to generate the restorative forces, making them difficult to use in daily life and the workplace.

The force field that people feel in their movements is not only determined by externally actuated force but also influenced by intrinsic musculoskeletal structures. When those structures are damaged by disease or injury, people are often able to adapt them into atypical movement patterns. For example, people with hereditary myopathies in which proximal weakness is a dominant feature exhibit waddling gait patterns because the calf muscles compensate for weakened hip flexors and lumbar hyperlordosis due to weak trunk muscles (*46*, *47*). This suggests that artificially placing tendons on a person's body to interfere with or limit joint movement can be a way to form a force field.

Our proposed solution for a lightweight wearable device that can facilitate specific motion incorporates the artificial bi-articular tendon, which is inspired by natural examples of muscles that cross two joints at once. Placing an artificial tendon across two joints couples the angles of the joints such that when one joint rotates, the tendon becomes taut and pulls the other joint, causing it to rotate. If one joint is fixed, then the tendon also restricts rotation of the other joint. Such artificial bi-articular tendons are applied to body-powered devices that allow a person to use one or more healthy joints to control a disabled one (*48*, *49*). They have also been used in robotic exosuits that use a single motor to assist movement in multiple joints (*50*, *51*). However, what would happen if a device's bi-articular tendon crossed two healthy joints? Because the tendon path is simultaneously affected by the angle of the two joints, the tendon may become taut or loose depending on the position of both joints. As a result, the person wearing the device experiences different levels of resistance in different joint chain postures.

An important factor in determining the performance of a wearable tendon system is its impedance. If tendon impedance is too high, then it causes discomfort that hinders movement, especially during dynamic motion, but if the impedance is too low, then it is difficult to limit movement to a desired range. A clutch system could be used to implement a device that can vary impedance with motion (*52*). However, if mostly soft materials could be used to passively change the impedance, then a new type of wearable device would be possible that could be incorporated as part of a garment.

In this study, we present an artificial tendon morphology that can design a desired force field on a wearable device without requiring a power source (Movie 1). The new tendon morphology can be incorporated into a passive full-body exosuit that uses the wearer's movements to vary the impedance that the suit places on target joints. This idea is inspired by human musculoskeletal anatomy, which allows the impedance of joints to be adjusted for different environments (*53*, *54*). Among various motor tasks, we first pay attention to lifting, which frequently results in injuries when performed with undesirable form (*1*, *2*), and among various lifting patterns, we focus on squatting, which is considered as the desirable form in various sites (*12*–*14*). To create a force field that hinders stooping to lift but facilitates squatting, we devised and installed bi-articular tendons that cross behind the wearer's back, hip, and knee joints. The tendons are pulled when the back and hip are flexed and released when the knees are flexed (Fig. 1 and figs. S15 to S17).

Movie 1. Overview of unpowered soft exosuit with body-powered variable impedance. This video summarizes the principles and performance of the developed exosuit.

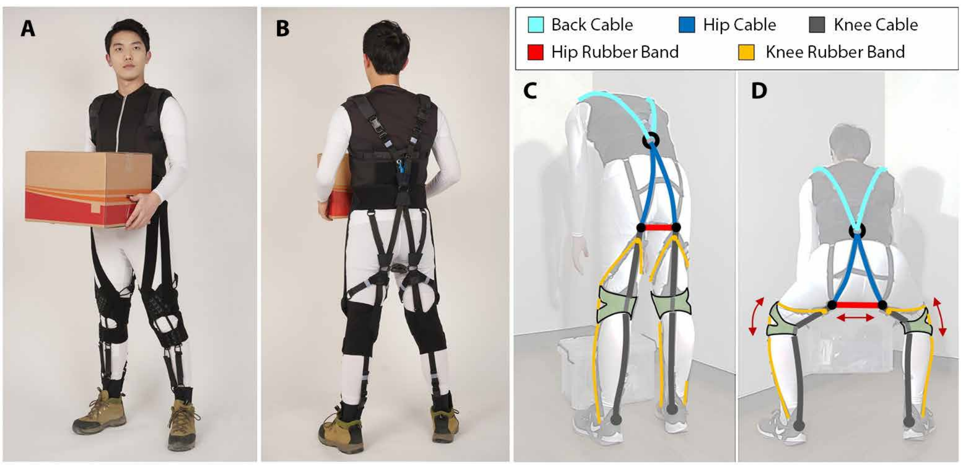

Fig. 1. Configuration of the passive exosuit with body-powered variable impedance. Individual wearing exosuit and holding a box photographed from the front (**A**) and the back (**B**). (**C**) Stooping to pick up the object. (**D**) Squatting to pick up the object.

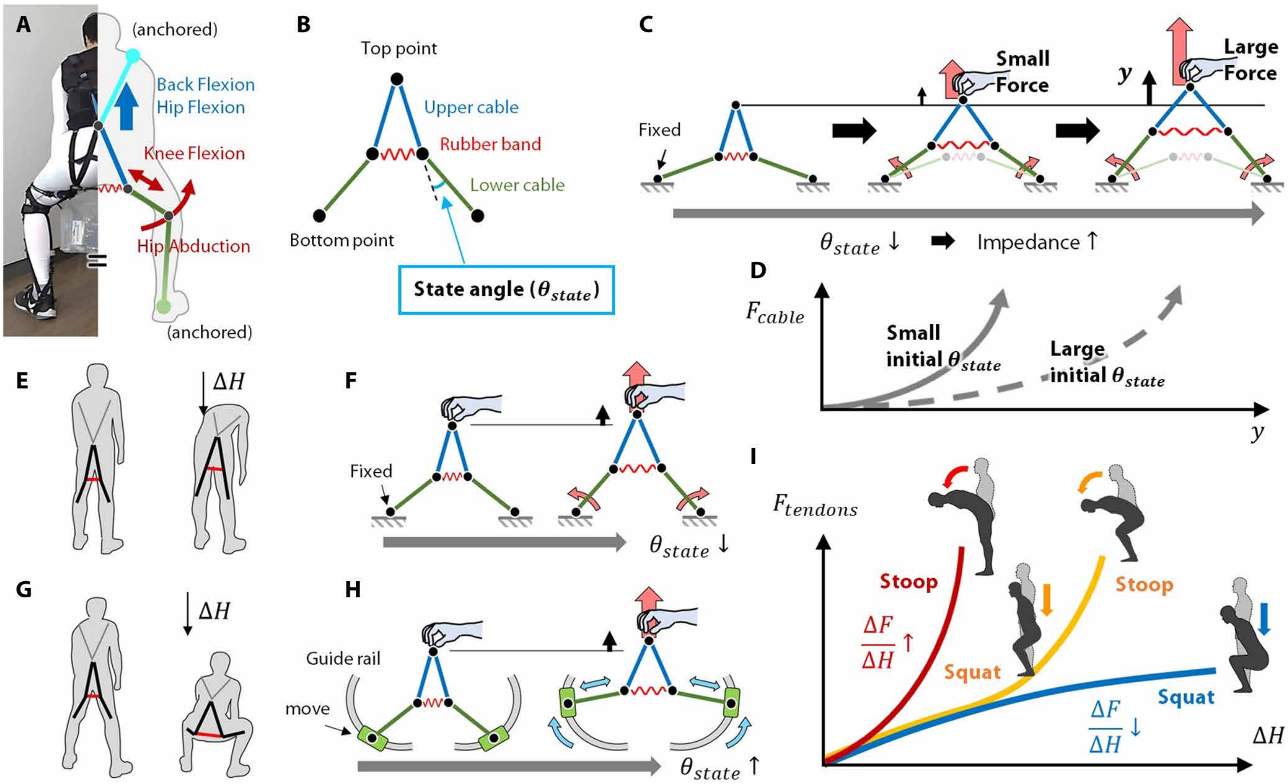

Fig. 2. Principle of body-powered variable impedance. (**A**) Arrangement of the tendons that create body-powered variable impedance in the exosuit. (**B**) A-shaped tendon structure. (**C**) Dependence of the tendon force on the state angle. The bottom points of the lower cables are fixed to the ground; the top of the upper cable is being manually pulled. (**D**) Tensile profile of the A-shaped tendon structure during deformation. (**E**) State angle during stooping. (**F**) Representative concept for explaining the deformation of the A-shaped tendon structure during stooping. (**G**) State angle during squatting. (**H**) Representative concept for explaining the deformation of the A-shaped tendon structure during squatting. (**I**) Dependence of the tendon force and impedance on lifting strategy. During stooping, tendon impedance increases because of the decreased state angle, which leads to a rapid increase in tension (red line). During squatting, tendon impedance decreases because of the increased state angle, which leads to a slow increase in tension (blue line). When the posture begins to change from squatting to stooping, the impedance pattern changes immediately to one that interferes with stooping (yellow line).

The key to the variable impedance of our design is allowing the tendons to slide laterally and change their angle. We achieved this by placing an orthogonal rubber band at the midpoint of two strands of nonparallel, bi-articular tendons connecting the hip and knees to create an A-shaped tendon structure (Fig. 2, A and B). When there is an angle difference between the upper cable and the lower cable (state angle), the cables tend to align by stretching the rubber band when the tendon is tight (Fig. 2C). When the state angle is large, the rubber band can be stretched easily under even small tension, which means the impedance of the bi-articular tendons becomes small (Fig. 2D). In contrast, when the state angle is small, the impedance of the bi-articular tendons becomes large (movie S5).

To minimize stooping, which involves flexion of the back and hip joints (Fig. 2E), we anchored the upper cables to the shoulders and routed their paths vertically down the back and hip joints. When the wearer stoops, the upper cables are pulled upward as the back and hip joints flex, decreasing the state angle (Fig. 2F). Because squatting involves flexion of the knee joints and abduction of the hip joints (Fig. 2G), we anchored the lower cables to the feet and routed them vertically upward behind the knees. When the wearer squats, hip abduction moves the lower cables in a lateral direction and knee flexion elongates them, increasing the state angle (Fig. 2H). As a result, high impedance is created when the wearer stoops to lift and low impedance is created when the wearer squats to lift (Fig. 2I and movie S2). Therefore the force field, designed through body-powered variable impedance, gives freedom of movement within the squatting posture range but hinders movement the farther the wearer's posture strays from squatting. This approach allows wearers to voluntarily select any motion pattern that is within the squatting posture range. Strategies for designing different types of trajectory-oriented force fields with body-powered variable impedance are introduced in the Supplementary Materials (figs. S3 and S4).

The design of the exosuit also allows for kneeling so that the wearer can lift large objects with relatively smaller BCF than when taking a squat (fig. S13). The left and right tendons of the suit are connected to each other through a ring-shaped pulley on the centerline of the body. This design makes the tendon slide horizontally without yielding noticeable resistance when the wearer's legs cross each other. This allows wearers to comfortably perform not only kneeling but also many other movements, including walking and climbing stairs (movie S1).

RESULTS

We modeled the exosuit and examined the force and energy profile produced by the tendons in each posture of the wearer (Supplementary Materials). We also compared the results with actual measurements.

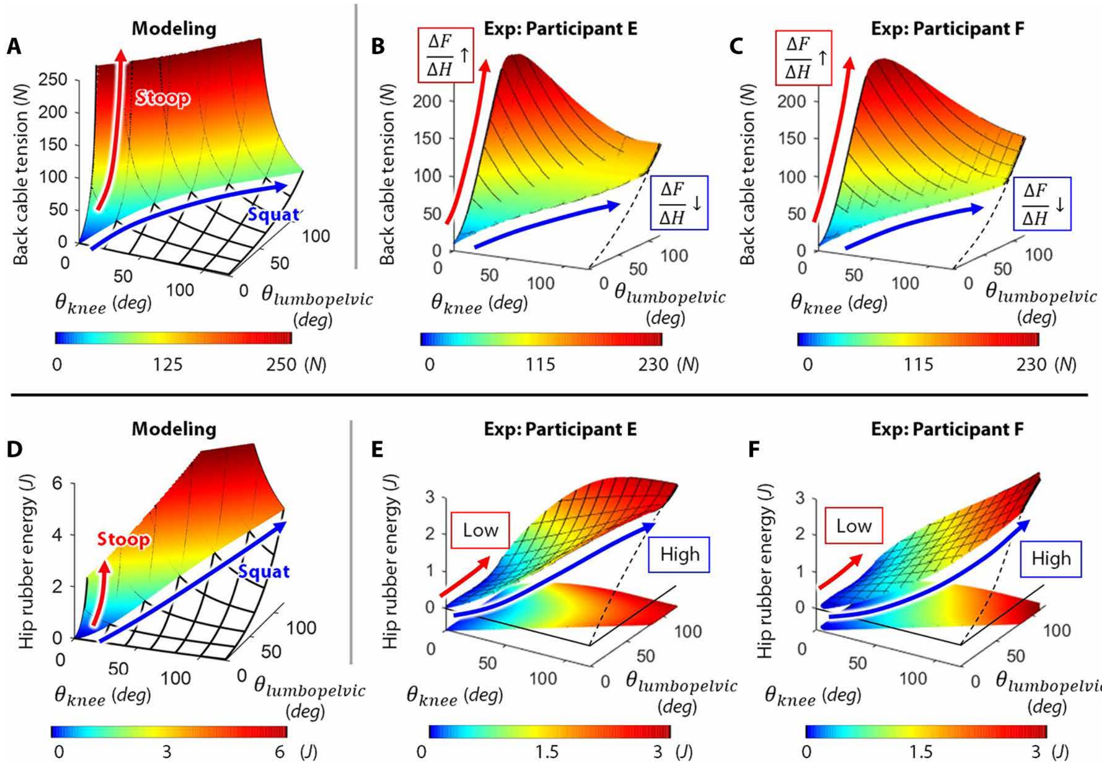

Fig. 3. Force and energy of the cable depending on posture. (Movie S2) (**A**) Force on the back cables estimated by the model. See Fig. 4A for the location of the lumbopelvic angle. (**B** and **C**) Force on the back cables measured by the experiments. (**D**) Energy stored in the hip rubber band estimated by the model. This plot is based on the assumption that no participants made a motion in which the back extension force exceeded 200 N, a supposition that the experiment confirmed. (**E** and **F**) Energy stored in the hip rubber band measured in the experiment.

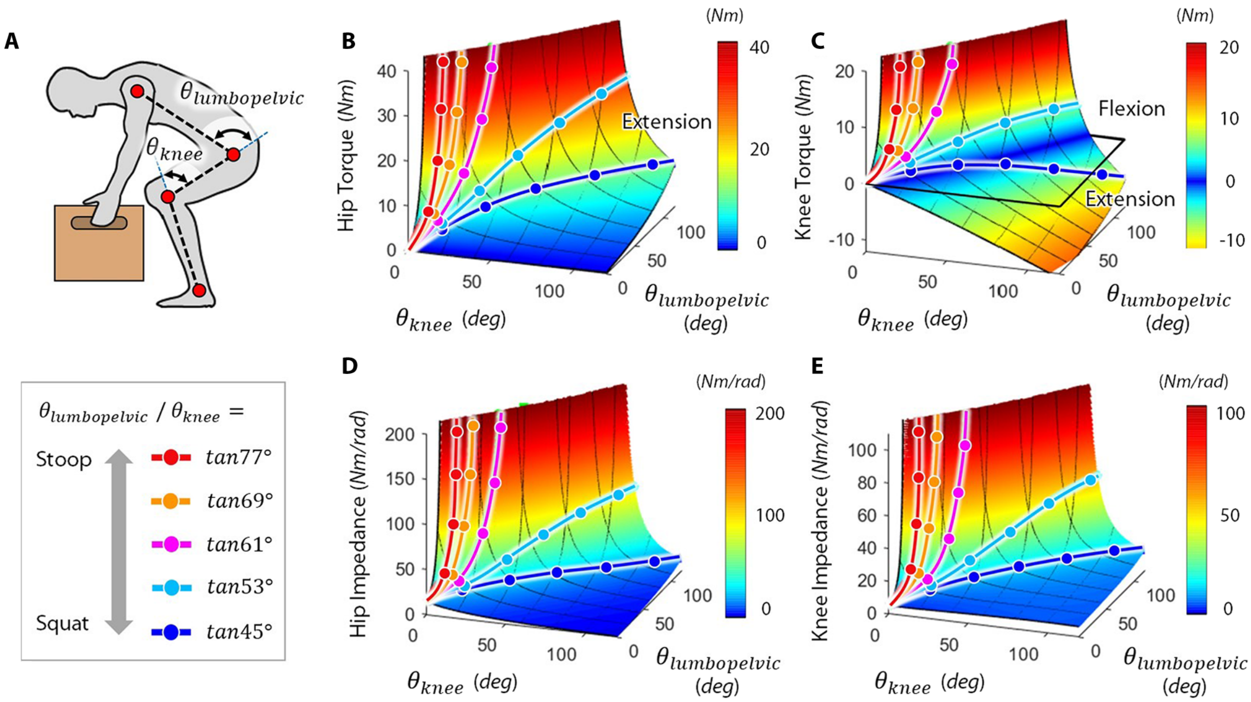

Fig. 4. Model of the torque and impedance created by body-powered variable impedance. (**A**) Angle parameters used for analysis. (**B**) Additive extension torque to the hip joint. (**C**) Additive flexion torque to the knee joint. The rubber band that traverses the kneecap (see fig. S2) changes the direction of the knee torque to extension in the squatting motion. (**D**) Additive impedance to the hip joint, resisting hip flexion. (**E**) Additive impedance to the knee joint, resisting knee extension.

The tension of the bi-articular tendon gradually increased as the lifting motion approached stooping and gradually decreased as it approached squatting in both the model and the measurements (Fig. 3, A to C, and movie S2). On the other hand, the potential energy stored in the hip rubber band is relatively large in the squatting posture range compared with the stooping posture range (Fig. 3, D to F, and movie S2). This result shows that the functionality of the exosuit can vary according to the wearer's movement pattern; it works as a spring during squatting and as a brake (or strut) during stooping (*54*). The exosuit model also shows the torque and impedance applied to the joints during various postures. The torque applied to the hip and knee joints increased as the lifting posture deviated from the squatting trajectory, which hindered stooping (Fig. 4, A to C). The additive joint impedance to the hip and knee joint also increased as the lifting posture approached the stooping trajectory (Fig. 4, D and E).

We performed experiments to assess the feasibility of the proposed suit design. Ten first-time users were recruited for a squat-lifting test and a free-lifting test. Each test was performed in with-suit and no-suit conditions. In the squat-lifting test, we measured the effect of wearing the suit on metabolic rate during squatting. The metabolic rate of 9 of 10 participants decreased, by an average of 5.26 ± 2.4% (n = 10; mean ± SEM; paired t test, P = 0.073; Fig. 5). Although the reduction in the metabolic rate was not statistically significant, the results show that the exosuit does not impose an energy penalty for squatting. On the contrary, the suit slightly decreased metabolic rate during squatting despite the suit's weight (850 g). The mechanical design, which enables the hip and knee rubber bands to absorb elastic energy only during the lowering motion of squatting, seems to contribute to this positive effect (Fig. 3, D to F).

In the free-lifting test, we measured the effect of the suit on the participant's voluntary lifting patterns. Lifting motion patterns were compared using the lifting postural index (LPI) (*55*)

$$\mathrm{LPI} = \frac{\Delta\mathrm{knee}}{\Delta\mathrm{ankle} + \Delta\mathrm{hip} + \Delta\mathrm{lumbar\ vertebra}} \quad (1)$$

where Δankle is the ankle plantar flexion angle, Δhip is the hip flexion angle, Δknee is the knee flexion angle, and Δlumbar vertebra is the flexion angle of the lumbar vertebra from the normal standing position. The lower the LPI, the closer the person's posture is to the stooping posture. A fully stooped posture and a fully squatting posture typically correspond to an LPI of 0.11 and 0.8, respectively. The suit increased LPI by 34.9% with statistical significance (N = 10; paired t test, P = 0.033, SMD = 0.79, 1 − β = 0.61; Fig. 6B). This increase in LPI was observed for 9 of 10 participants.

We also analyzed the range of motion of four major joints that are closely related to the lifting posture: trunk incline, spine flexion, lumbopelvic flexion, and knee flexion angle (Fig. 6, C to F; figs. S7 to S9; and tables S2 to S5). The results show that the participants changed their kinetic chain posture into a form that made the additive joint impedance by the exosuit smaller.

We additionally assessed the efficacy of the suit on the spinal loads with inverse analysis. We estimated the BCF on the L5-S1 spine of participants during the free-lifting test using musculoskeletal simulation software (OpenSim; SimTK; fig. S10). The peak BCF was decreased by 2.2 ± 0.94% with statistical significance (N = 10; mean ± SEM; paired t test, P = 0.046, SMD = 0.74, 1 − β = 0.55; Fig. 7B). This decrease in BCF was observed for 8 of 10 participants.

By additional analysis, we found that the suit force itself does not make a statistically significant effect on BCF, and the change in the voluntary lifting pattern of the participants is the major factor of the reduction in BCF (Supplementary Materials; fig. S12).

To sum up, the suit successfully changed wearers' free-lifting posture closer to the squatting form immediately, and most of the wearers were able to perform squat lifting with less energy consumption and reduced BCF.

DISCUSSION

Humans have, for many centuries, sought to artificially create a better body by donning garments and harnesses that passively

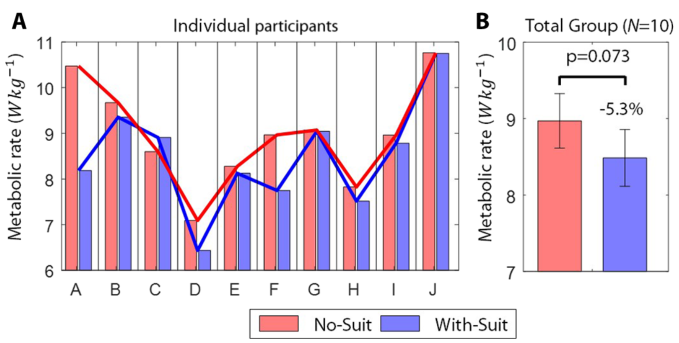

Fig. 5. Metabolic rate measured during the squat-lifting test. (**A**) Metabolic rate for each participant. Each letter on the x axis of the graph corresponds to each individual participant. (**B**) Average metabolic rate for the total participant group (N = 10; mean ± SEM).

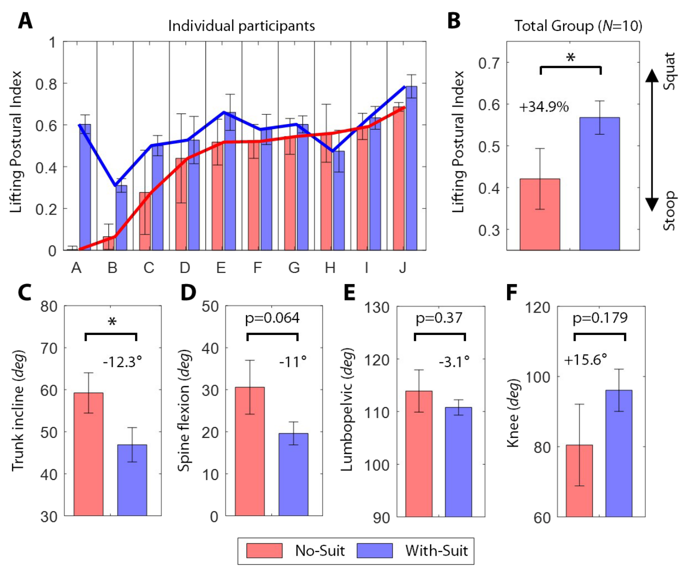

Fig. 6. LPI measured during the free-lifting test. (**A**) LPI for each participant (mean ± SD). Each letter on the x axis of the graph corresponds to each individual participant. Results are ordered from left to right on the basis of participants' LPI in the no-suit condition. (**B**) Average LPI for the total participant group (N = 10; mean ± SEM). (**C**) Average maximum trunk incline angle for the total participant group (N = 10; mean ± SEM). (**D**) Average maximum spine flexion angle for the total participant group (N = 10; mean ± SEM). (**E**) Average maximum lumbopelvic flexion angle for the total participant group (N = 10; mean ± SEM). (**F**) Average maximum knee flexion angle for the total participant group (N = 10; mean ± SEM). *P < 0.05.

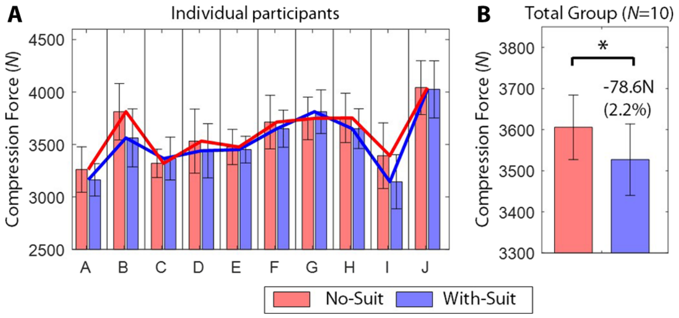

Fig. 7. BCF on L5-S1 estimated using OpenSim in the free-lifting test. (**A**) Peak BCF for each participant (mean ± SD). Each letter on the *x* axis of the graph corresponds to each individual participant. (**B**) Average BCF for the total participant group ($N = 10$; mean ± SEM). $^*P < 0.05$.

assist movement. Today, this involves developing state-of-the-art wearable robots, but simple passive devices still remain relatively popular (*56*, *57*). In this study, we developed a body-powered variable impedance technique that can be embedded in a garment. The proposed mechanism does not forcibly determine the wearer's body position. Instead, it changes the force field to make undesirable motions difficult and desirable motions easy, thereby inducing the wearer to voluntarily choose desirable ones.

The exosuit can be used as an assistive device to train people to lift small-sized but heavy objects by squatting. The 10 participants had no experience of lifting objects while wearing the suit before the experiment; the efficacy of the suit, which was evaluated by the experimental results, suggest that the suit induces fast motor adaptation of the wearers, resulting in the immediate reshaping of the lifting posture. Furthermore, 5 of the 10 participants did not have any weight training experience (table S1); the function of the suit is valid even for novices who lack lifting training. This immediate efficacy of the exosuit in reshaping the voluntary lifting posture may be useful in various sites, including some industrial settings (*12*), gyms (*13*), and rehabilitation centers (*14*), where the squat technique is recommended, particularly for untrained employees.

One limitation of the exosuit design is that it constrains forward-leaning motion and possibly interferes with movements not related to lifting objects, such as leaning forward to grasp objects. This may make the exosuit uncomfortable to wear for long periods of time. To alleviate this problem, we designed a mode control module that lets people electronically or manually adjust the exosuit's range of motion of the spine and hip flexion to permit freer movement (fig. S14 and movie S3).

Impedance is one of the factors that influence human behavior, but its efficacy may differ from person to person without elaborated customization. In our experiment, each of the 10 participants showed a different degree of postural change, and one outlier changed his posture closer to stooping after wearing the suit. As a result, the statistical power decreased below 0.8. Therefore, the statistically significant change in LPI and BCF should only be accepted as a demonstration of the feasibility. As a future work, further experiments with customized exosuit parameters and the resulting impedance will contribute to establishing a more effective strategy for controlling human behavior through the body-powered variable impedance.

Although the estimated BCF reduction was not substantial, we speculate that the actual effect of the suit could be more than the calculated effect if the anatomy of the spine is taken into consideration. When the spine is straight, the facet joints share about 20% of the spinal load, reducing the load on the discs (*58*, *59*). Therefore, for participants who decreased spine flexion after wearing the suit, the reduction in the actual stress transmitted to the disc is likely more than the reduction in the peak BCF.

There were several additional limitations of the evaluation of the device performance. First, the kind and amount of the information that can be collected during the lifting motion in the experiment were limited. Because the cables in the exosuit closely contacts with the body and slides on the wearer's body surface, a conventional load cell attached to the suit would cause substantial discomfort when the wearer performs lifting. Likewise, a surface electromyography sensor attached to the wearer's muscle would cause interference with the exosuit. Any design change to circumvent this discomfort or interference would degrade the performance or the compactness of the suit. Second, the effect of muscle co-contraction or detailed anatomy of the spine was not considered in the simulation. Furthermore, to the best of our knowledge, no currently available tool for inverse analysis provides a reliable way to assess the effect of soft wearable devices that continuously change the cable path like the exosuit used in this study. In future work, it is necessary to evaluate the effect of the exosuit more accurately when more advanced sensors or analysis tools become available.

Unfortunately, humans often choose to perform motor tasks in a way that is comfortable in the short term but might be harmful in the long term. Because the deleterious effects of undesired motion, such as spinal disc degeneration, develop slowly, people do not receive stimuli from them in time to adopt less harmful motor patterns (*60*–*62*). On the other hand, people immediately change their movements to reduce unpleasant stimuli such as mechanical or psychological stress, pain, fatigue, or discomfort. Body-powered variable impedance can provide the missing immediate feedback about undesired motions by generating the stimulus of impedance change in response to the undesired motion, resulting in immediate sensation of mechanical resistance and discomfort. By artificially giving a high priority to any appropriate posture, this process can induce the wearer to voluntarily choose the desired movement pattern. The proposed approach will facilitate the development of wearable devices that can alter motor patterns via a deliberately designed force field without a power source.

MATERIALS AND METHODS

Participants

Ten healthy male adults ($N = 10$; age, 25.4 ± 2.4 years; weight, 74 ± 7 kg; height, 175.9 ± 2.4 cm; mean ± SD; table S1) participated in the experiment. All participants were people who had no experience of wearing the exosuit (participants A, C, D, E, F, G, I, and J) or who had worn the suit for less than 5 min before the experiment (participants B and H). Participants B and H had simply moved their limbs with the suit to check the suit size but had not performed any actions related to the experimental protocol before the experiment. One additional participant participated in the experiment but was excluded from the analysis because the suit did not fit his body. All experimental procedures were approved by the Institutional Review Board of Seoul National University. Participants were informed about the experiment and provided written consent before participation.

Experimental design

Each participant was given 5 min of warm-up time to get used to the suit and try out lifting with it. All participants conducted a squat-lifting test and then a free-lifting test (fig. S5A). In each test, the order of the no-suit and with-suit conditions was randomly assigned for each participant. In each test, participants repeatedly lifted and lowered a 10-kg box with handles at a constant rate of 10 lifts/min for 6 min, from a 200-mm-high shelf to standing height. The size of the box was 450 mm by 290 mm by 150 mm. Each test condition was separated by a 20-min break. A force plate (FP6090-15-2000, Bertec, Columbus, OH, USA) recorded the ground reaction force (GRF) during the entire experiment. We developed a timer program that provided visual and auditory stimuli to help participants lift the box at a constant speed of 10 lifts/min. Each cycle of the lifting task was 6 s long: 2 s each for lifting and lowering and 1 s of waiting time in the fully lowered and standing postures (fig. S5B). In the squat-lifting test, participants were instructed to lift the box 60 times with a squatting movement. No detailed instruction on how to squat was provided. Instead, participants were instructed to follow the guidelines in the NIOSH manual (*11*). Metabolic gas data were collected with a portable respiratory gas analyzer (K5, COSMED, Italy). In the free-lifting test, participants were instructed to lift the box 60 times using any movements that they liked except for asymmetric motions, such as twisting and lateral bending, which could severely hinder evaluation of the LPI. Joint positions were recorded with motion-tracking cameras (Oqus500, Qualisys, Sweden; fig. S6). Additional tests were performed on two participants to measure tendon force from back cable and hip rubber band in various lifting postures with self-selected speed. Tension on the cables was measured by load cells (333FDX, Ktoyo, South Korea).

One limitation is that we did not randomize the order of the squat-lifting test and the free-lifting test; we intended to minimize any possible effect of fatigue during the metabolic cost measurement performed in the squat-lifting test. In future work, the efficacy of the exosuit can be evaluated more systematically either by assigning multiple sets of tests to participants in random order or by assigning only a single set to a single participant after recruiting sufficient participants.

Lifting posture index analysis

Adopting the concept of LPI from a previous study (*55*), we quantitatively assessed how close the participant's posture was to a squatting or stooped posture. The representative posture index per lift cycle was specified as the LPI at the point when trunk inclination reached its maximum. All 60 LPI readings collected over the entire 6-min lifting test were used for the analysis.

Estimation of spinal load using OpenSim

The OpenSim (SimTK, Stanford, CA; www.simtk.org) simulation platform was used for estimating the compression force, shear force, and moment on the L5-S1 spine. A full-body lumbar spine (FBLS) model was used for the simulation (fig. S10) (*63*). We increased the range of motion of the hips, lumbar, knees, elbows, and arms of the FBLS model following a recommendation of previous research (*64*). Marker position data (fig. S6), GRF data, and a 10-kg vertical load on the hands were specified in OpenSim. Because our experiment was performed on a single large force plate, we assumed that the measured GRF was symmetrically applied to both feet. To simulate holding a 10-kg box in the hands, we set the condition that an external load of 5 kg would be applied to each hand. We followed the simulation procedure used in previous studies (*64*, *65*). Scale model, inverse kinematics, inverse dynamics, static optimization, and joint reaction analyses were performed sequentially. A second-order Butterworth low-pass filter at a cut-off frequency of 6 Hz was applied during the static optimization. Muscle forces were calculated using the sum of the square of 324 musculotendon actuator (including 224 trunk muscle fascicles) forces as the cost function. The peak compression force, shear force, and moment for each cycle were selected as representative results (Fig. 7, fig. S11, and tables S7 to S9). All 60 spinal load readings collected over the 6-min lifting test were used for the analysis.

Metabolic rate analysis

A portable respiratory gas analyzer was used to measure metabolic rate (K5, COSMED, Italy) during the 6-min squat-lifting cycles. Average oxygen consumption and carbon dioxide production data over the last minute of the 6-min squat-lifting cycles were used for analysis. Energy expenditure was calculated using software supported by COSMED (*66*), referring to existing methods (*67*).

Visualization of force and energy data

A polynomial surface fitting (poly23) was performed using MATLAB (MathWorks, Natick, MA, USA) to express the force and energy data obtained in the experiment in the form of a continuous curved surface.

Statistics

The data were statistically analyzed in MATLAB (MathWorks, MA, USA). We used a paired t test to compare data from the no-suit and with-suit conditions. The level of statistical significance was set at $P < 0.05$. Statistical powers for LPI and BCF results were analyzed using G*Power software (F. Faul, Christian-Albrechts-Universität Kiel, Kiel, Germany) (*68*).

SUPPLEMENTARY MATERIALS

robotics.sciencemag.org/cgi/content/full/6/57/eabe1243/DC1
Materials and Methods
Figs. S1 to S17
Tables S1 to S11
Movies S1 to S5

REFERENCES AND NOTES

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