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Encouragement of Squat-Lifting: Feasibility Study of a Highly Usable Passive Knee Assistive Device

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Abstract— **Low back pain (LBP) is a serious occupational disease. A main contributing factor to LBP is a burden on the lumbar region, which can occur when lifting heavy objects. To reduce the risk of LBP, squat-lifting, which uses knee torque, is recommended as a preferred alternative to stoop-lifting, which uses hip and back muscles. However, people tend to avoid squat-lifting due to its relatively low metabolic efficiency, and the greater force it places on the knee and rectus femoris. Many lifting assistive devices have already been commercialized to cope with LBP. However, the majority are designed to support back or hip muscles, not to encourage workers to perform squat-lifting. Therefore, we fabricated a prototype knee assistive device that is easy-to-wear and light-weight, using shape-memory-alloy (SMA) wires that are inherently passive elements. We then tested the effectiveness of the device by measuring electromyography (EMG) levels in the rectus femoris during squat. At knee bending positions, the device reduced EMG levels in the rectus femoris by at least 15% on average.**

I. INTRODUCTION

Multiple original studies [1], [2], [3] and systematic reviews [4], [5] agree that manual lifting is a major cause of developing low back pain (LBP). When lifting heavy objects, two predominant methods are stoop-lifting and squat-lifting, as shown in Fig. 1. Squat-lifting relies on knee torque for lifting, while keeping the torques from the back low. Stoop-lifting is characterized by an inclined trunk and almost fully extended knees. In one definition, the postural index, which is the ratio between knee flexion from normal standing and the sum of ankle, hip, and lumbar vertebral flexions from normal standing, is approximately 0.80 for squat-lifting and 0.11 for stoop-lifting [6]. Squat-lifting reduces the maximum voluntary contraction in the erector spinae compared to stoop-lifting [7], and the shear forces applied to lumbar for stoop-lifting are estimated to be 180% greater than squat-lifting [8]. Furthermore, stoop-lifting is reported to have caused approximately 75% more stresses on discs and ligaments than squat-lifting [9]. Consequently, the governmental ministry in Japan recommends squat-lifting to lower the risks of LBP [10]. However, squat-lifting is 23% to 34% less metabolically (in terms of work per oxygen consumption) efficient than stoop-lifting [11]. The maximum oxygen consumption during squat-lifting is 14.3% more than that during stoop-lifting [12], and the maximum heart rate during squat-lifting is 6.5% more than that during stoop-lifting [12]. Furthermore, squat-lifting increases maximum muscle contraction in the rectus femoris [7]. As a result, although squat-lifting can be effective in preventing LBP because it exerts less shear stress on lumbar discs, it is more fatiguing and thus, less frequently used.

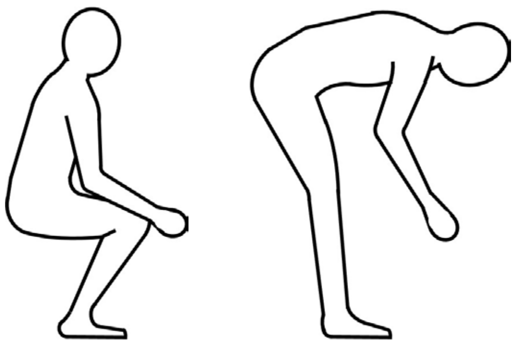

Fig. 1: Schematic diagrams of squat-lifting (left) and stoop-lifting (right).

To prevent or manage LBP, several wearable assistive devices have been developed and commercialized in recent years. The majority of these devices provide external support to the lumbar spinae when lifting objects. Depending on how external forces are applied, there are two types of assistive devices: active and passive. Active devices use actuators such as electromagnetic, electrohydraulic, and electropneumatic to generate the external forces; passive devices use intrinsically passive elements to store energy during negative works, and then release the stored energy during positive works.

Hybrid Assistive Limb (Cyberdyne, Ibaraki, Japan) [13], [14] and Mk2b [15], [16], a revised version of Robo-Mate [17], are examples of active assistive devices with two electromagnetic motors on the hip joints of the wearer. Both devices provide assistive torque to wearers' hip and trunk to support the lifting motion. In contrast, the Laevo (Laveo,The Netherlands) [18] is a passive assistive device that uses pads contacting wearers' chest, back, and upper legs, and a circular tube with spring-like characteristics that connects these pads to generate upward forces to the chest; the reactive forces are supported by the wearers' legs. A garment-like device [19], designed by Vanderbilt University, USA, is

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another passive device that uses elastic bands to provide extension momentum to the lumbar. Additional devices have been either commercialized or in development.

These assistive devices have trade-offs between effectiveness and usability. The active ones produce more effective support, but some require external energy sources and are heavier. The wearability is not high because these devices use rigid links and cuffs and allow limited degrees of freedom during use. The passive devices, on the contrary, are lightweight and are more practical to use when wearers need to move around frequently and in circumstances where external power sources are not available.

Assistive devices can also be categorized by supported area. Assistive devices for lifting are focused on protecting the lower back and hips. Other assistive devices provide support to the knees or ankles and can be utilized for walking assistance [20], [21], [22], [23], [24]. However, giving support to the lower back and hip joints does not change the wearers' behavior from stoop-lifting to squat-lifting. Consequently, we focused on a knee assistive device to encourage laborers to conduct squat-lifting to lower the risk of LBP.

To expand the use of squat-lifting, this study aims to develop a knee assistive device with high usability by using shape-memory-alloy (SMA) wires as passive elements. This device takes advantage of the superelastic property of SMA to provide external forces to the rectus femoris, encouraging workers to perform squat-lifting by compensating for the part of the necessary knee torque. In this paper, we describe the designing process and policies of the device and evaluate the effectiveness of the device experimentally by measuring the electromyography (EMG) of the rectus femoris.

II. KNEE ASSISTIVE DEVICE WITH SHAPE-MEMORY-ALLOY WIRES

The knee assistive device consists of two main parts: a cloth knee brace and SMA wires. The knee brace functions as a splint on which to install the SMA wires, while the SMA wires provide the external forces to support rectus femoris. The ends and the middle part of the wires, which are the vertices during bending, are fixed to the knee brace. This design positions the bending center of each wire at the rotational center of the knee on the sagittal plane. The device is shown in Fig. 2.

A. Knee Brace

A commercial hinged knee brace (54557, Mueller Sports Medicine Company, Germany) was chosen to be part of the assistive device. The hinges of the knee brace, which are used to prevent the medial-lateral motion of the knee to protect damaged knees, were removed. Therefore, the brace was used as a soft chassis on which to attach the SMA wires to provide supportive force while bending.

The brace is designed to be easy-to-wear by using two Velcro strips, which allow the wearer to put on and off the device without needing to remove his or her shoes.

Fig. 2: Knee assistive device built from a commercial knee supporter with shape-memory-alloy wires.

B. Shape-Memory-Alloy Wires

The SMA wires are made of nickel-titanium (Ni-Ti) alloy (Yoshimi Inc., Japan). Supportive forces are generated by the superelasticity property of SMA, which is more accurately described as pseudoelasticity. As shown in Fig. 3, four wires were joined together with a thermoplastic resin. Each strand of wire was put in a pocket at each side of the knee brace, thus there were a total of eight SMA wires per knee. The maximum knee torque for sit-to-stand movement is 0.38 Nm/kg for normal people [25]. Thus, for a 70-kg person, it requires 26.6 Nm to perform sit-to-stand movements. The number, length, and diameter of the SMA wires were selected to produce 10% of the required torque. This value (10%) was chosen so that the device would reduce the intensity of muscle activity required during the sit-to-stand motion without the wearer feeling a resistive force during the other activities, which would be perceived as a nuisance. The choice was subjective because, to the best of our knowledge, such a minimum resistive force that does not subjectively or physically disturb the motion of wearers has yet to be determined. Therefore, no current literature or data is available to determine an optimal percentage for the level of support provided by passive knee assistive devices. Furthermore, our device was designed to generate 10% of the necessary torques when the wearer is not lifting any additional weight. It should be noted that the support ratio of the device will be smaller than 10% while lifting additional weight. The performance tests we used to determine the values of the length, diameter, and the number of wires will be described in the next subsection.

C. Determination of the Length and Diameter of SMA Wires

We conducted experiments to determine the optimal length and diameter of SMA wires in terms of the forces generated by the wires. A push-pull scale (FB-100N, Imada Co. Ltd., Japan) was used to measure the restoring force when the wire was bent at set angles. One end of the SMA wire was fixed to the push-pull scale, and an external force was applied to the other end to bend the wire to a certain angle. The push-pull scale was aligned with the direction of the restoring force of the wire. The maximum knee angle of a sit-to-stand motion

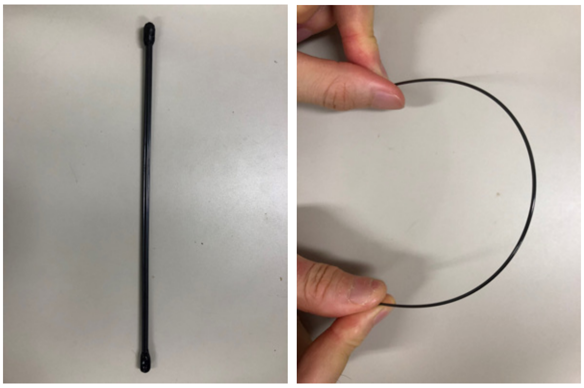


Fig. 3: The left panel is a single strand of the SMA wire. A thermoplastic resin was applied on the two ends of four wires to bind them together. The right panel shows a SMA wire under bending conditions.

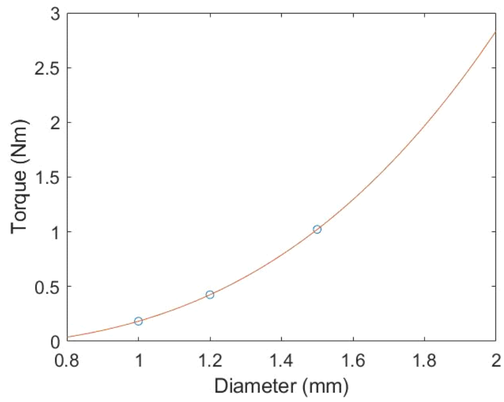


Fig. 4: The relationship between diameter and torque of SMA wires. All wires were 20 cm in length, and four wires of identical diameter were joined together. Torques were determined with the wires bent at 100°.

is approximately 100° [26]. Therefore, the resultant torque was measured at 100°.

To experimentally evaluate the diameter, SMA wires having diameters of 1.0, 1.2, and 1.5 mm, and a common length of 20 cm were compared. Four wires of the same diameter were joined together for the measurements, and the results are shown in Fig. 4. SMA wires of larger diameter generated greater torques during bending. The relationship between diameter and torque was nonlinear and quadratic.

For the length test, the diameter of the SMA wires was fixed to 1.0 mm, and the length ranged from 16 cm to 34 cm with an interval of 2 cm. Different from the diameter test, we only used one wire to measure the forces for this test. As shown in Fig. 5, the shorter SMA wires exhibited higher torque, whereas the torque is not significantly different when the wire length is more than 20 cm. Nonetheless, when shorter wires are used, the clothing area that supports the wires is also smaller, and the restoring force damages the cloth material of the knee brace.

Based on these performance tests, we determined that the optimal length, diameter, and number of wires is 22 cm, 1.8 mm, and 8, respectively. To make these determinations, inter- and extrapolation of the data were used. This combination achieves the aforementioned supportive torque, i.e. 10% of the moment.

III. EFFECTIVENESS TESTS

A. Participants

Three participants (one male, two females in 20s) agreed to test the EMG value while squatting. The average height and weight of the participants were 166.7±9.1 cm and 55.3± 5.9 kg, respectively. All participants were healthy with no injuries or diseases.

B. Experiment Procedures

We measured the EMG signals of the rectus femoris under four different conditions: 45° (half) or 90° (full)-squat with or without the knee assistive device. For each trial, the participants were asked to squat to the assigned angle, then

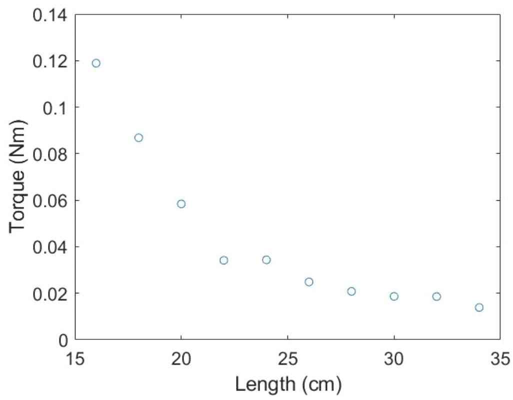


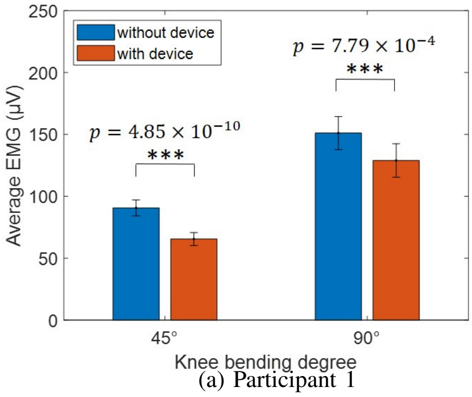
Fig. 5: The relationship between length and torque of SMA wires. The diameter of each wire was 1.0 mm. Torques were determined with the wires bent at 100°.

pause at the position for five seconds. A constant height chair was used as a mark to let the participant know whether his or her knee angle was close to the designated value. This task mimicked a situation in which the participant was expected to hold and lift an object by squat-lifting. The half squat is sometimes used in earlier studies (e.g. [27]).

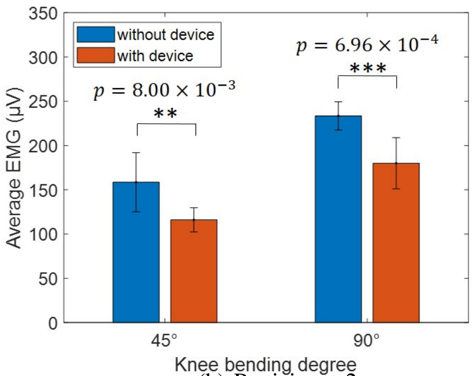
To avoid accidental injuries, the participants did not have any actual loads in their hands and were instructed to avoid quick movement. Each condition was repeated a total of 10 times by each participant. To prevent the muscle fatigue from affecting the results, the four conditions were tested in randomized order with each condition repeated five times in succession.

C. Data Analysis

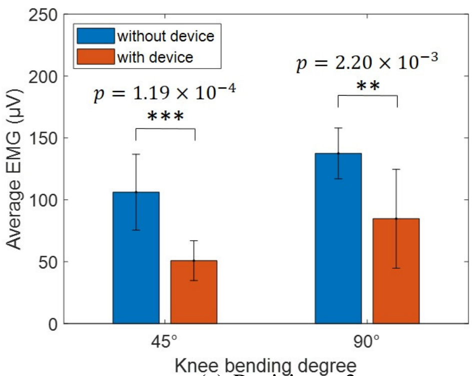
To process the EMG signals, we focused on the central three seconds within the five-seconds-pause. During the central three seconds, relatively stable signals were recorded. We calculated the mean voltage of the absolute envelope of each trial, and then compared the difference between the trials done with and without the knee assistive device.



(a) Participant 1

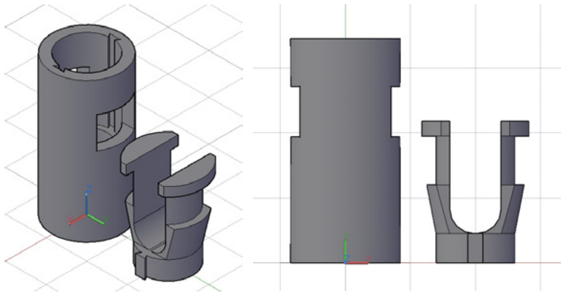


(b) Participant 2



(c) Participant 3

Fig. 6: EMG values of the rectus femoris. **: $p<0.01$, ***: $p<0.001$.


(a) CAD images of the fixation mechanism.

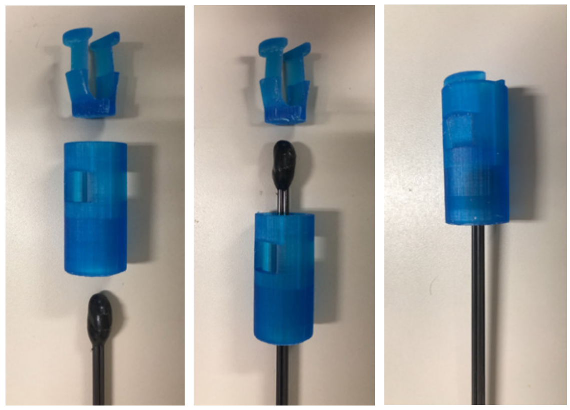

(b) Manufactured by a three-dimensional printer.

Fig. 7: Fixation mechanism. (a) CAD images. (b) From left to right, the panels illustrate how to confine the SMA wires in the tube.

Although we planned to exclude samples where the variation exceeded two standard deviations, there were no outliers among the samples. However, we did exclude some trials, in which the EMG signals were intermittent because of false skin-electrode contact as invalid.

D. Results

The mean EMG signals and the p-values of Welch's t-test of the three participants are shown in Fig. 6. There are significant differences between the conditions with and without the knee assistive device for both angles. For each of the three participants, the mean EMG signal decreased by 27.7%, 26.8%, and 52.1% when squatting 45°; and decreased by 14.7%, 22.9%, and 38.4% when squatting 90°. The effectiveness of the knee assistive device varies for each participant, which may be because the weight of each participant is different.

IV. DESIGNING THE SMA WIRE FIXATION MECHANISM FOR OPTIMAL USABILITY

To increase usability, a mechanism for easily installing and uninstalling the SMA wires is proposed in Fig. 7. The mechanism consists of two parts. One part looks like a cylinder with two square holes, which is used to confine the horizontal movement of the wires. The other part is a lid with two "ears" to confine the vertical movement of the wires. There are two convex squares on the ears that fit into the two square holes in the cylinder. The lid can be easily removed by pressing the ears. A detailed procedure of utilizing the mechanism is shown in Fig. 7b. With this mechanism, users can turn on and off the supportive torque without taking off the knee assistive device. The bottom end of the SMA strand is inserted into the pocket of the knee brace while the top is anchored by the mechanism. This mechanism has no effect on the assistive performance; however, it contributes to the usability of the assistive device.

V. CONCLUSIONS

Many published reports and government agencies recommend using squat-lifting instead of stoop-lifting to reduce the risk of low back pain. To encourage people to perform squat-lifting, we made a prototype of a light-weight and easy-to-wear passive knee assistive device. This device consists of a commercial knee supporter and two strands of SMA wires. One device is worn on each knee. The SMA wires do not lose their elasticity even when they are bent at larger angles and generate restoring forces. A fixation mechanism for the SMA wires was devised to easily turn the assistive device on and off without taking off the knee brace. The device was designed to support approximately 10% of the maximum torques for healthy people weighting 70 kg to perform sit-to-stand movements. Four different conditions were compared to experimentally evaluate the effectiveness of the prototype: squats at 45° and 90° with and without the assistive device. The EMG values of rectus femoris were collected while three participants performed each condition in a randomized order. The results indicate that, with the knee assistive device, the participants decreased the activity of the rectus femoris muscle by 14.7%–52.1%. The participants were lighter than 70 kg, so the observed effects were higher than expected. The proposed knee assistive encourages people to change their lifting behavior from stoop-lifting to squat-lifting. Nonetheless, further usability tests are needed to evaluate the practical utility of the proposed device and include subjective feedback from occupational users. Furthermore, although we did not use any loads in this study, if the participants had lifted actual weights, then the magnitude of reduction of the EMG signals during use of the assistive device would have been lower.

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