Soft Exosuit Based on Fabric Muscle for Upper Limb Assistance

Seong Jun Park, Kyungjun Choi, Hugo Rodrigue, *Member, IEEE*, and Cheol Hoon Park

***Abstract*—Upper limb exosuits (ULEs) are wearable robots that can assist arms to lift or support an object. However, heavy wearable equipment often interferes with the free movement of a person's arms, which have a wide range of motions. This article presents the design and manufacturing processes for a ULE that does not restrict the range of motion of the person wearing it and that can be folded to a small, palm-sized volume for portability. The light and flexible ULE employs a fabric woven from shape memory alloy (SMA) springs (FWS) as an actuator. The FWS uses SMA springs coiled with microdiameter SMA wires, and it exhibits the soft and flexible characteristics of a fabric. The proposed lightweight ULE weighs 540 g and supports the muscular strength for the bending motion of an elbow. The experimental results indicated that the proposed ULE reduced the muscle activity of subjects by approximately 50–70% when they held a load of up to 10 kg under the condition that their temperature was controlled to be 50 °C or lower. This result suggests that the proposed ULE can be used to provide strength assistance for delivery or construction workers.**

***Index Terms*—Fabric muscle, shape memory alloy (SMA) spring, soft exosuit, upper limb assistance.**

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Seong Jun Park is with the Department of Robotics and Mechatronics, Korea Institute of Machinery and Materials, Daejeon 34103, South Korea, and also with the School of Mechanical Engineering, Sungkyunkwan University, Suwon-si 16419, South Korea (e-mail: sjpark61@kimm.re.kr).

Kyungjun Choi is with the Department of Robotics and Mechatronics, Korea Institute of Machinery and Materials, Daejeon 34103, South Korea, and also with the Department of Mechatronics Engineering, Chungnam National University, Daejeon 34134, South Korea (e-mail: choikj@kimm.re.kr).

Hugo Rodrigue is with the School of Mechanical Engineering, Sungkyunkwan University, Suwon-si 16419, South Korea (e-mail: rodrigue@skku.edu).

Cheol Hoon Park is with the Department of Robotics and Mechatronics, Korea Institute of Machinery and Materials, Daejeon 34103, South Korea (e-mail: parkch@kimm.re.kr).

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I. Introduction

WEARABLE robots can help assist or strengthen the motor abilities of a person in their daily lives as well in industrial and rehabilitation-related fields. People use wearable robots like clothing, and therefore, the wearability of these robots, which is defined as the comfort provided by an object or a device attached to the body of a person [1], needs to be guaranteed. To this end, several recent studies have focused on developing soft wearable robots or exosuits, which exhibit flexible and lightweight characteristics, to allow people to perform movements freely while wearing these robots [2]–[11]. Among the various types of exosuits that are being developed, upper limb exosuits (ULEs) support the arms performing a wide range of motions (ROMs) under changing working conditions. Thus, ULEs need to realize high wearability for ensuring free and comfortable movement to those wearing them, irrespective of whether they are supported by an assistive force. In addition, it should be easy to wear and remove ULEs, and they should allow the person to perform movements as freely as before wearing the ULE. The arm of a person is light, i.e., it accounts for only 4.3% of the body's weight [12]. Therefore, a heavy actuator or device attached to the arm can disturb the free movement of the arm in unassisted scenarios or work conditions that do not require a large assistive force. Thus, ULEs should use light, soft actuators, and components to minimize the increase in inertia and assist in the muscular strength of the arms. The ULEs also need to provide high portability; i.e., people should be allowed to wear these exosuits freely when they require assistance, regardless of the time and location. People can easily carry ULEs if they can be folded or rolled to a small size, like a shirt, which allows them to hold these robots in their hands.

Many exosuits that utilize a motor-tendon mechanism have been developed to enable the free movement of wearers while employing lightweight components for assistive parts of the arm [13]. When people wear these exosuits, light, and flexible cables, which transfer the power of the motors, are attached to their arms instead of heavy and hard motors. The use of such light and flexible cables helps prevent the exosuits from interfering with the movements of the wearer. However, the number of motors and cable routings needs to be increased in proportion to the degree of freedom of joints for supporting the upper limb. Such an increase can complicate the design of ULEs, increase the weight of these exosuits, and reduce their wearability and portability. Gaponov et al. [7] developed a ULE that did not include hard links or joints around the arms of the wearer; their ULE supported the shoulders and elbows of the wearer for rehabilitation by operating a direct current (dc) motor placed at the back of the wearer. The weight of the components attached to the arms of a wearer was 0.32 kg. Unlike these lightweight components, the weight of the motor and the frame in the back used to fix the motor is 2.8 kg, which accounts for over 70% of the weight of the entire system. Samper-Escudero et al. [9] introduced a ULE that supported one side of a shoulder and an elbow of a wearer in their daily activities or rehabilitation. This ULE used an external power supply, and the total weight of the backpack containing the two motors and their electronic components required for operating this ULE was 890 g. The weight and size of the backpack increase when using the battery as a power source, and this makes it less portable. Lessard et al. [10] presented a ULE that supported one-side shoulder and elbow joints of a stroke patient that wears this exosuit for rehabilitation. This ULE comprised six motors, a controller, and a battery; its total weight was 1.3 kg. These ULEs enhance physical functions related to the upper limb; however, these ULEs cannot be folded or rolled like clothes to a small, portable size because of the volume occupied by motors and rigid components.

Fabrics or textiles have recently gained interest as potential materials for developing exosuits designed as everyday clothes that do not disturb the movements of wearers based on their inherent advantages, such as flexibility, softness, and air permeability [13]. Accordingly, researchers have developed light and flexible fabric-type actuators, i.e., fabric muscles, to replace the existing motor-tendon mechanism.

Fabric muscles perform the linear actuation of contraction and relaxation similar to skeletal muscles. Like skeletal muscles are distributed over the skeleton, fabric muscles can also be distributed on body parts that require direct and uniform strength assistance. Therefore, the use of a fabric muscle facilitates the easy and simple design of exosuits. Existing fabric muscles are classified based on their operation method as those based on joule heating and those on pneumatic operation [14]–[16]. Fabric muscles based on pneumatic operation require additional devices such as compressors, regulators, and valves in addition to the controller and battery, and they generate a loud noise during operation. In contrast, fabric muscles based on joule heating comprise the controller and battery, and they do not generate noise during operation. However, joule heating leads to an increase in the temperature of the fabric muscles, and thus, these fabric muscles need to be operated within a certain range of temperatures that ensures the safety of wearers even when their bodies are directly in contact with fabric muscles when generating sufficient force for strength assistance.

This article proposed a fabric woven from shape memory alloy (SMA) springs (FWS) based on joule heating. In addition, we developed a lightweight ULE (540 g, including the controller and the battery) to support the muscular strength of wearers performing elbow flexion motions by applying the SFM as an actuator (see Fig. 1). We herein present the design and manufacturing processes of the SFM and ULE. The ULE features high wearability and portability because it does not limit the ROM of the wearers and can be folded into a small palm-sized volume. Furthermore, this article developed an assistance algorithm by applying sEMG sensors and evaluated the assistive effects of the proposed ULE.

Fig. 1. Upper limb exosuit based on fabric muscles. The proposed ULE supports elbow joint flexion motions of wearers without disturbing their free motions.

The rest of this article is organized as follows. Section II discusses the design and fabrication of the ULE. Section III evaluates the ULE. Section IV presents the assistance algorithm of the ULE. Finally, Section V concludes this article.

II. Design and Fabrication of ULE

A. Microdiameter SMA Spring-Based Fabric Muscle

An SMA spring contracts because of the phase change from martensite to austenite caused by an increase in temperature attributed to electric heating. The SFM designed in a previous study [17] comprises 500 μm-diameter SMA springs, and it has the inherent softness and flexibility of fabric. Only a small battery is required to actuate the SFM, and its stroke, generating force, and actuation speed are controlled based on the magnitude and the time of the currents supplied. Although SFMs have advantages such as high power density, high strain, lightweight, and silent operation, they suffer from the problem of a slow relaxation speed.

This problem was addressed in this article by applying the following methods. The relaxation speed was determined by the cooling speed of an SMA spring. Based on this principle, the surface area-to-volume ratio of the SMA spring was increased to enable the faster dissipation of heat. The surface area-to-volume ratio increases, and the cooling speed becomes faster with a decrease in the diameter of an SMA wire used for manufacturing an SMA spring. Furthermore, small-sized fans were attached to the breathable cover of the SFM to supply cooling air to the SMA spring by force and for increasing the cooling speed.

In this article, an SMA spring thread was manufactured by continuously winding a NiTi SMA wire with a diameter of 40 μm on a core wire with a diameter of 180 μm in the form of a coil spring. The SMA wire diameter is thinner than that of a human hair and a skeletal muscle fiber having a diameter of 100 μm and 50 μm, respectively. When the core wire in the SMA spring thread is removed, only the SMA spring remains.

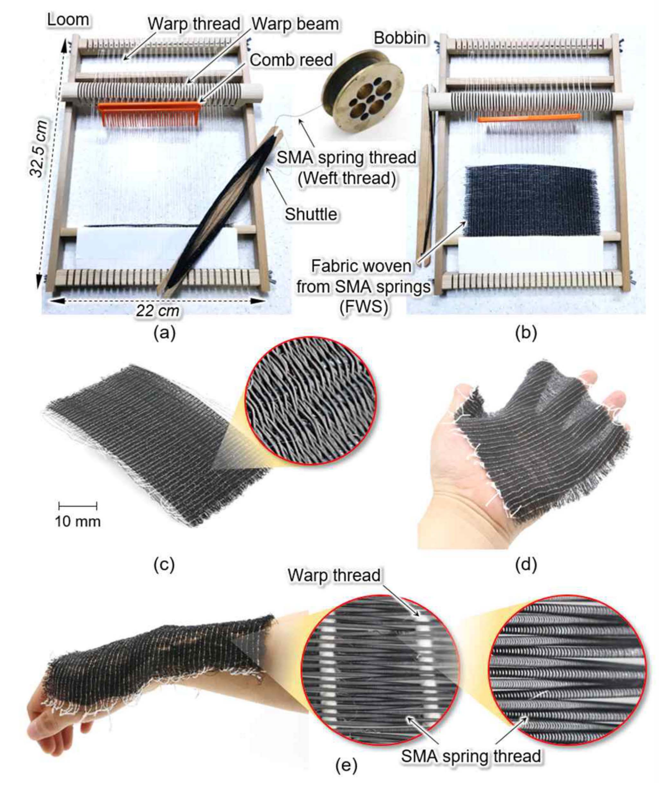


Fig. 2. (a) Prepare the loom for weaving the SMA spring. The loom consists of a warp thread, warp beam, comb reed, and shuttle. The warp threads are placed vertically in the length direction of the fabric, and the weft thread crosses the warp threads in the horizontal direction. The SMA spring thread is used as the weft thread. The SMA spring thread wound on the bobbin is transferred to the shuttle and wound on it. (b) Shuttle is repeatedly passed between the warp threads based on the size of the FWS to be fabricated. The comb reed aligns the SMA spring thread tightly by sweeping the weft threads downward. (c) After weaving to the desired length, cut the warp thread, and remove it from the loom. (d) To remove the core wires in the weft thread, the FWS is immersed in the acid solution for 30 min. Once the core wire is removed, the FWS becomes silky and very light, with a texture similar to that of a thin fabric. (e) FWS is similar to a skeletal muscle, which is composed of muscle fibers.

We developed a process to fabricate an FWS by weaving SMA spring threads. Previous studies focused on weaving or knitting SMA wires to create SMA-based actuators [18]–[20].

However, as the SMA wire has a low contraction rate of 2–5%, it cannot generate skeletal muscle-level contractile displacement. A fabric woven from SMA wire suffers from the same limitation. On the other hand, a fabric made of SMA springs can greatly improve the contractile displacement, making it suitable for suit-type wearable robots [21]. The SMA spring thread is thin and flexible, with a diameter of 220 μm; thus, the fabric can be directly woven with it.

Fig. 2 shows the process of weaving FWS using SMA spring thread. It is necessary to weave the SMA spring without removing the core wire to maintain the stiffness required for weaving while maintaining the shape of the thread. After weaving the FWS to the required size with core wire, the core wire is removed by immersing it in an acidic solution. This results in a soft and flexible FWS.

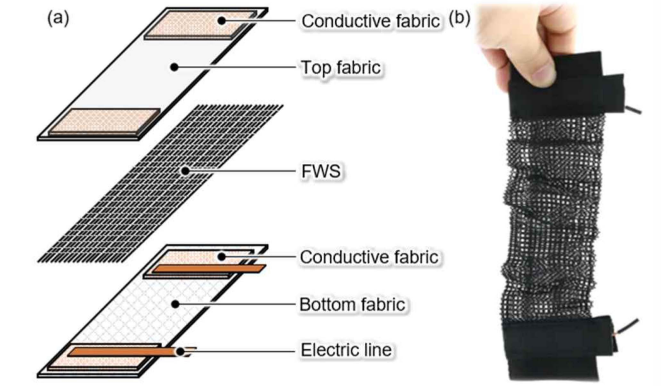


Fig. 3. (a) Configuration of the SFM. First, we stitched the conductive fabrics (1168, Adafruit, NYC, USA) to the top and bottom fabric. Then, we stitched the FWS on the bottom fabric. The electric lines supply the external current to the FWS. They are fixed together with the FWS by sewing them together with the conductive fabrics of the bottom fabric. The top fabric is then covered over the bottom fabric and sewn. (b) Fabricated SFM.

SFM was fabricated by inserting the FWS between breathable covers and connecting electrodes. Fig. 3 shows the fabricated SFM and its configuration. One SFM uses an FWS composed of approximately 800 SMA springs and has a load capacity of approximately 40 N. Two SFMs with the 40-N load capacity were fabricated and overlapped to produce an SFM with an 80-N load capacity; the mass was 36 g. An SFM can be fabricated based on the size of the body part where the SFM is attached and the load capacity required. The number of SMA springs inserted in the SFM can be increased when an SFM with a higher load capacity is needed. Alternatively, several SFMs can be overlapped [17].

Isotonic and isometric tests were conducted to evaluate the generating force and the contraction displacement of the SFM. In the isotonic test, the upper end of the SFM was fixed to a frame. An 8 kg weight was added at the bottom end of the SFM. A laser displacement sensor (LK-G500, Keyence Corp, Japan) was installed on the floor under the SFM. The maximum stroke and contraction strain of the SFM were measured when currents of 10, 12, 14, and 16 A were supplied for 1.5 s, respectively.

Furthermore, the maximum temperature of the SFM was measured by attaching a thermocouple (TT-K-30-SLE, Omega Engineering Inc., South Korea) to the SMA springs in the SFM. A battery of 22.2 V was employed for the power supply; the experiment was repeatedly conducted five times, and Fig. 4(a) shows the experimental results. The contraction displacement and maximum temperature of the SFM increased with an increase in the supplied current.

The initial length of the SFM was 150 mm at a room temperature of 26 °C. The length of the SFM increased to 190 mm when an 8-kg weight was attached at the bottom of the SFM. When the supplied current was 10 A, the temperature of the SFM increased to 51 °C and its length contracted to 133.2 mm. The contraction displacement is 56.8 mm, and the contraction strain

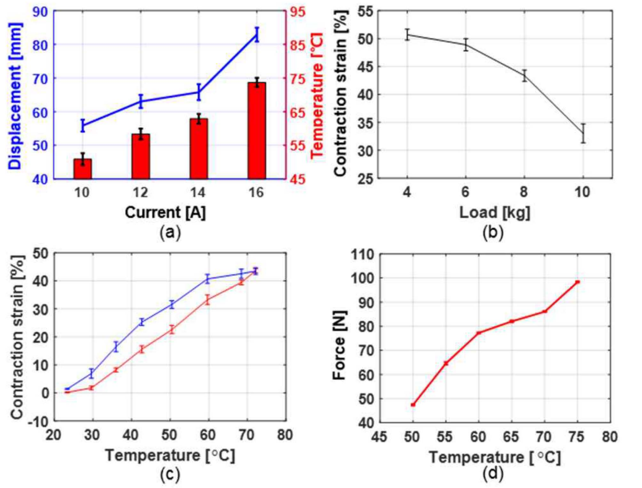


Fig. 4. (a) Comparison of maximum displacement and maximum temperature of SFM depending on input current. (b) Contraction strain of SFM for different loads. (c) Contraction strain versus temperature during heating (red line) and cooling (blue line) under a load of 8 kg. (d) Force generated by SFM depending on maximum temperature of SFM in the isometric test.

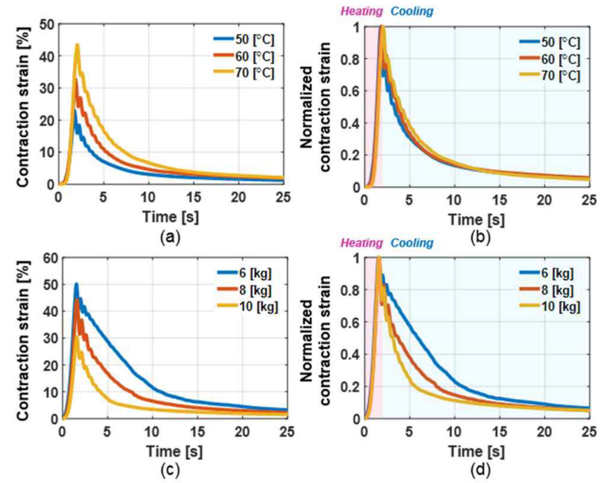


Fig. 5. Comparison of (a) measured and (b) normalized contraction and relaxation curves for different joule heating temperatures, and comparison of (c) measured and (d) normalized contraction and relaxation curves for loads of 6, 8, and 10 kg.

is 30% based on the following:

$$\text{Contraction strain} \quad [\%] = \frac{dl}{l_0} \times 100 \tag{1}$$

where l_o denotes the increased length of the SFM when the weight is added to the bottom end of the SFM; dl represents the contraction displacement of the SFM after the current was supplied. The mass of a bundle of 400 SMA springs used in the SFM was 5.2 g. The SFM could lift an 8-kg weight of approximately 1500 times the mass of the SMA spring bundle; the maximum contraction strain was 45.5% when a current of 16 A was supplied.

To measure the relationship between contraction strain and load, the SFM was fixed to the frame in the same experimental setup, as in Fig. 4(a), and the maximum contraction strain was measured for loads of 4, 6, 8, and 10 kg [see Fig. 4(b)]. The contraction strain was maintained above 40% up to the load of 8 kg. However, when the load was 10 kg, it decreased to 33.4%.

The contraction strain depending on temperature during heating and cooling of the SFM was measured under 8-kg load condition, and it was observed that the SFM has a hysteresis of 10 °C between heating and cooling temperatures [see Fig. 4(c)].

In the isometric test, the length of the SFM was increased and fixed at 190 mm. One end of the SFM was fixed to a load cell (LSB200, FUTEK, CA, USA), and the other end was fixed to the frame. The force generated by SFMs was measured while increasing the temperature from 50–75 °C at an interval of 5 °C. The temperature of the SFMs wax s maintained via proportional-integral (PI) temperature control; the amount of current supplied to the SFM was controlled based on temperature feedback received from a thermocouple attached to the SFM. The experiment was repeatedly conducted five times. Fig. 4(d) shows the result of measuring the generating force of the SFM based on temperature.

Experimental results indicate the SFM generated a force of 48 N at a temperature of 50 °C. The force generated by the SFM increased proportionally with an increase in temperature. Consequently, the SFM generated a force of up to 98.4 N at a temperature of 75 °C. This experiment confirmed that SFM could be applied to wearable robots because it generates a sufficient force even below 50 °C; this is not a dangerous temperature for the human body.

Under the load of 8 kg, the SFM was heated using different joule heating temperatures and then relaxed using natural convection. The contraction and relaxation displacements were measured and normalized for comparison. The maximum displacements occurred after joule heating for 1.5 s, and the contraction and relaxation rates were similar despite the different joule heating temperatures [see Fig. 5(a)]. The relaxation time to recover 80% of the maximum displacement was 5.7 s [see Fig. 5(b)]. In addition, under the different load conditions, the SMF was heated up to 70 °C and then relaxed by natural convection. As the load increased, the maximum contraction strain decreased, and the relaxation speed increased [see Fig. 5(c)]. At the load of 10 kg, the relaxation time required to recover 80% of the maximum displacement was 4.1 s [see Fig. 5(d)].

The SMA spring thread was annealed at 400 °C for 30 min. If the heat treatment condition is not appropriate during the manufacturing of SMA springs, then the contraction displacement is degraded as the actuation is repeated [22]. To observe the degradation with respect to the number of actuation cycles of the SFM, 20 000 cycles of repeated actuation were performed under a load of 3 kg. The contraction displacement did not decrease during the entire cycle, and the uniform contraction–relaxation performance was maintained. This experiment confirmed that the heat treatment condition for the SMA spring thread for SFM was appropriate.

Fig. 6. ULE developed to assist the elbow in flexion and extension, (a) front and back view of exosuit. (b) Unfolding components of ULE. (c) ULE can be folded to be stored in a small bag and held by one hand.

B. ULE Based on Fabric Muscle

The structure of a ULE for supporting elbow flexion motions is as follows: 1) The SFM is attached to the location of the biceps that the human uses to flex the elbow. The SFM is contracted by battery power, and it provides the assistive force required by the wearer to bend the elbow. Fig. 6(a) shows that a ULE comprises a shoulder anchor, a forearm anchor, an SFM, and a controller package (controller and battery). The upper part of the SFM is connected to the shoulder anchor; the bottom part is connected to the forearm anchor. The controller package is located at the lower back, which receives the minimum effect when loads are applied [4]. All components can be easily replaced because they are all modular.

1) Lightness and Flexibility: The ULE supports arms under a wide ROM and frequently changing working conditions. Thus, the components of the ULE need to minimize the increase in arm weight and inertia. SFM can be attached directly to the biceps' region supported by the ULE because it is lightweight, and it slightly increases the weight and inertia to the level of everyday wear.

Shoulder anchors based on straps and buckles are firmly fixed to the body of the wearer to transfer the force to the forearms when SFMs are contracted. Straps made of sponges and mesh fabric provide comfort to the wearer's shoulder. Buckles are used to adjust the ULE to fit the wearer's body size.

The SFM can be easily attached to or detached from the shoulder anchor using a light and flexible zipper (see Fig. 6(a) yellow line). The shoulder anchor, which includes the buckle and the zipper, weighs 70 g. The forearm anchor comprises a lightweight, nonstretch fabric for transmitting the contractile force of the SFM to the forearm; it can be adjusted with BOA (BOA Technology Inc., Denver, USA) based on the thickness of the wearer's forearm.

The controller package comprises a controller (Arduino micro, Boston, USA), current drivers (24V13, Pololu Corp., Las Vegas, USA), and a battery (Anhui Enrichpower Battery Co., LTD, China). Both the width and length of the controller package are 140 mm; the total weight is 227 g. As all components (shoulder anchors, forearm anchors, BOAs, and SFMs) of the ULE, except the controller and the battery, were made of fabric, they are soft and flexible. Since they are fixed via stitching, their flexibility is not hindered, and the lightweight property can be maintained. The mass of the produced ULE is 310 g; the total mass, including the ULE, controller, and battery, is 540 g. Given that the mass of a t-shirt is approximately 460 g, the ULE is as light as daily clothes.

2) Portability: The portability of the proposed ULE is high as it is light and flexible. People can wear lightweight ULEs as comfortable as daily clothes for a long time, even in scenarios where the ULE is not powered because of a discharged battery. All components of the ULE are fabric-based, and therefore, they can be folded or rolled down to a small size that can be carried with one hand or placed in a bag [see Fig. 6(b) and (c)]. This ULE is easy to carry, and thus, people can easily wear this ULE whenever and wherever strength assistance is required.

3) Wearability: Among the components of the proposed ULE, shoulder and forearm anchors have the closest relationship to wearing comfort because these components are directly fixed to the body of the wearer. These should be designed to provide a comfortable fit while being tightly fixed to the wearer's body. An elbow is bent when the biceps, connected via ligaments between the shoulder bones and forearm bones, contract. Based on this mechanism of elbow movements, the SFM is placed between the shoulder and forearm anchors.

The bands of shoulder anchors come down from the front shoulders, pass through armpits, and are crossed at the back. The back of the shoulder anchors was made of crossed bands to prevent interference from shouldering movements. The shoulder anchor is firmly attached to the shoulder, and the chest [see Fig. 6(b)], thus, becomes a supporting reference point. Then, the contractile force of the SFM is transmitted to the forearm. Since the shoulder anchor is tightened by simply inserting the arms and pulling the strap, people can quickly place the ULE in less than five seconds (see Video S1).

The BOA was applied to allow the wearers to adjust the perimeter of the forearm anchors based on the thickness of their forearms and to easily untie or tighten these anchors. The wearer tightens the BOA to the forearm only when strength assistance is required. If the BOA is released in scenarios where strength assistance is not required, it is in the same state as the sleeves of daily clothes, and it frees the wearers from pressure. The inner lining of the forearm anchor is made of an antislip fabric so that it can be wrapped around the forearm more firmly. Further, the strap at the bottom of the SFM and the forearm anchor were connected with a buckle (see Fig. 6(a) pink line) to allow adjustments to the length according to the wearer's arm length.

III. Evaluation of the ULE

A. Evaluation of the Range of Motion

If the ROMs for cases with and without ULE are similar to each other, the ULE provides the wearer with comfortable

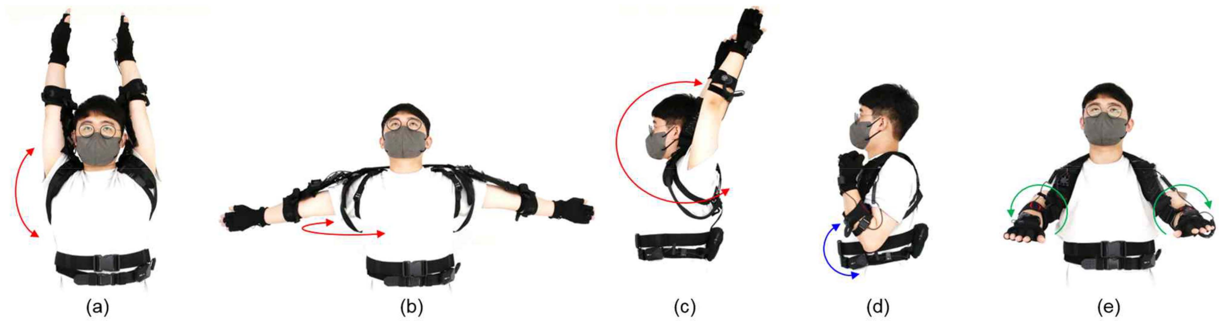


Fig. 7. Photograph of the motion for test. (a) Shoulder abduction and adduction. (b) Shoulder horizontal flexion and extension. (c) Shoulder flexion and extension. (d) Elbow flexion and extension from rest pose. (e) Forearm supination and pronation.

TABLE I
COMPARISON OF ROM OF THE UPPER LIMB

Movement	Range of motion (degree)	
	Without ULE	With ULE
Shoulder adduction and abduction	182.9	181.3
Shoulder extension and flexion	210.6	207.5
Shoulder horizontal extension and flexion	91.5	88.4
Elbow extension and flexion	138.7	135.4
Forearm supination and pronation	152.8	152.1

TABLE II
SUBJECT INFORMATION

Subjects	Age (years)	Height (cm)	Weight (kg)
1	28	177	95
2	25	174	76
3	27	174	95
4	25	179	74
5	24	175	73
6	24	169	57
7	23	171	70
8	25	167	58
9	24	170	69
10	25	172	67

wearability such as that of daily clothes, without disturbing the motions of the wearer.

Previous studies [10], [11], [23] compared the ROM of people before and after they wore wearable robots for upper limb assistance. These studies evaluated the ROM based on the motions of the upper limb joints. However, this article excluded hands and fingers in the evaluation process because these parts are not related to the motions of a person wearing the ULE.

As shown in Fig. 7, a subject is asked to repeatedly perform five types of motions related to shoulders and arms 10 times before and after wearing the ULE. A motion capture device (Perception Neuron, Florida, USA) is utilized to measure and compare the ROM of the subject.

Ten inertial measurement unit (IMU) sensors attached to the upper limb of the subject measured the motions of the upper limb joints at a sampling rate of 120 Hz. When measuring the ROM while wearing ULE, the subject wore both the motion capture device and the ULE simultaneously. The ULE was not in a strength assistance state.

Table I summarizes the results of comparing the ROM of the subject before and after wearing the ULE. The ROM for the shoulders of the subject before wearing the ULE was similar to that after wearing the ULE. This result was obtained because the proposed ULE did not interfere with shoulder motions. A shoulder joint rotation motion [see Fig. 7 (a)–(c)] cannot be readily performed or might apply considerable pressure to an arm when a heavy wearable device or a hard link is attached to the arm. The results of measuring the ROM for the elbows and forearms indicated that the range of elbow extension and flexion, and the forearm supination and pronation motions before wearing the forearm anchors of the proposed ULE were similar to those after wearing the forearm anchors [see Fig. 7 (e)]. These results show that SFMs and forearm anchors are light and flexible and have a trivial effect on the motion of the elbow or forearm.

B. Evaluation of Assistive Force

The assistance performance of the proposed ULE based on dumbbell curl motions of 10 healthy adult men who wore the proposed ULE and lifted a dumbbell are explained below. All subjects wore the same ULE because the ULE can be adjusted to fit 10 subjects (see Table II) with different body sizes and shapes. In this experiment, subjects held a switch in their hand, which does not hold a dumbbell, and they activated the ULE by pressing the switch when an assistive force was required. The muscle activity of subjects under the no-suit condition was compared with that under the suit-assistance condition to analyze the assistance performance of the ULE. To this end, sEMG sensors (Cometa Mini Wave, Milano, Italy) were attached to the biceps of the right arms of the subjects. This experiment is conducted based on a predetermined motion sequence, as shown in Fig. 8 (a). Before the motion measurement, all subjects were provided a sufficient amount of rest. The experimental protocol was approved by the Institutional Review Board at the Chungnam National University (202101-SB-013-01).

The subject prepares to lift a 5-kg dumbbell in the initial state with the SFM of the ULE relaxed under the suit-assistance condition [see Fig. 8 (a)-i)]. An electric current is supplied to the SFM of the ULE when the subject presses the switch while starting to bend the elbow. While the subject slowly lifts the

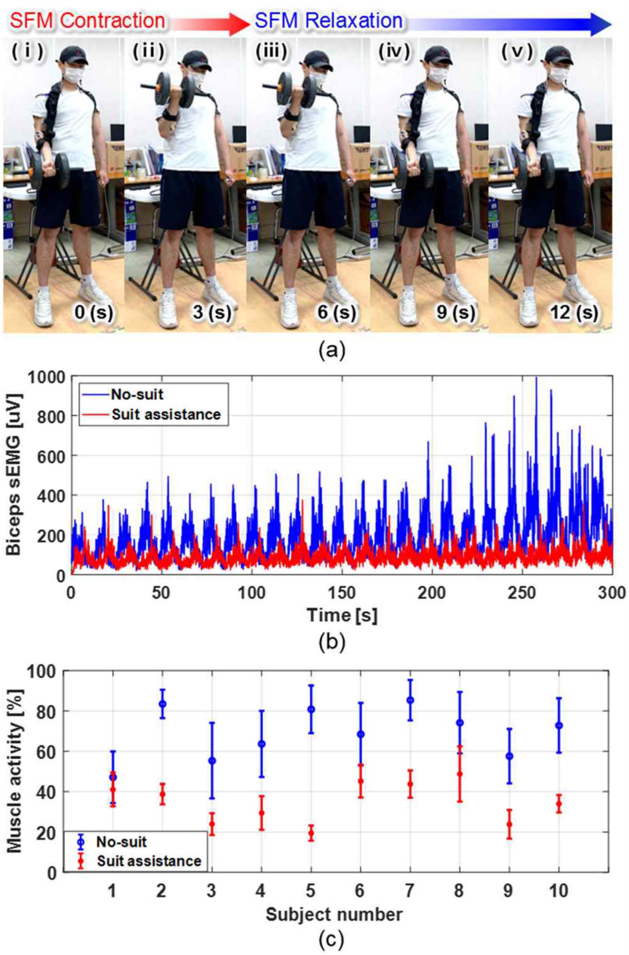


Fig. 8. Evaluation of the performance of the ULE for generating assistive force. (a) Order of lifting and putting down a dumbbell. (b) Comparison of sEMG signals obtained from the biceps of a representative subject before and after the subject was supported by the ULE. (c) Comparison of muscle activity of all subjects before and after they were supported by the ULE.

dumbbell for 3 s, the SFM contracts to support the strength of the biceps [see Fig. 8 (a)-ii)]; no more current is supplied to the SFM. Under this condition, the subject continues to lift the dumbbell with his elbow bent [see Fig. 8 (a)-iii)]. After this motion, the subject slowly unfolds his elbow for 3 s [see Fig. 8 (a)-iv)]. Finally, the subject rests with his elbow extended for 3 s before repeating the motions [see Fig. 8(a)-v)].

All subjects repeatedly performed the aforementioned dumbbell curl motions 25 times in the same manner under the no-suit and suit-assistance conditions to measure the signals derived by the sEMG sensors attached to their biceps. Fig. 8(b) shows sEMG signals obtained from the biceps of a representative subject. The sEMG data were recorded at a 2 kHz sampling rate and filtered with a low-pass filter of 60 Hz and a high-pass filter of 3 Hz. After rectification, the peak and mean values were calculated.

During one cycle of a dumbbell curl motion, sEMG signals obtained from the biceps reached the peak in the dumbbell lifting motion. Both the peak and mean values of the sEMG signals were decreased signals under the suit-assistance condition. The peak of sEMG signals observed under the no-suit condition decreased by up to 70% under the suit-assistance condition. We observe a notable point in the experimental results shown in Fig. 8(b). Under the no-suit condition, the biceps of the subject became fatigued as the subject repeatedly performed dumbbell curl motions; the peak of the sEMG signals at the biceps gradually increased to reach 1000 μV. However, the peak of the sEMG signals from the biceps was maintained in the range below 380 μV despite repeated dumbbell curl motions under the suit-assistance condition. This result indicates that the ULE reduces fatigue in muscles and assists wearers [24]. Thus, wearers can perform repetitive arm motions under specific loads for longer.

The comparison of muscle activity of all subjects when they lifted 5 kg dumbbells [see Fig. 8(c)] shows that the muscle activity under the suit-assistance condition decreased by 43% on average and 60% at maximum compared to that under the no-suit condition.

IV. Assistance Algorithm of ULE

A. sEMG-Based Driven Method

Several assistance algorithms that use different types of sensors have been widely utilized to estimate the intention of wearers, provide feedback on muscular movements, and generate an appropriate assistive force [25]. Methods for providing motion commands to wearable robots include switch input, voice commands, and sensor-based intention recognition. Among these, the switch input is a method of pushing a pressure switch installed at the tip of a finger or under the chin; it is a simple method than can accurately deliver the wearer's intention. Intention delivery through voice commands is a simple and intuitive method. While a wearer performs a specific task, pushing a switch may not be easy. It may be inconvenient to operate an external switch for an older person or a patient with a physical disability. With the development of voice recognition technology, motion commands through natural language have great potential in industrial wearable robotics [26].

Sensor-based intention recognition measures and analyze the biological signals or movements that occur when the wearers move and identifies their intention to provide motion commands to the ULE automatically. This method utilizes various types of sensors such as force, displacement, IMU, and EMG sensors.

The sEMG sensors have been widely used to measure the muscle activity and fatigue of the wearers because their electrodes can be easily attached to the skin and measure EMG. In this article, the sEMG sensor was applied to the intention recognition of the wearer's flexion–extension motion.

It is necessary to select the muscle location that generates the EMG that best reflects the movement of the elbow joint and attach the electrode to enable the ULE to recognize the intention of the wearer using the sEMG sensor accurately. To this end, sEMG sensors (MyoWare Muscle Sensor AT-04-001, Advancer Technologies, NC, USA) were attached to three locations (biceps, triceps of the upper arm, and antebrachial muscle of the

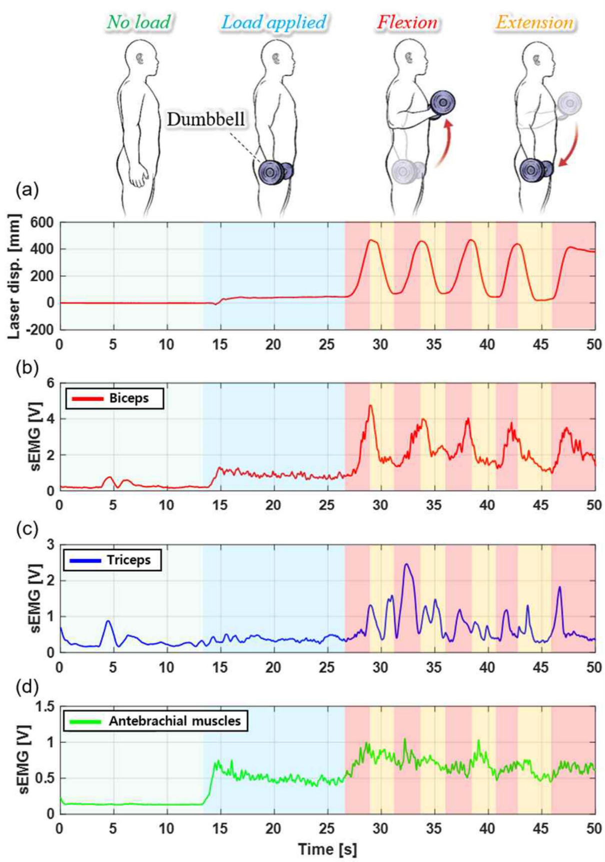


Fig. 9. (a) Elbow flexion and extension motions of the wearer. (b) Output of sEMG sensors installed at the biceps, (c) triceps, and (d) forearm muscle of the wearer.

forearm), and sEMG signals obtained for designated motions were compared with the elbow motions of wearers.

Laser displacement sensors installed on the ground measured the displacement of a target face attached to the back of the hands of the wearers to obtain the motion data of elbow flexion and extension. The wearer waited for 13 s extending the elbow with bare hands (no-load state) and then for another 13 s extending the elbow with a 5-kg dumbbell in hand (load applied state). Subsequently, they repeated lifting and lowering the dumbbell five times with a cycle of 5 s (flexion and extension state).

Fig. 9(a) shows the elbow flexion and extension motions of wearers. Fig. 9(b)–(d) shows sEMG signals obtained from the biceps, triceps, and antebrachial muscles, respectively. When a no-load state changed into a load applied state, the values of sEMG signals obtained from biceps and antebrachial muscles increased and were maintained.

In elbow flexion and extension motions with a cycle of 5 s, the values of sEMG signals clearly increased during the flexion motions (see Fig. 9(b), red sections) and decreased during extension motions (see Fig. 9(b), yellow sections) only at the biceps based on elbow movements. The values of sEMG signals obtained from triceps and antebrachial muscles did not clearly reflect elbow motions.

Biceps were verified as the most accurate muscles where sEMG signals can reflect elbow flexion and extension motions based on measurement results. Therefore, attaching the electrode of sEMG sensor to the biceps was evaluated as the most appropriate solution for recognizing the intention of wearers.

The intention recognition method was designed to provide assistive force by applying joule heating to the SFM of the ULE when the wearers lift loads by bending their elbows and the values of sEMG signals obtained from the biceps exceed the designated threshold. The threshold was set to 24% of the maximum voluntary contraction (MVC).

The mannequin was equipped with a ULE to verify the accurate operation of intention recognition; sEMG sensors were attached to the right biceps of the subjects. A 1-kg dumbbell was attached to a hand of the mannequin wearing the ULE. A laser displacement sensor installed at the ground measured the contraction–relaxation displacement of the SFM by measuring the displacement of the target face attached at the lower end of the SFM. The subject performed elbow flexion and extension motions with a cycle of 15 s twice. Furthermore, the subject was asked to bend his elbow for 3 s and extend it for 12 s for a cycle while holding a dumbbell with one hand.

Detailed processes of verifying the intention recognition for the elbow flexion and extension are provided as follows.

1) The subject prepared to lift a dumbbell in the initial state where the SFM of the ULE worn by the mannequin was relaxed [see Fig. 10(a)-1)].
2) As soon as the subject bent his elbow, the values of sEMG signals at the biceps increased; these values reached the threshold 0.85 s after elbow flexion was initiated.
3) The SFM of the ULE began contracting [see Fig. 10(a)-2)]; the SFM continued to contract, and the forearm of the mannequin was raised [see Fig. 10(a)-3)].

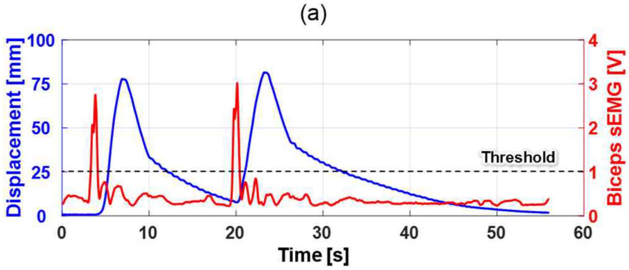
Furthermore, the final contraction displacement of the SFM was 75.6 mm [see Fig. 10(a)-4, (b)], and the forearm of the mannequin was bent at an angle of 90° based on the upper arm (see Fig. 10(a)-4), see Video S2). The experimental result indicated that the intention recognition method applying sEMG sensors attached to the biceps exhibited adequate performance.

B. Assistance Method of ULE

The ULE provides an assistive force to help wearers flex their elbows or maintain flexion based on intention recognition using sEMG sensors when loads are applied to the biceps. Fig. 11 depicts the assistance method of the ULE.

The ULE determines the intention of elbow flexion by comparing the values of the sEMG signals obtained and the threshold. Therefore, a difference of the output of the sEMG signals in a load-applied state and an elbow flexion state should be distinguished. The loads changed in the range of 0, 1, 3, 5, 7, and 10 kg to compare the output of sEMG signals in a load-applied state and a flexion state for determining the threshold (see Fig. 12). The maximum difference between sEMG output voltages in the load-applied and flexion states was measured to be 2.1 V for 5 kg, 2.2 V for 7 kg, and 1.9 V for 10 kg. At a 5 kg or higher load, the difference in the sEMG output voltage

(a)



(b)

Fig. 10. Evaluation of the performance for intention recognition motions based on sEMG sensors attached at the biceps. (a) Elbow of mannequin bends according to elbow flexion motion of the subject. (b) When the subject performs an elbow flexion motion, the output of sEMG signals obtained from the biceps of the subject and displacement of the SFM forcing the elbow of the mannequin to bend are measured.

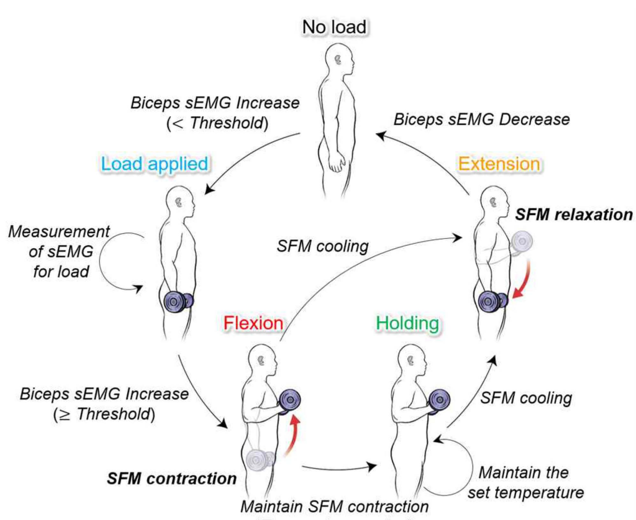


Fig. 11. Assistance method of ULE applying sEMG sensors. When a load is applied to the ULE wearer in a no-load state, the output of the sEMG sensors at the biceps increases slightly. When the wearer bends his elbow to lift the load, the output of sEMG sensors attached at the biceps of the wearer increases simultaneously. Currents are supplied to contract the SFM of the ULE when the output of the sEMG sensors exceeds the threshold. Contractile force generated by the SFM supports elbow flexion. When the wearer holds the load, assistive force is provided based on temperature control. The SFM is cooled to be relaxed when the wearer performs an elbow extension motion.

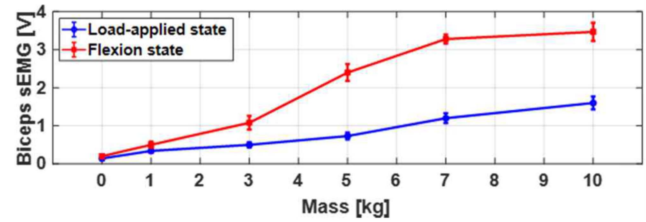


Fig. 12. Output of sEMG sensors at biceps of wearers in a load applied state (blue line) and a flexion state (red line) according to weight of a dumbbell.

is significant enough to be 1.9 V or greater. These are sufficient differences to recognize the intention. However, when a load of 0 or 1 kg was applied, the difference between the sEMG output voltages was 0.2 V or below. The intention for elbow flexion cannot be easily distinguished with this slight difference. Thus, an intention recognition-based assistance method operates only for 5 kg or higher load conditions, and the threshold was established as 2 V.

The amount of assistive force generated by the ULE differs based on the heating temperature of the SFM when ULE wearers perform flexion and holding motions. This study measured the assistive effects of the ULE based on the heating temperature to determine the operating temperature of the SFM in a load-applied state. sEMG sensors are attached to the biceps of the right arms of the subjects wearing ULEs. The temperature of the SFM increased at intervals of 5 °C from 50 to 75 °C under the conditions of 5, 7, and 10 kg applied loads. The output of sEMG sensors was measured when ULE wearers bent their elbows while holding dumbbells and maintained this motion. PI control is performed to control the temperature of the SFM.

The experimental protocol is described as follows. The subject prepares to lift a dumbbell with one arm in an initial state where the SFM of the ULE is relaxed. The subject bends his elbow slowly for 3 s to lift the dumbbell (see Fig. 13, red sections). When the subject lifts the dumbbell, joule heating is applied to the SFM of the ULE to support elbow flexion. Then, the heating temperature of the SFM is maintained at a constant value while the subject keeps the elbow bent (see Fig. 13, green sections). The ULE does not provide assistive force at 26 °C before the SFM is heated; the sEMG output voltage at this temperature is the reference for comparison. The subject places his right upper arm parallel to his body and bends his elbow so that his forearm is at a 90° angle from his upper arm. Then, the subject holds the elbow in a flexion state. The subject repeatedly performs the aforementioned motions five times, and the sEMG signals obtained from the biceps are measured. This article compared the output of the sEMG signals at the biceps according to the heating temperature and load (see Fig. 13).

The comparison results are as follows: For a load of 10 kg, the output of sEMG signals at the biceps at 50, 60, and 70 °C decreased by 49.2%, 49.4%, and 70.3% compared to that at 26 °C, respectively. For a load of 7 kg, the output of sEMG signals at 50, 60, and 70 °C decreased by 70.2%, 72.9%, and 75.9% compared to that at 26 °C, respectively. At 75 °C, the assistive

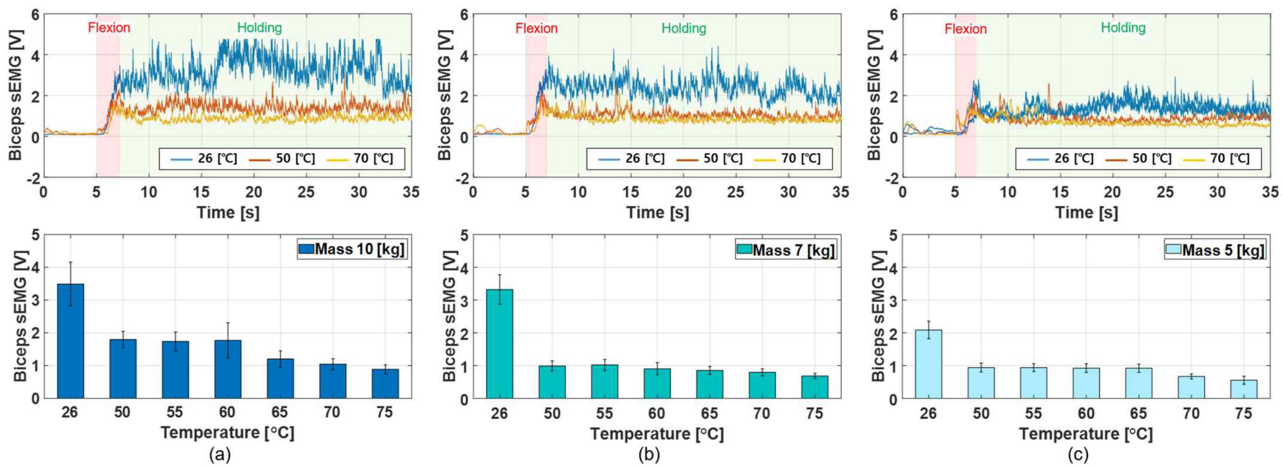


Fig. 13. Comparison of output signals of sEMG sensors at biceps measured according to the heating temperature of the SFM of the ULE when subjects continued lifting a dumbbell with a mass of (a) 10 kg, (b) 7 kg, and (c) 5 kg.

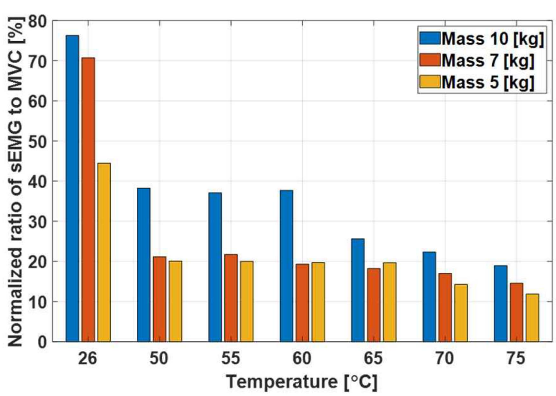


Fig. 14. Evaluation of the performance of the ULE for generating supportive force. The average output of sEMG sensors attached at the biceps of ULE wearers were measured according to heating temperature of the SFM of the ULE while they maintained lifting loads of 5, 7, and 10 kg, respectively. Subsequently, the normalized ratios of the average output of sEMG sensors to the MVC were compared.

effect of the ULE did not increase significantly compared to that of 70 °C. The output of sEMG signals at the biceps at 50 °C decreased by 54.9% compared to that at 26 °C for an applied load of 5 kg. The output of the sEMG signals did not decrease by over 60%, even when the temperature increased up to 65 °C.

Fig. 14 shows the result of normalizing the ratios of the outputs of sEMG signals based on the heating temperature of the SFM to the MVC. The maximum value (MVC) of sEMG signals at the biceps was obtained when subjects maintained lifting loads in the state of the absence of assistance (at a temperature of 26 °C). The MVC was applied as a denominator to calculate the aforementioned ratio, which can be effectively used to compare the degree of reduction of force required for a ULE wearer to lift loads affected by muscular strength support provided by the assistive force of the ULE. When the subject continued lifting the load of 10 kg with one arm, the output of the sEMG signals compared to the MVC was 76.3% at 26 °C. However, this output decreased to 38.2% at 50 °C, which indicates that the amount of force required for the subject to lift the load decreased by approximately 50% compared to that at 26 °C. The ratio of decrease in the output of sEMG sensors did not significantly change when the temperature increased up to 60 °C; it was maintained at a similar level. For an applied load of 7 kg, the outputs of the sEMG sensors compared to the MVC were 70.7% at 26 °C and 21.1% at 50 °C. This result indicated that the force required for the wearer to lift the load decreased by approximately 70%.

Improvement of assistive effects occurred the most for all load conditions when the temperature of the SFM increased to 50 °C. Even if the temperature was increased above 50 °C, the assistive effect was not significantly improved, particularly at 5 or 7 kg. The experimental results verified that the proposed ULE provided assistive effects of 50–70% for 5–10 kg loads when we controlled the temperature to be within 50 °C.

IV. Conclusion

In this article, a soft and flexible FWS was developed and applied to a lightweight, compact, portable, and comfortable ULE. Individuals can easily wear this ULE when they need muscular strength support regardless of the time and the location. The proposed ULE provides wearers with a high level of wearing comfort without disturbing their natural movements. As they can realize the same ROMs in both states of wearing or not wearing ULEs, they can wear these exosuits as daily clothes comfortably for a long time even when power is not supplied to the ULEs. In addition, an experiment was conducted to analyze the motions of subjects lifting dumbbells. The experimental result showed that the muscle activity of the subjects assisted by the ULEs decreased by approximately 60% compared to that of the subjects not assisted by the ULEs. Thus, the ULEs enabled wearers to perform the same motions using muscular strength at half the level of that before they wore the ULEs. This result indicates that support from ULE reduces fatigue applied to muscles and assists wearers to exercise their muscular strength consistently for a long time. Therefore, they can perform repetitive motions under the condition of specific loads for longer periods of time. Another experiment was conducted to examine the motions of subjects who lifted a 10-kg dumbbell and continued to hold it with one hand under the condition where the temperature of the SFM was controlled at or below 50 °C. It would be ideal to use a predefined voltage pattern to ensure rapid actuation while ensuring that controller or sensor-related issues do not cause problems in control. This would allow the sensor to function as a safety feature rather than a control feature. The experimental result showed that the muscle activity of SFM wearers decreased by approximately 50% compared to that of nonwearers.

The proposed ULE has the advantages of excellent wearing comfort, simple structure and design, and lightness. Thus, it is expected to be extensively applied in not only industries but also diverse areas such as rehabilitation and military fields.

However, the intention recognition and support method of the proposed ULE is insufficient for application to a wide range of arm motions under a frequently changing working environment. Additional research is required in the future to recognize and support the intention of multi joints quickly, including the shoulder and the elbow. Another downside that remains to be addressed with ULEs is that their cooling speed is slower than their heating speed and significantly slower than human movements. This could hinder the rapid and sustained operation required when executing certain tasks. Future work will focus on increasing the cooling and cyclic speed of the ULE to increase their usability and optimizing the control algorithm of the SMAs in order to limit their temperature within a safe margin.

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Seong Jun Park was born in South Korea, in 1992. He received the M.S. degree in mechatronics engineering from Chungnam National University, Daejeon, South Korea, in 2019. He is currently working toward the Ph.D. degree in mechanical engineering with Sungkyunkwan University, Suwon, South Korea. He is currently a Researcher with the Department of Robotics and Mechatronics, Korea Institute of Machinery and Materials, Daejeon, South Korea. His research interests include soft actuators and soft wearable robots.

Kyungjun Choi was born in South Korea, in 1986. He received the B.S. and M.S. degrees in mechatronics engineering from the Chungnam National University, Daejeon, South Korea, in 2012 and 2015, respectively. He is currently working toward the Doctorate degree in mechatronics engineering in Chungnam National University, Daejeon, South Korea. He is currently a Researcher with the Department of Robotics and Mechatronics, Korea Institute of Machinery and Materials, Daejeon, South Korea. His research interests include artificial muscles, soft actuators, and wearable robots.

Hugo Rodrigue (Member, IEEE) was born in Montreal, QC, Canada, in 1985. He received the B.Eng. degree in mechanical engineering from McGill University, Montreal, QC, Canada, in 2008, the M.S. degree in industrial engineering from the Ecole Polytechnique de Montreal, Montreal, QC, Canada, in 2010, and the Ph.D. degree in mechanical and aerospace engineering from Seoul National University, Seoul, South Korea, in 2015. He is currently an Assistant Professor with the School of Mechanical Engineering, Sungkyunkwan University, Suwon, South Korea. His research interests include soft actuation, smart materials, and soft robotics.

Cheol Hoon Park was born in South Korea, in 1973. He received the B.S. and M.S. degrees from Yonsei University, Seoul, South Korea, in 1998 and 2000, respectively, and the Ph.D. degree from the Korea Advanced Institute of Science and Technology, Daejeon, South Korea, in 2014, all in mechanical engineering. He was with the Hard Disk Drive Division, Samsung Electronics, in 2000, and was with the Korea Institute of Machinery and Materials (KIMM), Daejeon, South Korea, in 2007. He is currently a Principal Researcher with the Department of Robotics and Mechatronics, KIMM. His research interests include soft actuators, soft wearable robots, and magnetic bearings.