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METHODS

Fabrication of an E-Textile Bioelectrode Array With Screen-Printed Wiring and an Ionic Liquid Gel Toward Cutaneous Whole-Body Electromyography

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ABSTRACT We propose an e-textile bioelectrode array that consists of screen-printed silver paste wiring, a thermoplastic polyurethane insulation layer, and ionic liquid gel-embedded knit textile pads for wearable whole-body medical and healthcare monitoring. The technical challenges in integrating bioelectrodes onto knit textile include forming an insulation layer on silver paste wiring and fixing an ionic liquid gel on the wiring since the screen-printed silver paste wiring is removed by the organic solvent in conventional insulation ink and the adhesive force between the wiring and ionic liquid gel is weak. Our proposed e-textile bioelectrode array structure includes a laser-patterned hot-melt urethane film thermally attached to silver paste wiring on a polyurethane film and fabric to avoid wiring disconnection. An ionic liquid gel is formed and fixed in the conductive polymer-coated knit textile pad, which is fixed to an electrode with a hot-melt urethane film to improve the adhesive force. A 6 × 3 bioelectrode array of 1 cm^2 ionic liquid gel pads is fabricated for electromyography. The impedance between the electrode and skin is 1 kΩ at 1 kHz, which is the same as that of a medical electrode. The adhesive force of the ionic liquid gel to the wiring is improved to 10 N. Finally, the erector spinae muscles are measured with our e-textile bioelectrode array, and the distribution of the flexion relaxation phenomenon can be measured to investigate back pain, leading to a medical diagnosis for plastic surgery or other divisions and healthcare applications.

INDEX TERMS Bioelectrode, conductive polymer, electronic textile, EMG, health monitoring, ionic liquid, screen ink, screen printing, wearable device.

I. INTRODUCTION

In recent years, the development of smart wearable and e-textile technology incorporating bioelectrodes [1], [2], accelerometers [3], [4] and other sensors [5], [6] has attracted much attention to digitize a person's health and cognitive status, analyze them with artificial intelligence (AI), represented by machine learning, and provide advice [7], [8]. Smart wearable devices using e-textiles have the advantage of measuring whole-body vital data, while existing smart watches, including Apple Watch and Fitbit, can only monitor the ulnar artery passing through the wrist, and the resultant data are limited to the heart rate and wrist temperature [9]. Therefore, whole-body vital data monitoring with smart wearables can offer more precise and a wider variety of medical diagnoses and health management [10] since the blood vessels, nerves, and organs are located throughout the whole body. Among smart wearable technologies, e-textile technologies, where

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sensors and electronic components are mounted and wired on a soft flexible textile, can offer a whole-body sensor system due to the advantages of excellent elasticity and compatibility with the human body [11], [12].

In previous studies on e-textiles, bioelectrodes for detecting basic vital electrocardiography (ECG) and electromyography (EMG) data were developed [13], [14], but multiarray bioelectrodes with long-term use for the whole body cannot be fabricated. Most e-textile bioelectrodes utilize dry electrodes, which have the advantage of long-term use without replacement, but their contact electrical resistance for human skin is very large, resulting in large EMG noise [13], [15]. In contrast, wet-gel-type bioelectrodes exhibit low contact electrical resistance for human skin, but they cannot be used long term since wet gels consisting of water and ionic components dry out after several hours [16], [17], [18]. Therefore, a long-term wet-gel-based bioelectrode array is required for back pain measurement or other applications in everyday life usage [19], [20].

To develop a long-term wet gel bioelectrode array, an ionic liquid gel is the ideal electrode material since an ionic liquid is a liquid salt at room temperature that can be gelatinized by polymerization [21], [22], [24]. However, the patterning and fixation of an ionic liquid gel on an electrode array is problematic since it is a wet gel and has no adhesive force to silver paste wiring. Our previous study reported that the conductive polymer poly(3,4-ethylenedioxythiophene) polystyrene sulfonate (PEDOT:PSS) can be patterned on a knit textile to form a 1 cm square electrode and that 1 mm-wide wiring and an ionic liquid can be fixed on the PEDOT:PSS-coated knit textile [20]. The ionic liquid can be fixed on the knit textile, but PEDOT:PSS has a large electrical resistance, and even several centimeter-long wiring for a multielectrode array is impossible due to the several hundred Ω electrical resistance. In addition, the candidate wiring material with low electrical resistance is silver paste, but forming an insulation layer on the wiring to eliminate electrical shorts between the wiring and human skin is difficult because insulation ink with an organic solvent dissolves the silver paste wiring, which is weak against organic solvents when printed with an insulating material ink. Thus, an e-textile-based bioelectrode structure and its fabrication process to make low electrical resistance wiring with an insulation layer and fix an ionic liquid gel are in high demand for applications including diagnosing the movement of muscles in the lower back and other parts of the body, such as for examining lower back pain.

This study proposes a structure and manufacturing process for bioelectrode arrays using an ionic liquid gel on fabric, as shown in Fig. 1. First, a polyurethane film with screen-printed silver paste is bonded onto a cloth, and an insulation layer pattern is formed by laser cutting a hot-melt polyurethane film to achieve low electrical resistance wiring with an insulation layer. This is because the silver paste has low electrical resistance, and the hot-melt type of urethane film can cover the silver paste wiring without dissolving the silver paste due to thermal adhesion and the lack of an organic

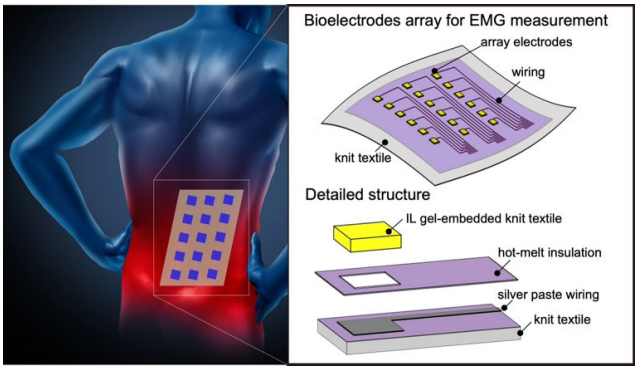

FIGURE 1. Proposed device structure of the.e-textile bioelectrode array for EMG measurements.

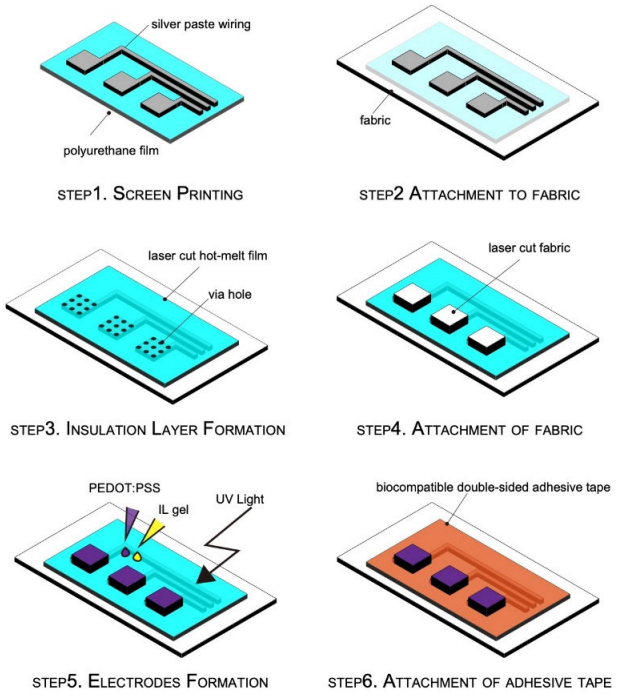

FIGURE 2. Fabrication process.

solvent. Polyurethane film was selected because it is soft and stretchable and does not affect the original texture of the fabric. Second, an ionic liquid gel is formed on the electrode pad after the knit textile is preadhered. This structure makes peeling off the ionic liquid gel difficult because it is integrated into the knit textile bonded to the silver paste wiring and urethane film substrate. Using the proposed structure, a prototype bioelectrode array using the ionic liquid gel is fabricated. The adhesive strength to fabric, insulation properties, and electrical properties as a bioelectrode are evaluated. Finally, EMG measurements, including of the flexion relaxation phenomenon (FRP) and other phenomena, are performed with an array for low back pain diagnosis.

II. EXPERIMENTAL METHOD

A. FABRICATION PROCESS

Fig. 2 shows the fabrication process of the bioelectrode array. Step 1 is screen printing of the silver paste on the

polyurethane film. As polyurethane has lower Young's modulus (2 MPa) than conventional polyimide film, has high flexibility and strength like rubber and is commonly used for the outer skin of clothes, it was used as a substrate material for high conformability to the human skin. Stretchable silver paste (SSP2801, TOYOBO CO., LTD.) is printed on a 100 μm thick polyurethane (PU) film (DUS-202, Sheedom Co., Ltd.) by a screen printer (DP-320, NEWLONG SEIMITSU KOGYO Co., LTD) and stainless mesh screen mask with 28 μm diameter 280 meshes and 10 μm thick resist. The silver paste is first printed not on the knit textile but on the PU film to avoid disconnection of the silver paste wiring caused by the three-dimensional woven structure of the knit textile. In the wiring configuration, the bioelectrode pads are wired to outer circuits using silver paste.

In Step 2, the polyurethane film with screen-printed silver paste is bonded onto a 500 μm thick knit textile (TX1009, Texwipe) using a 100 μm polyurethane hot-melt sheet (SHM101-PUR, Sheedom Co., Ltd.). Thermal bonding is performed using a hot press (JL-CO005B, QUICK ART) at 100°C for 15 seconds.

Step 3 is patterning of an insulation layer over the wiring. Fine via holes are formed on the hot-melt sheet in advance using a laser processor (VLS4, Laser Works Inc.).

The laser output power is 12 W, and the processing speed is 5 cm/s. Then, patterned hot-melt sheets are thermally bonded on the PU films with wiring to insulate the wiring part to avoid direct contact between the wiring and the skin.

Step 4 is adhesion of the textile that holds the ionic liquid gel in place. First, a laser processor is used to cut the knit textile to a size of 1 cm $\times$ 1 cm, the same size as the electrode shape. The textile is then thermocompressed onto the electrode pad, which are thermally bonded by the hot-melt sheet formed in Step 3. The condition for this process is 100°C for 15 seconds.

Step 5 is gelatinization of the ionic liquid on the textile. Ionic liquids are salts that are in a liquid at room temperature and can be used for long hours, in contrast to electrolytes that use water as a solvent, and therefore are used as gels for bioelectrodes. After 50 μL/cm^2 of the conductive polymer formulation permeates into each of the textiles attached in step 4, the knit textiles are dried in a constant-temperature oven at 80° for 30 minutes. The conductive polymer formulation consists of 80 mL of a PEDOT:PSS dispersion (Clevious TM PH1000, Heraeus), 20 mL of ethylene glycol (Sigma–Aldrich), 40 μL of 4-dodecylbenzenesulfonic acid (Sigma–Aldrich) and 1 mL of 3-methacryloxypropyltrimethoxysilane (Sigma–Aldrich). Next, 20 mL/cm^2 of the ionic liquid (1-ethyl-3-methylimidazolium-ethyl sulfate, Sigma–Aldrich) is added to the electrode, followed by 25 mL/cm^2 of the ion gel. The ion gels are prepared with 1-ethyl-3-methylimidazolium-ethyl sulfate, poly(ethylene glycol) diacrylate (Sigma–Aldrich), and the photoinitiator 2-hydroxy-2-methylpropiophenone (Sigma–Aldrich) at the ratio of 0.6/0.35/0.05.

Step 6 is attachment of the biocompatible tape that bonds the skin to the e-textile bioelectrode. Biocompatible double-sided tape (SH 1577-S, Innovect Co., Ltd.) is applied to the e-textile bioelectrode surface after the electrode pads are cut out using a laser cutting machine. The screen-printed wiring is connected to a bioelectrode amplifier chip (RHD2132, Intan Technology) via a flexible printed circuit board to amplify and record myoelectric signals.

B. EVALUATION

The fabricated bioelectrode arrays for EMG were evaluated in terms of the following parameters: 1) the patterning resolution of the silver paste wiring and the impedance of the wiring before and after hot-melt film insulation, 2) the impedance between the conventional and proposed bioelectrodes and human skin, and 3) the adhesive strength of the electrode to the textile substrate.

For 1), silver paste wiring with different widths was screen printed. The electrical resistance of the silver wiring printed on the hot-melt polyurethane film was measured before and after insulating it with the hot-melt film using an LCR meter and analyzer (IM3533-01, HIOKI), which can measure the impedance ranging from 100 m to 100 MΩ by sweeping the frequency from 1 mHz to 200 kHz, and the results were compared. For 2), the impedance between the electrode and the skin was measured with an LCR meter in the range of 100 to 10 MHz while the e-textile bioelectrode was attached to the forearm. Medical patch EMG electrodes (Muscle Sensor Surface EMG Electrodes H124SG Covidien, Adafruit Industries LLC) were also attached to the forearm and examined for comparison, after wiping human skin with moistened cloth containing alcohol and removing the skin erosion. For both electrodes, ten measurements were performed, and the mean and standard deviation were determined. For 3), a peeling test was performed using a tensile tester (FTN1-13A, AIKOH Engineering) to evaluate the adhesive strength of the electrode. A schematic diagram of the peel-off test is shown in Fig. 3(a), where a 1 cm $\times$ 1 cm silver paste electrode pad was screen-printed in the center of the PU film and a 1 cm $\times$ 1 cm ionic liquid gel electrode was created. There are two methods for fixing the ionic liquid on the silver paste: direct gel molding using a mold (Fig. 3(b)) and the proposed method of attaching fabric electrodes with a hot-melt film (Fig. 3(c)). The electrodes were fixed with instant adhesive (NW-15, NICHIBAN Co. LTD.), and the polyurethane film, which is the substrate, was pulled by a force tester to measure the force required to peel off the ionic liquid gel electrode pad at a peel rate of 20 mm/min.

III. EXPERIMENTAL RESULTS

A. FABRICATED DEVICE

Fig. 4(a) presents a photograph of the fabricated device, which shows a highly flexible 6 $\times$ 3 bioelectrode array on the knit textile. Fig. 4(b) displays a magnified view of the bioelectrode pads, which shows that the ionic liquid gel soaks

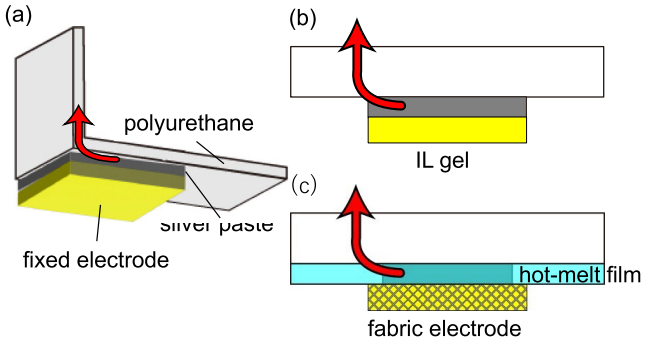

FIGURE 3. Adhesion test conditions (a) experimental setup, (b) ionic gel only, and (c) ionic gel embedded in the knit textile.

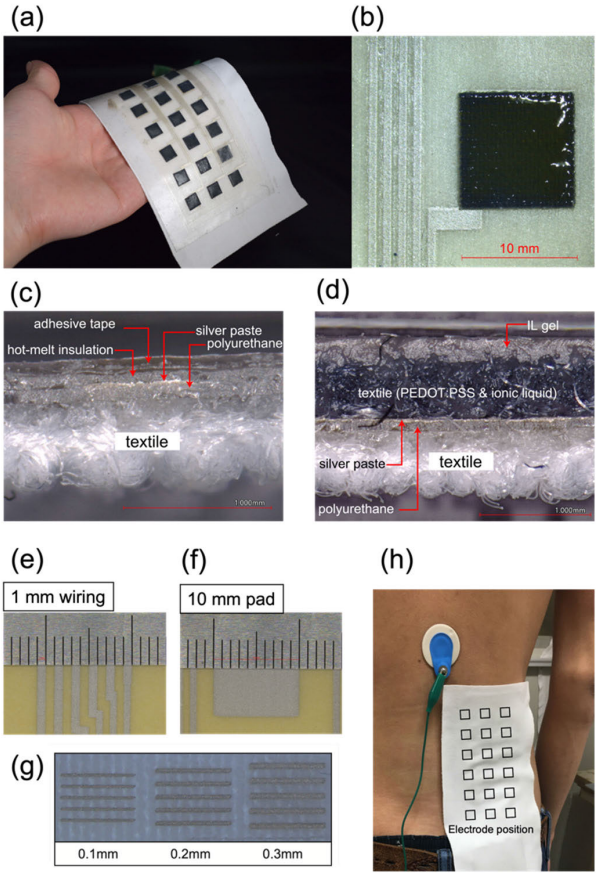

FIGURE 4. Fabricated device photographs (a) Fabricated e-textile bioelectrode array, (b) PEDOT:PSS/ionic liquid gel fabric electrode, (c) cross-sectional view of the insulation layer on the silver paste wiring, (d) cross-sectional view of ionic liquid gel/PEDOT:PSS embedded in the knit textile pad, (e) 1-mm-wide screen-printed silver paste wiring, (f) printed silver paste pad for the ionic liquid gel, (g) resolution of the screen-printed silver paste wiring, and (h) attachment of the e-textile bioelectrode array to a human waist.

into the cloth and is fixed on the textile substrate. Fig. 4(c) and Fig. 4(d) show cross-sectional images of the insulated wiring and electrode pads, respectively. The wiring is completely insulated by the hot-melt film. PEDOT:PSS and the ionic liquid gel are soaked into the knit textile, and both electrical and mechanical connections with the wiring are obtained. Figs. 4(e), (f), and (g) show the resolution of the silver paste

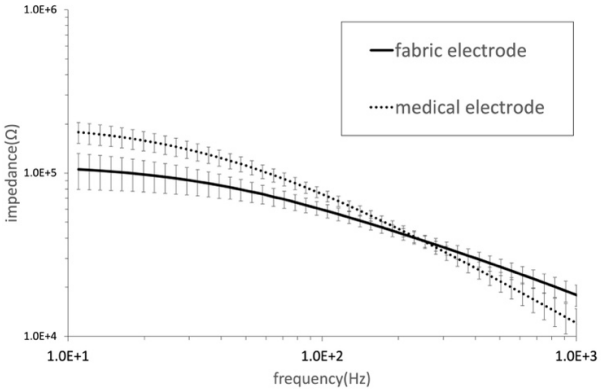

FIGURE 5. Comparison of the modulus of the impedances of the fabricated bioelectrode and a conventional medical electrode.

wiring that can be formed. The lines and spacing of the wiring are 1 mm by 1 mm, the electrode pad is printed in a 1 cm square, and the results indicate that wiring down to 0.1 mm wide can be formed. The wiring can be printed with sufficient accuracy to support arrays of skin electrodes (the typical area is approximately 1 cm^2). Fig. 4(h) shows the attachment of our fabricated bioelectrode to the human back with biocompatible double-sided tape.

B. INSULATION CHARACTERISTICS OF THE HOT-MELT POLYURETHANE FILM

The electrical resistance of the wiring of the bioelectrode before and after insulating the wiring with the hot-melt polyurethane film was investigated because conventional insulation screen ink decreases the electrical resistance of the silver printed silver paste due to its contained organic solvent. The sheet resistance of the printed wiring was approximately 0.1 Ω/sq. The change in resistance after insulation of the wiring with the hot-melt polyurethane film was small, and the resistance after insulation was 98.04% of the resistance before insulation. This is because the hot-melt film does not contain solvents and does not dissolve the silver paste wiring.

C. IMPEDANCE BETWEEN BIOELECTRODES AND SKIN

The electrical impedances between human skin and PEDOT:PSS/ionic liquid gel textile electrodes were measured and compared with those of commercially available medical bioelectrodes. Since the skin and the electrode are in contact with the ionic liquid gel, the impedance between the electrode and the skin can be considered as an equivalent circuit model with resistance and capacitor components connected in series. Fig. 5 shows the impedance results of the fabricated electrode and the medical circular electrode of 1.5 cm in diameter versus the measured frequency. The impedance modulus of the fabric electrode at 500 Hz was 26.0 kΩ (standard deviation (SD) = 3.01kΩ). On the other hand, the impedance magnitude of the commercial medical electrode measured under the same conditions was 21.6 kΩ (SD = 3.00kΩ). Thus, the electrical characteristics of the cloth electrode are equivalent to those of common wet electrodes, and our electrodes can be used for EMG measurement.

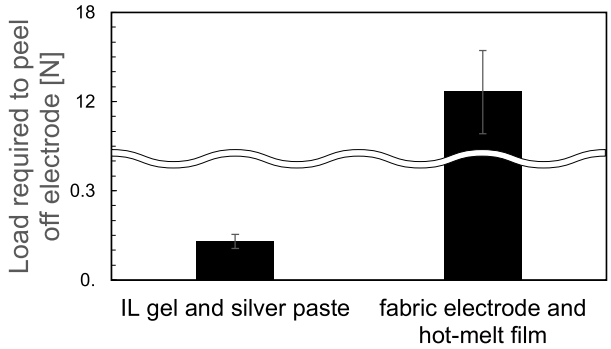

FIGURE 6. Comparison of the adhesive force of the ionic liquid gel only and the ionic liquid gel embedded in the knit textile.

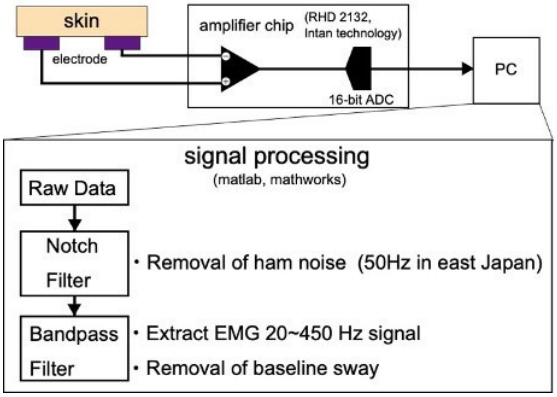

FIGURE 7. Experimental setup for measurement and its signal processing.

D. ADHESIVE STRENGTH OF BIOELECTRODES TO THE SUBSTRATE

Fig. 6 shows the results of the adhesive strength of the ionic liquid gel bioelectrodes to the silver paste wiring. The vertical axis is the peak force required to pull off the 1 cm × 1 cm ionic liquid gel electrode (N = 3). The ionic liquid gel electrode peeled off at 100 mN when the gel was directly fixed on the silver paste electrode, whereas the ionic liquid gel absorbed in the textile electrode required 10 N to be peeled off. For EMG measurement, the bioelectrodes may rub against the skin during body movements, resulting in peeling off of the gel. The 10 N adhesive force between the silver paste wiring and the ionic liquid gel with the textile pad is sufficiently strong for a wearable EMG measurement device since the conventional adhesive force between polyurethane films was reported to be 5 N in a previous study [23].

IV. DEMONSTRATION OF EMG MEASUREMENT

A. EMG MEASUREMENT OF THE FOREARM

EMG recording of the forearm with our e-textile bioelectrode array and the conventional medical electrode was produced, and their resultant signal was compared. EMG recording with our electrode was approved by Research Ethics Committee of the University of Tokyo (No. 19-243). Fig. 7 shows the experimental setup. A bioelectrode array and conventional medical electrode were placed on the forearm and signals from the bioelectrodes and medical electrodes were connected to an amplifier (RHD2132, Intan Technology),

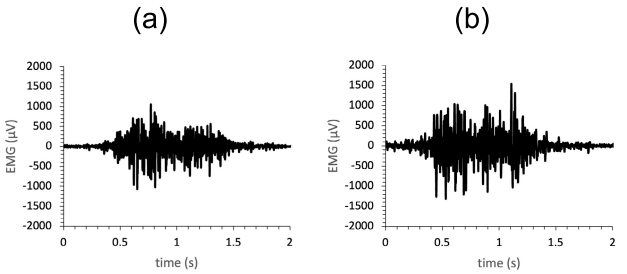

FIGURE 8. Forearm EMG measurement with wrist flexion (a): fabric electrode (b): medical electrode.

which was connected to a PC for data collection. Measurements were taken outside an electromagnetic shield to demonstrate our electrode array for daily application. The acquired signals were processed by standard EMG signal processing which consists of notch filter to remove 50 Hz power supply noise and then 4th Butterworth bandpass filter to pass the signal ranging from 20 Hz to 450 Hz. Figure 8 shows that our bioelectrode array can detect similar EMG signal to the conventional medical electrode.

B. EMG MEASUREMENT OF LOWER BACK

EMG recording of human back pain with our e-textile bioelectrode array was demonstrated. EMG is widely used as a noninvasive method for monitoring muscle activity, and erector spinae muscle measurements are particularly useful for screening patients with lower back pain. By using our bioelectrode array, we investigated the flexion relaxation phenomenon (FRP), a phenomenon in which the erector spinae muscles lose activity when a person is in the maximum forward bending position [24]. In healthy subjects, the FRP can be found, while the FRP cannot be seen in patients with lower back pain. Therefore, an e-textile bioelectrode array was attached to the lower back to monitor the EMG signals of the erector spinae muscles during the movement to raise the upper body to detect the FRP. To detect the FRP with conventional medical electrode, experienced medical staff must place the electrodes in the correct position and diagnose the pain in the hospital. To detect the FRP at home, the bioelectrode array is useful for patient because EMG can be detected from a large area of the back and a precise placement of the electrodes is not required. A bioelectrode array was placed on the spine and above the erector spinae. A common medical electrode was placed on the spine as a reference electrode. Measurements were taken outside of an electromagnetic shield The subject bent over at the electrode placement point for 5 seconds and held the maximal forward bending position for 5 seconds. The patient then rose for 5 seconds and held an upright position for 5 seconds. EMG signals were measured during the 20 seconds of this operation.

Fig. 9 shows heatmaps of EMG signals at four representative timings. The bioelectrode array can detect appropriate and inappropriate EMG signals simultaneously, and only the appropriate signals can be selected, while the conventional

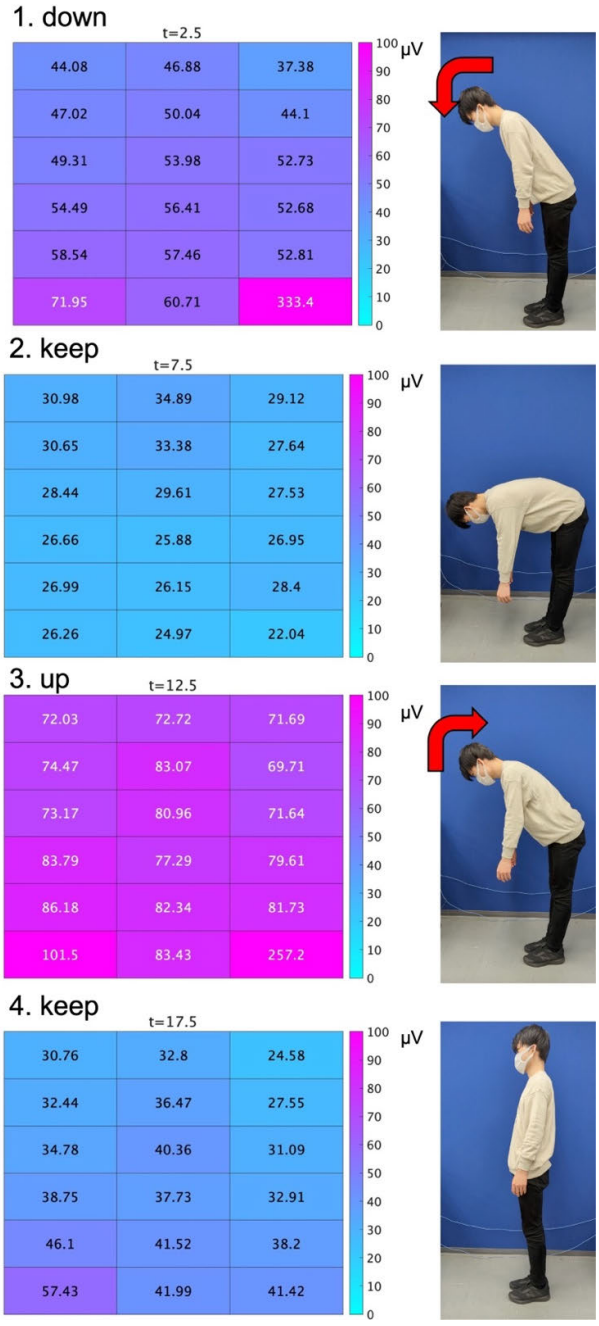

FIGURE 9. EMG heatmap obtained with the e-textile bioelectrodes array.

one-by-one electrode must be pasted several times to place the electrode on the correct location of muscles for appropriate signals. In the heatmap, some electrodes exhibited outliers depending on the posture because they were momentarily separated from the skin. Although the electrode conditions for attachment to the skin are different, the bioelectrode array can offer appropriate EMG signals.

Figure 10 shows the measurement results of the two electrodes for the measurement data used as the basis for the heat map of Figure9. Figure 10 (a) shows the overall measurement, and (b) and (c) show the data for two seconds in particular. Although the overall trend is the same, different signals were obtained for each electrode.

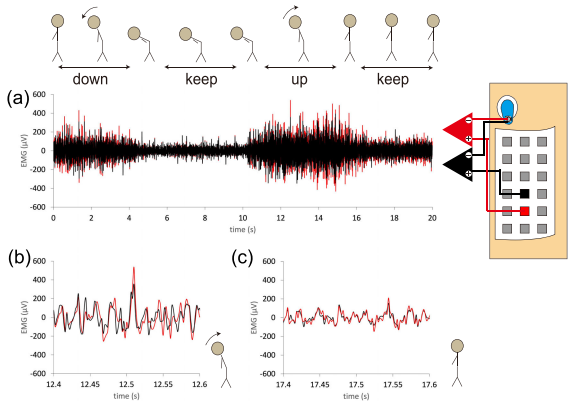

FIGURE 10. Measured EMG signals of lower back.

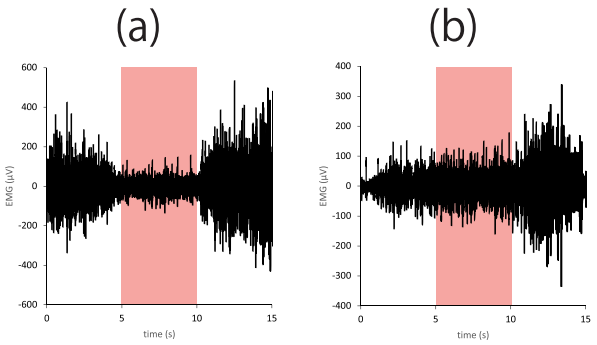

FIGURE 11. (a) EMG in healthy subjects, (b) EMG in patients with low back pain.

Finally, differences in EMG were measured depending on the presence or absence of low back pain. Figure 11 shows the results. (a) is for non-low back pain patients, and (b) is for low back pain patients. The common feature was that myoelectricity was generated during the process of upper body movement, while no myoelectricity was generated during the last 5 seconds. The difference in the amplitude of the last 5 seconds in the steady state may be due to individual differences in the amount of noise and differences between measurements depending on the skin condition and electrode contact conditions. The subjects with low back pain generated electromyograms during the second 5-second period when they maintained their upper body in an inverted position. On the other hand, the electromyograms show that the subjects without back pain do not use their back muscles during the second 5-second period. This difference is the difference between subjects with and without low back pain and those with and without FRP. Therefore, the device proposed in this study has the performance to be used for the diagnosis of low back pain.

V. CONCLUSION

The fabrication process of an e-textile bioelectrode array, which consists of silver paste wiring, a hot-melt polyurethane film insulation layer, and ionic liquid gel-embedded knit textile electrode pads, was proposed and developed for whole-body EMG. The silver paste wiring, which was insulated with a patterned hot-melt polyurethane film, exhibited a low electrical resistance change of 2%. The knit textile electrode pads

for the ionic liquid gel improved the adhesive force between the ionic liquid gel and silver paste wiring, resulting in a 10 N adhesive force for a 1 cm square electrode pad. In addition, the impedance between the electrodes and the skin was 26.0 kΩ (SD = 3.0kΩ), which is comparable to that of commercially available medical bioelectrodes for EMG. When conventional wet electrodes are used for EMG measurement, the electrodes much be attached one by one to the body. The device that we have developed, however, enables measurement simply by attaching a knit textile with many integrated electrodes and wiring to the body and allows measurement over a long period of time, making it highly compatible with long-term EMG monitoring. In actual measurements, motion artifacts did not affect the EMG signal processed with a bandpass filter, and power supply noise could be removed with a notch filter. Finally, our e-textile bioelectrode array could detect the FRP, which is one of the diagnostic criteria for lower back pain. Therefore, by expanding the area of our e-textile bioelectrode array, whole-body EMG can be expected for future healthcare and medical diagnosis.

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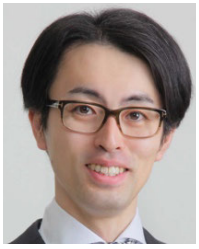

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