Stretchable, Multifunctional Epidermal Sensor Patch for Surface Electromyography and Strain Measurements

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1. Introduction

As a noninvasive technique to record the muscle activities, surface electromyography (sEMG) has been widely used for human–machine interactions (HMIs), prostheses control, and clinical evaluations of muscle function. Nevertheless, the conventional sensor systems made of rigid, bulky components cannot provide a reliable, conformal interface for accurate, continuous measurements of the epidermal physiological signals. Herein, a skin-interfaced, multifunctional epidermal sensor patch with characteristics of mechanical softness, large stretchability, and wearable conformability for multimodal measurements of sEMG signals and associated skin deformations from various muscle activities and joint motions is reported. The sensor patch features two pairs of stretchable sEMG electrodes and two thin, miniaturized strain sensors, which are connected by stretchable filamentary serpentine interconnects in an open-meshed structure. Experimental and computational studies reveal the design and operation of the sensor patch, which exhibit stable and repetitive performance even under a 30% stretching strain level. Demonstrations of the sensor patch on the wrist for simple sign language recognition and on the lower back for the flexion-relaxation phenomenon illustrate its potential for the comprehensive assessment of the muscle activities and related motions of muscle joints.
physiological parameters remains as a challenge due to the lack of mechanically compliant multifunctional sensors that match the physical properties of the skin. The current solutions are mainly limited to the use of separate sensor modules with various sensing functionalities. However, those sensors have bulky form factors, thereby leading to poor data acquisition, undesired irritations, and mechanical constraints to the skin. Only until recently, multifunctional sensor systems that are entirely integrated with ultrathin, stretchable epidermal electronics have made drastic advances\(^\text{[7,8,11,20–22]}\) and exhibit impressive capabilities in noninvasive monitoring for a variety of physiological signals with high fidelity. Of particular interest, sEMG–strain sensor coupled systems that provide a comprehensive assessment of muscle activities and associated movements of muscle joints are gaining much attention due to their applications in sign language or posture recognition,\(^\text{[15]}\) swallowing disorder evaluation,\(^\text{[8]}\) lower back pain (LBP) diagnosis,\(^\text{[7,17]}\) etc. For example, a flexible submental sEMG and strain sensor patch designed with open honeycomb network structures were developed to monitor both the muscle activity and laryngeal movement for rehabilitation of oropharyngeal swallowing disorders.\(^\text{[8]}\)

Although the sensor patch can form a robust interface with the submental muscle area, which generally undergoes relatively minimal deformations, the limited stretchability (< 10%) of the sensor patch is not applicable for the regions like the hand joints or the lower back, which require large strain (> 20%) accommodation.\(^\text{[23]}\)

In this work, we present a stretchable, multifunctional epidermal sensor patch (ES-Patch) that is capable of concurrently measuring EMG signals and associated deformations of the skin from various muscle activities or joint motions. The ES-Patch consists of two pairs of stretchable, skin-interfaced sEMG electrodes and two thin, miniaturized strain sensors, which are connected by stretchable filamentary serpentine interconnects in an open-meshed form. The sEMG electrodes and strain sensors of the ES-Patch are arranged in a symmetrical T-shaped configuration that is capable of covering both muscles and related joints on the hand, submental area, and lower back, respectively. This configuration allows for mechanical softness and large stretchability (≥ 30%) for conformal attachment to complex surfaces of the skin. The mechanical and electrical characteristics of the ES-Patch are determined by experimental and computational studies and are also compared with the conventional sensor designs. Validations of the ES-Patch on the hand for simple sign language recognition and the lower back for the flexion–relaxation phenomenon (FRP) demonstrate the feasibility and effectiveness of the system.

2. Results

Figure 1 shows the basic layouts and configurations of the stretchable, multifunctional ES-Patch that can accommodate large strain during muscle activities and related joint motions for multimodal measurements. As shown in Figure 1a, the epidermal sEMG and strain sensor system features two pairs of stretchable, skin-interfaced sEMG electrodes for recording of the sEMG signals; two thin, miniaturized strain sensors for measuring the skin deformations; stretchable filamentary serpentine interconnects; and contact pads for interfacing with the external data acquisition circuits. The skin-interfaced sEMG electrodes are constructed by the uniform, stretchable serpentine metal traces and are supported by the patterned polyimide (PI) film in the same configuration, as schematically shown in Figure 1b. The associated optical image of the sEMG electrode (in-plane dimensions, 10 mm × 10 mm) is shown in Figure 1c, with the enlarged microscopic image of the serpentine metal traces (Cr/Au, 165 nm in thickness and 60 μm in width) on the right. The exposed serpentine metal traces in an open-meshed structure enable mechanical softness, large stretchability, skin breathability, and can directly contact the skin for the accurate, long-term recording of sEMG signals. The schematic exploded view in Figure 1d and optical images in Figure 1e show the configuration of the thin, miniaturized strain sensor (in-plane dimensions, 2.15 mm × 2.3 mm), which consists of the straight, narrow metal wires (Cr/Au, 55 nm in thickness and 15 μm in width) sandwiched by the patterned polyimide films. The strain sensor is engineered with a perforated polyimide substrate to enhance its sensitivity for strain measurements and the small planar dimensions combined with the island-bridge structure design are used to achieve large stretchability of the epidermal sensor system. Particularly, the sEMG electrodes and strain sensors are connected by the stretchable serpentine interconnects and arranged in a symmetrical T-shaped configuration to match the anatomical structures of various muscles and related joints, such as the submental area and the lower back. The detailed fabrication process of the sensor system is described in Experimental Section. For ease of use and accessibility, the epidermal sEMG and strain sensor system is connected to a flexible anisotropic conductive film (ACF) cable and then integrated with a stretchable medical adhesive tape (Figure S1, Supporting Information) to create a soft, breathable sensor patch for instant use (Figure 1f). The ES-Patch can be directly mounted onto the forearm and maintain intimate contact with the skin even under large mechanical compression, as shown in Figure 1g,h, which demonstrates a reliable, compliant interface for multimodal epidermal data acquisitions. Furthermore, the design features and performances of each sensor in the ES-patch were investigated.

The mechanical design features and dynamic characterization results of the thin, miniaturized strain sensor of the ES-Patch are shown in Figure 2. The design criteria to improve the sensitivity of the strain sensor and avoid the disturbance from the interconnects are that 1) the resistance of the strain sensor should be much larger than that of its interconnects and 2) the resistance change under the mechanical deformation is mainly attributed to the strain sensor rather than its interconnects. Therefore, a thin, perforated strain sensor (resistances: ~2.98 kΩ) connected by stretchable serpentine interconnects (resistance: ~125 Ω) was designed and fabricated by regulating the thickness and width of the metal trace via multiple steps of metal depositions and patterning, respectively. The perforated substrate reduced the rigidity of the strain sensor adhered on the stretchable tape substrate and therefore enhanced its sensitivity to mechanical strains from the deformed skin. As demonstrated by the finite element analysis (FEA) results shown in Figure 2a, the maximum principal strain distributed in the strain sensor based on the perforated polyimide substrate is much larger than that of the conventional design, which is entirely encapsulated by a block of polyimide
film, under a global 30% tensile strain applied to the stretchable tape substrate. The details of the FEA models and the related parameters are shown in Figure S2, Supporting Information. The dynamic responses of the perforated strain sensor to various tensile strain levels were quantitatively characterized. As shown in Figure 2b, a stable and repeatable performance over the strain ranging from 0% to 30% could be observed. The tests were conducted by combining a fatigue testing system (CARE Measurement & Control Co. Ltd.) at a loading frequency of 3 Hz and a multimeter (34465 A, KEYSIGHT) for in situ resistance measurements. Figure 2c shows the normalized resistance changes as functions of the tensile strain for calibrating the strain sensors. The calculated gauge factors from the linear fitting curves are $1.132 \times 10^{-2}$ and $3.16 \times 10^{-3}$ for the perforated and conventional strain sensor, respectively. The higher sensitivity of the perforated strain sensor further validates the rationality of the design. In addition, the stable response of the strain sensor in the cyclic test over 1000 repetitive stretching and releasing cycles at a 30% strain level verifies the mechanical durability of the perforated strain sensor (Figures 2d,e). In addition to validating the design and performance of the strain sensors, the sEMG electrodes were similarly evaluated.

**Figure 3** shows the mechanical and electrical characterizations of the skin-interfaced sEMG electrodes of the ES-Patch. The microstructured serpentine metal traces, which determine the sensing area, mechanical reliability, and wearable comfortability of the sEMG electrodes, are crucial to the fidelity of the measured sEMG signals. Generally, a high filling factor will increase the effective contact area and decrease the contact impedance of the electrodes with the skin but will also yield poor skin breathability and mechanical stretchability. As a compromise, the sEMG electrodes were designed with a filling factor of 66.93%
to enable a large stretchability of 30%. Figure 3a shows the scanning electron microscope (SEM) images and the associated FEA results of the stretchable serpentine metal traces under the given tensile strains of 10% and 20%, respectively. The in situ observations of the deformations were conducted by the stretcher (MT 01714, PHENOM) that is compatible with the SEM instrument (Figure S3a, Supporting Information). The FEA-simulated profiles of the serpentine structure agree well with the experimental measurements. It shows that the maximum principal strain is concentrated in the inner edge of the circular arc and is slightly larger than the elastic strain of Au. It is pointed out that the strain will not cause rapid fracture of the serpentine structure, as demonstrated by a previous theoretical model[25] and by the stable resistance response over 1000 stretching and releasing cycles at a 30% strain level (Figure S4, Supporting Information). The details of the FEA models and the related parameters are shown in Figures S3b,c, Supporting Information. To quantitatively evaluate the signal quality collected by the skin-interfaced electrodes, we compared the sEMG signals of the skin-interfaced electrodes and the standard gel-based electrodes. As shown in Figure 3b, the electrode pairs for double-differential measurements were mounted closely on the forearm. An additional gel-based
electrode was adhered on the back of the hand as the ground electrode. Although the examined areas of the epidermal sensor patch and the gel-based electrodes are different, the muscle activity during the hand-close motion is approximately similar.\textsuperscript{[26]} The sEMG signals were recorded simultaneously during the muscle activity by a home-made wireless, 64-channel sEMG data

Figure 3. Mechanical and electrical characterization of the skin-interfaced sEMG electrodes of the ES-Patch. a) SEM images and associated FEA results of the skin-interfaced sEMG electrodes under the given tensile strain. The scale bar is 200 $\mu$m. b) Optical image of the gel-based electrodes and the stretchable, skin-interfaced sEMG electrodes mounted closely on the forearm. c) The recorded sEMG signals from the gel-based electrodes (G-electrodes) and the stretchable, skin-interfaced electrodes (S-electrodes) of the ES-Patch, respectively. Comparisons of the commonly used features for the measured sEMG signal including d) the frequency spectrum, e) MAV, and f) MPF. g) The measured sEMG signal of the ES-Patch from the hand-close gesture after being worn for 17 h on the forearm and the electrode response under dynamic large compression during sEMG measurement.
acquisition system. Figure 3c shows the measured sEMG signals from the gel-based electrodes and the stretchable electrodes, respectively, for the two repetitive hand-close motions each with a duration of 10 s. A few of the frequently used features for sEMG signal analysis including the frequency spectrum, the mean absolute value (MAV), and the mean power frequency (MPF) are compared and shown in Figure 3d,e,f, respectively, which demonstrates the excellent performance of the skin-interfaced electrodes for the sEMG measurements.

In addition, the adhesion of the electrode on the skin is critically important for the measurement of the sEMG signal. To examine the adhesion strength and durability of the stretchable tape, the energy release rates of the stretchable tape with the PMMA plastic and the forearm skin were measured by the 90° peeling test (Figure S5, Supporting Information). It is shown that the adhesion strength of the tape is sufficient and is comparable with that of the commonly used silicone-based adhesives. Importantly, the adhesive of the tape is applicable for repetitive usage (>20 times). Furthermore, we attached the ES-Patch on the forearm and conducted an in situ compression of the skin, which would cause local bending or wrinkling of the ES-Patch (Figure S6a-i, Supporting Information). After the ES-Patch was compressed for 1000 times on the forearm, the adhesion of the ES-Patch was still strong enough to sustain intimate contact with the skin (Figure S6a-ii, Supporting Information), as demonstrated by the measured sEMG signals in Figure S6b and Movie S1, Supporting Information. The measured sEMG signal was not disturbed by the external large, dynamic compressive deformation of the skin. To investigate the skin response, adhesion, and electrode dehydration after long-term wearing, the ES-Patch was worn on the forearm for more than 17 h, including an 8 h sleep session and 1 h bicycle ride. The adhesion of the ES-Patch was sufficient for retaining close contact with the skin (Figure S7-i, Supporting Information), even under a large compression of the skin (Figure S7-ii, Supporting Information). After removing the ES-Patch, a little bit of sweat and a slight wrinkling of the skin was visible but no irritation was observed (Figure S7-iii, Supporting Information). The skin recovered to its natural state soon after the removal of the ES-Patch for 10 min (Figure S7-iv, Supporting Information). The measured water vapor transmission rates (WVT) of the stretchable tape and its TPU film at 38 °C and 90% relative humidity are 9.652 g m⁻² h⁻¹ and 58.33 m⁻² h⁻¹ (W3/030, Labthink), respectively. The low WVT of the stretchable tape reduces the skin breathability of the ES-Patch and leads to the visible sweat shown in Figure S7-iii, Supporting Information, which can be improved by the microperforated design of the silicone gel adhesive of the stretchable tape. In addition, we also measured the sEMG signal after long-term wearing of the ES-Patch, as shown in Figure 3g and Movie S2, Supporting Information. During the measurement, the ES-Patch was adhered on the forearm and underwent external large, dynamic compression deformation of the skin. The recorded data in Figure 3g indicate that the dynamic deformation of the skin has a little effect on the measured sEMG signal.

The applications of the stretchable and multifunctional ES-Patch for the comprehensive assessment of muscle activities and associated motion of muscle joints are shown in Figure 4. The first example involves a simple demonstration of sign language recognition, which is critical for wearable HMI applications such as myoelectric prostheses control, sign-to-speech translation, and virtual reality. Devices for decoding the sign language have been developed and are mainly based on sEMG electrodes and piezoresistive strain sensors, respectively, but a combination of sEMG and strain sensors has been rarely investigated. Figure 4a shows the optical image of the ES-Patch adhered on the back of the hand, where muscle activities are highly coincided with the joint motions from various gestures. Here, the strain sensor of the ES-Patch is considered as the additional signal feature for the sEMG-based sign language recognition, and offers a simple yet robust method to improve the recognition accuracy, instead of using additional training and enhanced classification algorithms. As shown in Figure 4b, although the sEMG signals are similar in both the waveform and amplitude for the two distinct gestures, that is, hand close and wrist flexion, the difference in the concurrently collected strain signals is easily distinguishable, without the need for any complex algorithms. Figure 8, Supporting Information, shows the optical image of the custom-built data acquisition system for the ES-Patch, which consists of a wireless, 64-channel sEMG data acquisition system and a multimeter.

Another potential application is the use of the ES-Patch to detect the muscle activities of the lower back (Figure 4c). Chronic nonspecific low back pain (CNLBP) is one of the most significant health problems worldwide and substantially affects individuals due to the high prevalence and cost. The FRP, which represents a well-studied neuromuscular activity of the lumbar spine during trunk flexion, has been used as an important indicator clinically to distinguish CNLBP patients from healthy individuals. Figure 4d shows the schematic cartoon snapshots to illustrate the continuous postures of the subject during the experimental investigations of the FRP including I) stand straight, II) begin to stoop, IV) full trunk flexion, and V–VII) recover to the initial state. At the beginning stage of the trunk flexion (stage I–III), the muscles of the lower back continue to contract and the sEMG signal will increase. As the subject continues to stoop forward to the full trunk flexion state (stage III–IV), the muscles eventually relax and the sEMG signal decreases to a resting level, that is, myoelectric silence. During the recovery to the initial state (stage V–VII), the muscle activity of the lower back will be immediately observed again. Interestingly, the FRP is common in healthy people without CNLBP, but in patients with the CNLBP, this phenomenon is usually absent. Figure 4e shows the sEMG and strain signals of the lower back of a healthy subject (Male, 28 years old) captured by the ES-Patch during the trunk flexion, as shown in Figure 4d. Initially, the strain signal was invisible but it coincided with a small sEMG signal when the subject stood straight (stage I). As the subject started to stoop, both the sEMG and strain signals increased slightly. When the subject continued to stoop forward to reach the full trunk flexion, the sEMG signal decreased to a resting state but the strain signal increased rapidly (stage I–IV). Then, the subject held the posture for ~3 s and both the strain and sEMG basically remained unchanged (stage IV). Finally, the instant response of the muscle activity was captured at the onset of the subject recovery to the initial stage though the strain signal
decreased gradually. In contrast, for the patient with CNLBP, the measured sEMG signal was totally different, as shown in Figure 4f. The patient (female, 65 years old) with CNLBP was recruited from the Zhejiang Provincial Hospital of Chinese Medicine. When the patient reached the full trunk flexion (stage IV), the muscles of the lower back still continued to contract and could not relax, which yielded a continuous sEMG signal during the whole measurement. These results demonstrate the feasibility of the ES-Patch to capture the FRP and provide a simple yet effective tool for the clinical diagnosis of CNLBP.

Figure 4. Comprehensive assessment of the muscle activities and associated motion of the skin or muscle joints on the wrist and the lower back via the stretchable, multifunctional ES-Patch. a) Optical image of the ES-Patch mounted on the back of the hand. b) The measured sEMG (from the left pair of the electrodes) and strain signals (from the lower one) for the two distinct gestures (hand close and wrist flexion). c) Optical image of the ES-Patch mounted on the lower back of a healthy subject. d) Schematic cartoon snapshots to illustrate the continuous postures of the subject during FRP investigation: I) stand straight, II–III) begin to stoop, IV) full trunk flexion, V–VII) recover to the initial state. The measured sEMG (from the left pair of the electrodes) and strain signals (from the upper one) from e) a health subject and f) from a patient with CNLBP.
3. Conclusion

In summary, we present an easy-to-use, skin-mountable sensor patch that is applicable for multimodal measurements of sEMG and strain signals during muscle activities and related joint motions. Applications of the epidermal sensor patch were demonstrated on the wrist for strain-assisted sign language recognition and the lower back for the FRP. The sensor patch possesses excellent mechanical softness and stretchability, skin breathability, and exhibits stable and repeatable responses even under the 30% strain level. The results reported here provide a significant foundation for the construction of stretchable, multifunctional epidermal sensor systems and create engineering opportunities for widespread applications such as wearable HMIs, health-monitoring devices, and clinical evaluations of muscle function.

4. Experimental Section

Fabrication of the sEMG and Strain Epidermal Sensor System: To prepare the sEMG and strain sensor system, polyimide precursor (ZKPI 305IIE, POME) was first spin coated on a cleaned glass slide at 3000 rpm for 60 s and then baked on a hotplate at 80 °C for 60 min, 110 °C for 60 min, and 230 °C for 120 min, to yield the polyimide-supporting layer. Thereafter, Cr/Au (5 nm/50 nm) metal layer was deposited by e-beam evaporation and patterned by the photolithography and lift-off process to form the strain sensors on the polyimide-supporting substrate. Next, another layer of Cr/Au (5 nm/160 nm) metal layer was deposited on the polyimide supporting substrate by following the same process to form sEMG electrodes and stretchable interconnects. Finally, polyimide precursor (ZKPI 305IIE, POME) was spin coated on the device at 4000 rpm for 60 s and baked to form the polyimide encapsulating layer. The open-meshed configurations of the bilayer polyimide films were defined by inductively coupled plasma (ICP) etching with patterned Al as the mask. The device was released from the glass slide by entire immersion in buffered oxide etch (BOE).

Fabrication of the Epidermal Sensor Patch: The released epidermal sensor was first cleaned in deionized water and held with a temporary glass substrate. The ACF cable was connected to the pads of the sensor and printed circuit board (PCB) by heating and pressing. Then the device was flipped over in deionized water and held again with a temporary glass substrate. Finally, the device was adhered on the stretchable transparent medical tape with the dimensions defined by a laser cutting machine to yield the epidermal sensor patch for ease of use.

Characterization of the ES-Patch: For the measurement of the cyclic resistance change of the strain sensor under the repetitive stretching and releasing strain ranging from 0% to 30%, the strain sensor was adhered on the stretchable soft tape substrate (2.5 cm × 6 cm) and then baked at 60 °C for 60 min, and 230 °C for 120 min, to yield the epidermal sensor patch for ease of use.

Conflict of Interest

The authors declare no conflict of interest.

Data Availability Statement

Data available on reasonable request from the authors.

Keywords

electromyography, epidermal sensor patches, multimodals, strain sensors

Supporting Information

Supporting Information is available from the Wiley Online Library or from the author.

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