Development of 3D Textile Electrodes for Electrocardiography Measurement

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***Abstract* — Flexible dry electrodes are essential for wearable biosignal monitoring, but conventional textile-based designs often suffer from high impedance and unstable skin contact. In this study, a 3D-structured textile electrode was fabricated by embroidering conductive threads onto polydimethylsiloxane (PDMS) substrates using a computerized embroidery technique. Compared to conventional flat textile electrodes, the proposed electrode exhibited an 80.6% lower impedance at 10 Hz and a 5.42 dB higher signal-to-noise ratio (SNR). During dynamic measurements, the proposed electrode maintained signal fidelity, achieving a Pearson correlation of 0.893 with reference Ag/AgCl electrodes. These results demonstrate that the proposed method effectively improves signal quality and motion stability in dry electrodes for wearable electrocardiography (ECG) monitoring.**

I. Introduction

In response to societal trends such as aging populations and increasing healthcare costs, wearable devices for daily health monitoring are gaining significant attention. This shift underscores the growing importance of non-invasive electrodes for biosignal acquisition. Traditionally, wet Ag/AgCl electrodes have served as the clinical and commercial standard due to their high conductivity and stable signal-to-noise ratio (SNR). However, their prolonged use presents challenges, including gel drying, poor breathability, signal degradation, and skin irritation.

As a result, recent research has focused on the development of alternative electrode technologies. Among them, conductive fiber or textile-based electrodes have emerged as promising alternatives, offering advantages such as flexibility, lightweight form factor, breathability, mechanical compliance, and seamless integration into clothing. Textile electrodes fabricated via knitting [1], embroidery [2], and coating [3] have demonstrated feasibility for wearable biosignal monitoring and exhibit excellent durability under repeated washing and bending conditions [4]. Nonetheless, these face key limitations, namely high skin-electrode impedance and the requirement for consistent, firm contact with the skin to maintain signal quality [5].

To address these challenges, previous studies have explored various strategies, such as incorporating microstructures or surface patterns to increase effective contact area and reduce impedance [6] or utilizing 3D structures to enhance contact pressure and ensure sustained skin contact [7]. Textile electrodes have similarly adopted techniques such as towel embroidery [8] and electrostatic flocking [9], though these methods often struggle with precise control over complex 3D geometries and involve high fabrication complexity and cost.

In this study, we propose a novel fabrication method for textile electrodes using conductive threads embroidered onto polydimethylsiloxane (PDMS) substrates via a 3D embroidery technique. The effectiveness of the proposed design was validated through skin–electrode impedance measurements and electrocardiogram (ECG) signal acquisition. Experimental results demonstrated that the 3D-structured electrodes lead to improved ECG signal fidelity under both static and dynamic conditions. These findings suggest that the proposed fabrication method offers a promising solution for wearable ECG monitoring systems requiring both low-impedance and high-quality signal acquisition in real-world environments.

II. Method

A. *Fabrication of 3D Textile Electrodes*

To construct textile electrodes optimized for high-fidelity ECG acquisition, we employed a 3D embroidery technique using commercial silver-coated polyamide conductive thread (HC 40-Madeira, 300 Ω/m). PDMS substrates were fabricated via a casting method using molds (3 × 3 cm, 1 mm thickness) produced by 3D printing. To reduce needle rebound during embroidery, PDMS (Sylgard 184, Dow Corning) was prepared with an elevated curing agent ratio (100:11), vacuum-degassed for 2 hours, and then cured at 40 °C for 24 hours.

The resulting PDMS substrates were placed on cotton fabric, and conductive thread was embroidered in a 3 cm diameter circular pattern designed using Ink/Stitch software. Two types of electrodes were fabricated for comparative evaluation: (1) Planar and (2) Flat (conventional embroidery without PDMS).

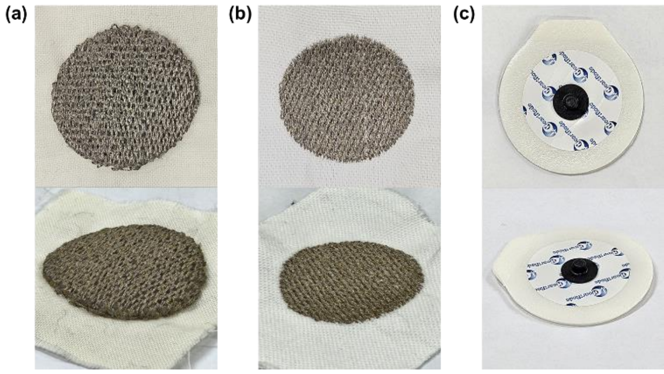

Fig. 1 (a) Fabricated Planar electrode, (b) Fabricated Flat electrode, (c) Commercial Ag/AgCl (gel type) electrode.

B. Skin–Electrode Impedance Measurement

Skin–electrode impedance was measured on the participants' forearms using a bipolar configuration. A commercial Ag/AgCl reference electrode was placed at the elbow, while the textile electrodes under evaluation were positioned on the forearm. To ensure consistent and uniform contact pressure across measurements, an elastic arm sleeve was applied over the electrodes. Impedance values were recorded across a frequency range of 10 Hz to 10 kHz using a precision LCR meter (Gwinstek LCR-6100).

C. Electrocardiogram Measurement

ECG signal quality was assessed using a commercial biosignal acquisition system (BioBrain Bios-s40) with a sampling rate of 250 Hz. A standard Lead I configuration was employed, with signal electrodes placed on the left arm (LA) and right arm (RA), and the reference electrode positioned on the lower left abdomen (LL; commercial Ag/AgCl electrode). To enable direct performance comparison, two electrode sets were simultaneously attached during measurement: the test electrodes (Planar and Flat types) applied to the left wrist, and the commercial Ag/AgCl electrodes applied under identical conditions.

ECG signals were recorded for 3 minutes under two conditions: static and dynamic (induced by repeated sitting and standing). For analysis, the first and last 10 seconds were excluded, and the central 2 minutes and 40 seconds of data were used. All signals were preprocessed using a 60 Hz notch filter and a 0.5–30 Hz bandpass filter to eliminate line noise and irrelevant frequency components.

R-peaks were detected using a local maximum-based algorithm with a minimum inter-peak interval of 500 ms to prevent false detection of R-peaks. For each detected R-peak, a beat segment was extracted from –150 ms to +350 ms relative to the R-peak. The mean waveform of all extracted beats was defined as the signal, and the difference between each individual beat and the mean waveform was defined as the noise. Based on these definitions, the SNR was calculated as the ratio of signal power to noise power.

Additionally, Pearson correlation analysis was conducted between beat segments extracted from the textile electrodes and those from the concurrently measured Ag/AgCl electrodes, using the latter as the reference, to assess how closely the textile electrodes reproduced the standard signal waveforms under dynamic conditions.

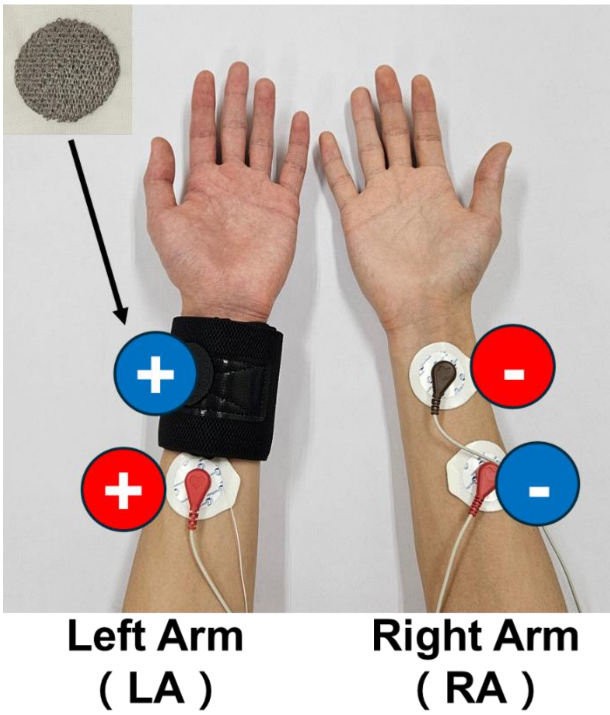

Fig. 2 Electrode placement for ECG measurement.

III. Results

A. Skin–Electrode Impedance

Fig. 3 presents the skin–electrode impedance measurements for the three electrode types, revealing clear differences across the entire frequency range. In general, lower skin–electrode impedance is considered to enable higher quality signal acquisition during biosignal monitoring [5].

As textile electrodes are dry electrodes, higher skin–electrode impedance was observed compared to conventional wet Ag/AgCl electrodes. However, when the proposed 3D embroidery fabrication method was applied, a noticeable reduction in impedance was achieved.

At 10 Hz, a frequency band particularly important for physiological signal measurement, the Planar electrode exhibited an impedance of 2.69 MΩ, which represents a reduction of 80.57% compared to the Flat electrode (13.86 MΩ).

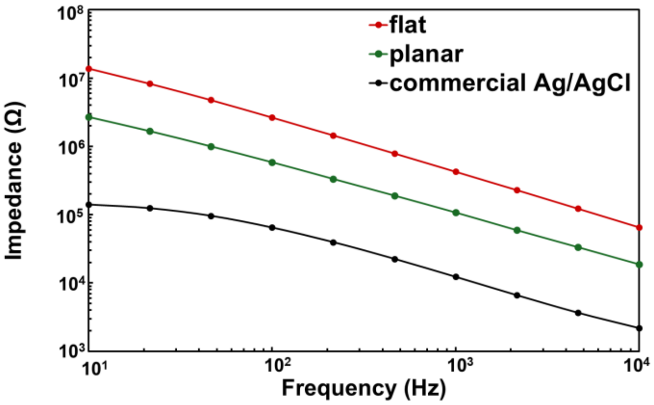

Fig. 3 Skin-electrode impedance results.

B. *Electrocardiography*

Fig. 4 presents the visualization and SNR analysis of the measured ECG signals. In the mean waveform analysis, the Planar electrode captured more distinct ECG features than the Flat electrode. A higher SNR indicates a cleaner signal with less noise. Based on this metric, the Planar electrode exhibited a higher SNR value of 9.92 dB compared to the Flat electrode at 4.5 dB. When compared with signals obtained using commercial Ag/AgCl electrodes as the reference, the Planar electrode showed a smaller difference in SNR—5.7 dB under static conditions and 11.41 dB under dynamic conditions—than the Flat electrode. These results indicate that the Planar electrode achieved quantitatively improved performance compared to the conventional textile electrode.

Fig. 5 presents the results of a Pearson correlation analysis between signals recorded with the textile electrodes and those acquired using commercial Ag/AgCl electrodes under dynamic conditions. Based on the average Pearson correlation coefficient calculated from all extracted individual beats, the Planar electrode achieved a mean correlation of 0.893, which was higher than that of the Flat electrode (0.642). These results suggest that the structured surface of the Planar electrode effectively attenuated motion artifacts, resulting in ECG waveforms more similar to those of commercial Ag/AgCl electrodes.

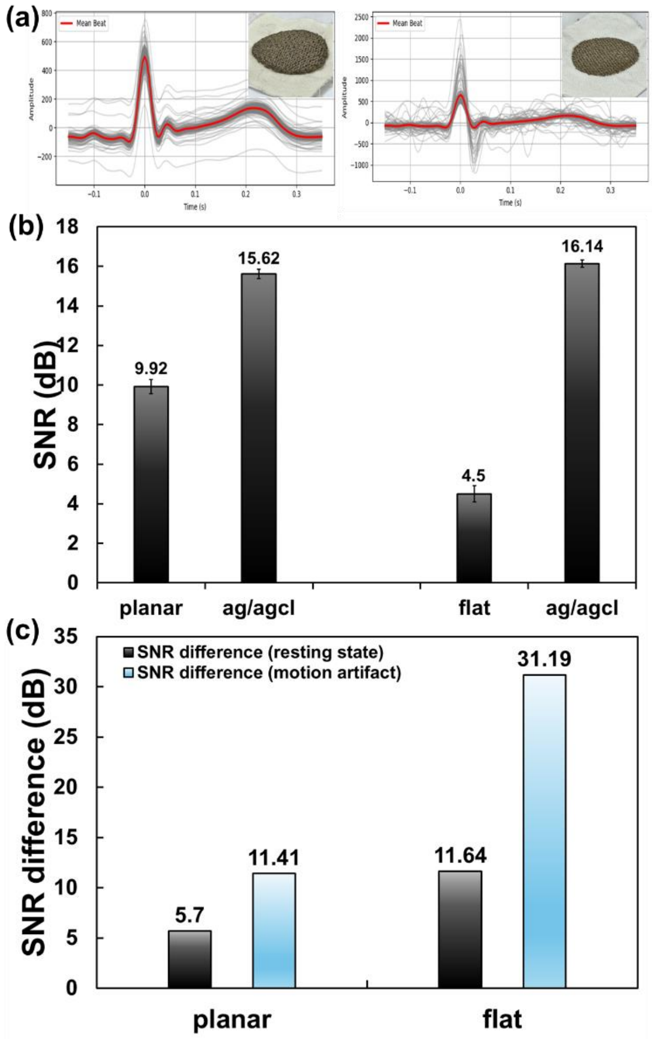

Fig. 4 (a) Mean ECG waveforms recorded using Planar and Flat electrodes, (b) SNR results, (c) SNR difference analysis results.

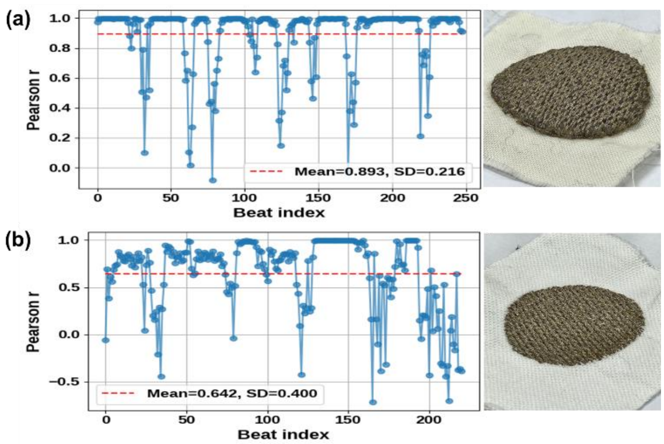

Fig. 5 Pearson correlation results, (a) Planar electrode, (b) Flat electrode.

IV. DISCUSSION

In this study, we proposed a flexible textile electrode fabricated by 3D embroidery onto PDMS substrates to address the limitations of traditional textile electrodes, such as incomplete skin contact and poor signal quality. The skin–electrode impedance analysis showed that the 3D textile electrodes exhibited lower impedance than conventional textile electrodes. These improvements can be attributed to the presence of 3D surface structures, which have been reported to enhance skin contact pressure and reduce impedance by increasing the effective contact area [9].

Compared to traditional textile electrodes, the proposed 3D textile electrodes also demonstrated superior signal performance, including a higher SNR and a stronger similarity to heartbeat waveforms recorded by commercial Ag/AgCl electrodes. Even under conditions involving periodic body movement, the proposed electrodes maintained clear signal quality and exhibited enhanced robustness against motion artifacts. Although the improved contact conditions are assumed to contribute to signal quality, the actual increase in contact area was not directly validated in this study. Therefore, further investigations are required to quantitatively assess the pressure and contact characteristics using artificial skin, human phantoms, or real skin models.

V. CONCLUSIONS

In this study, a flexible dry electrode with a 3D structure was fabricated by integrating PDMS substrates with computerized embroidery. Unlike traditional textile electrodes, the proposed 3D structure enabled more stable skin–electrode contact, resulting in reduced impedance and enhanced signal quality. The electrode also demonstrated improved robustness to motion artifacts compared to conventional textile electrodes, suggesting its potential applicability in ambulatory biosignal monitoring. Although the precise contribution of increased contact area has not been quantitatively verified, the overall findings support the effectiveness of this structural approach for enhancing dry electrode performance.

To further improve the electrode's functionality and adaptability, future work will investigate the design of various 3D geometries beyond simple planar forms by modifying the PDMS curing mold. In addition, introducing porosity into the PDMS substrate will be considered to enhance material properties such as moisture permeability and wearing comfort. These efforts, along with systematic evaluation of long-term usability, user comfort, and integration into wearable bioelectronic platforms, will help expand the applicability of the proposed technique to other electrophysiological signals such as Electromyography and Electroencephalography.

ACKNOWLEDGMENT

This work was supported by National Research Foundation of Korea (NRF) grants funded by the Korean government (RS-2024-00355530).

This research was supported by the MSIT(Ministry of Science and ICT), Korea, under the ITRC(Information Technology Research Center) support program (IITP-2025-RS-2023-00258971) supervised by the IITP (Institute for Information & Communications Technology Planning & Evaluation).

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