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ORIGINAL ARTICLE

Conditions for Textile Electrode Sensors to Monitor Cardiac Activity in Daily Life

Jae Kwan Kim[1,2] · Sangin Park[1] · Hyun-Seung Cho[3] · Jin-Hee Yang[3] · Soo-Hong Lee[2] · Jeong-Whan Lee[4]

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Abstract

This study determined the conditions for textile electrode sensors combined with a garment-type wearable platform to monitor daily cardiac activity. Six types of textile sensors were implemented by manipulating the size and configuration of textile electrodes. Each sensor was attached to a chest band to fabricate experimental garments, which were sequentially worn by ten healthy males. Their electrocardiogram (ECG) signals were measured using the modified Lead II method. We sampled ECG signals at 1 kHz using a Biopac ECG100 device. The signal power ratio of the ECG signals acquired from each electrode was calculated, and statistical analysis was performed. The results indicate that convex electrodes detect higher quality signals than flat electrodes. However, no significant difference was observed in the performance of cardiac activity acquisition in terms of electrode size. These results validate that the type of configuration, namely flat or convex, significantly impacts the performance of cardiac activity signal acquisition. Additionally, correlation coefficients, mean error, and Bland–Altman plots revealed a strong positive correlation, low mean error, and adequate agreement, respectively, between ECG and all textile sensors. The study's findings form the basis for implementing a wearable platform that integrates textile sensors with garments.

Keywords Textile electrode sensor · Cardiac activity · Wearable platform · Smart healthcare

1 Introduction

In the healthcare market, a paradigm shift from therapeutic to preventive and predictive purposes is accelerating to satisfy the growing medical needs of aging societies and the increasing number of chronically ill patients. In addition, there is a growing demand for smart wearable devices that can detect vital signs without interfering with users' voluntary actions at work, in sports activities, at home, or in clinical environments [1]. Therefore, personalized smart healthcare has become highly significant. Smart healthcare involves generating and collecting healthcare data using ICT or biotechnology. These healthcare data can be classified into genetic information, personal health records (PHRs), and electronic medical records. Among them, PHRs are the data obtained from the daily life activities of a person, including their blood sugar level, blood pressure, electrocardiogram (ECG), amount of exercise, and diet information. These data are primarily generated and collected through IoT devices and smartphone applications [2, 3].

In general, cardiac activity is an important metric for body condition monitoring, particularly in terms of preventive treatment in sports as well as general medical treatment

✉ Soo-Hong Lee
shlee@yonsei.ac.kr

✉ Jeong-Whan Lee
jwlee95@kku.ac.kr

Jae Kwan Kim
kimjk@kist.re.kr

Sangin Park
sipark@kist.re.kr

Hyun-Seung Cho
hyunseung-cho@yonsei.ac.kr

Jin-Hee Yang
sjinnie7@yonsei.ac.kr

1 Bionics Research Center, Korea Institute of Science and Technology, Seoul 02792, Korea

2 School of Mechanical Engineering, Yonsei University, Seoul 03722, Korea

3 Institute of Symbiotic Life-TECH, Yonsei University, Seoul 03722, Korea

4 Department of Biomedical Engineering, Konkuk University, Chungju 27478, Korea

[4]. Among the various biometric information that can be measured from the human body, ECG signals reflect the heart's activity state depending on activity conditions and the occurrence of lesions. Therefore, if ECG signals can be detected in an unconstrained and non-intrusive manner during daily activities, they can serve as essential PHR data.

However, it is practically difficult to attach the 12 clinical-standard electrode leads to the entire body for 24 h to detect ECG signals in daily life [5–13]. In addition, monitoring using conventional sensor technology often causes many problems; for example, it has been reported that skin irritation and allergic contact dermatitis can be caused by the adhesive parts and gels used in conventional ECG sensors [14]. The ultimate aim of wearable sensor technology is to enable continuous monitoring of human physiological responses while minimizing the burden of sensor attachment [15]. Accordingly, we assumed that it is necessary to embed electrodes in a wearable platform to achieve smart healthcare and detect ECG signals in an unconstrained and non-intrusive manner in daily life. Furthermore, we focused on the results of previous studies [6, 11–13, 16–20] that investigated the types, configurations, and locations of electrodes, which significantly impact the efficiency of obtaining ECG signals through a wearable platform. Particularly, the effect of the contact textile electrode configuration on the cardiac activity signal acquisition was analyzed using six types of contact textile electrodes based on the configuration conditions of the electrodes. Furthermore, by comparing the cardiac activity signal sensing performance of each electrode condition, we determined the conditions of each type of textile electrode that is most suitable for cardiac activity signal acquisition. In addition, noise from muscle movement should be considered as a significant factor affecting the reliability of cardiac activity measurement; however, previous studies related to developing textile sensors have not examined this point in depth. Thus, the aim of this study was to determine the conditions for textile electrode sensors on a garment-type wearable platform to monitor daily cardiac activity while considering muscle movement conditions.

2 Background

2.1 Wearable Cardiac Activity Signal Sensing Technology

A typical wearable cardiac activity signal sensing platform comprises a textile electrode unit corresponding to the input interface, textile signal wire unit, device unit, and wearable platform on which they are mounted. In the early 2000s, research on wearable cardiac activity signal sensing technology involved attaching various small devices and wires to the conventional band- or garment-type platforms. Concurrently, the development of electrodes and signal wires for wearable cardiac activity signal acquisition started using carbon fibers and metallic yarns. After 2006, electrodes and signal wires for sensing ECG signals began to be developed in textile forms using different materials, such as silver- or copper-based metallic yarns, carbon fibers, and conductive polymers. Wearable cardiac activity signal sensing technology advanced further from the beginning of 2010, with textile electrodes and signal wires being developed for sensing ECG signals using graphene, silver nanowires, different conductive polymers, and silver nano-material-based metallic yarns. Moreover, research and development on system-on-textile technology began to produce conventional control unit-based electrical circuits in the form of textiles [7, 8, 10, 21–25].

2.2 Textile Electrodes for Cardiac Activity Signal Sensing

In the clinical field, disposable Ag/AgCl metallic electrodes are used to measure ECG signals. However, allergic reactions may occur when these electrodes are attached to the human body surface using an electrolyte gel. Therefore, they cannot be used for a prolonged period as they are easily oxidized. To address this, textile electrodes that can replace Ag/AgCl metallic electrodes have been developed in the field of wearables; these textile electrodes can be attached to the human body for a prolonged period to monitor cardiac activity signals [6–13, 18–22]. Typically, textile electrodes are based on resistance, piezoelectric voltage, capacitance, and inductive capacity measurement methods. They are usually categorized depending on whether they are in contact with the human body.

Contact-type textile electrodes are the most common type used for wearable cardiac activity sensing. They acquire ECG signals only when the skin and electrodes are in contact. Their limitation is that the signal acquisition terminates or measurement noise occurs if a gap exists between the skin and an electrode or the position of the contact point changes owing to the movement of the wearer. ECG signals measured using contact-type textile electrodes are generally based on the resistive or piezoelectric measurement principle.

Song et al. [7], who developed contact-type textile electrodes, designed them in the jacquard weaving style using silver-based metallic yarn to measure ECG signals. They fabricated four types of electrodes based on each combination of electrode configuration (convex and flat types) and conductive paste application (with and without) and compared the cardiac activity signal sensing performance of the four electrodes. They determined that the sensing performance was best when the ECG signals were measured using a convex-type electrode coated with a conductive paste. Cho et al. [8] developed silver yarn-embroidered textile electrodes to evaluate the ECG sensing performance of four types of biometric signal sensing smart garments (chest-belt-type, cross-type, X-type, and curved-X-type). They measured the displacement of the electrodes in the standstill and moving states using ECG measurement experiments and analyzed the ECG signals. They reported that the "cross-type" garment was the most effective for ECG measurement. Cho and Lee [13] examined the cardiac activity signal sensing performance considering various induction positions of the contact-type cardiac activity sensing electrodes. The results indicate that the parts on the garment where the detection rate of valid signals was the highest, signal size was large, and signal quality was the best despite the movement of the body were between the vertical and sidelines of the nipples, below the chest circumference line.

One measurement principle of the non-contact-type textile electrode for cardiac activity signal sensing involves a magnetic-induced conductivity-based heart rate sensing method. This method is relatively less affected by motion compared to the electric field-based heart rate sensing method. Herein, the measurement principle is that when a time-varying magnetic field is applied to stimulate tissues in the body (e.g., muscles), eddy currents are generated and the re-induced magnetic field triggers a change in the inductance of the excitation coil, which varies the oscillation frequency [26].

Gi et al. [18] developed a textile-based inductor-type sensor structure using a computer embroidery method and compared the obtained mechanical ECG signals with Lead II ECG signals. The experimental results indicate that the textile-based inductor-type sensor is feasible for measuring ECG signals, as it exhibited a high correlation with the corresponding signals. Koo et al. [17] analyzed the effect of a textile-based inductive coil sensor on the heart rate measurement. They explored eight potential positions of the inductive coil sensor through a pilot test and selected three sensor positions. The result verifies that the position of the textile-based inductive coil sensor affects the quality of measurement result significantly. Gi et al. [19] investigated the effects of the position and shape of the electrode on the cardiac activity signal acquisition in magnetic-induced conductivity-based cardiac activity sensing. They reported that the dual-electrode type of the oval–oval shape is better suited for comprehensively overcoming the individual differences in the heart location.

3 Methods

3.1 Participants

Ten healthy males aged between 22 and 29 years (mean 25.2 ± 3.4) participated in the experiment of this study. Their average body shape was suitable for the cardiac activity measurement experiments. All participants were right-handed and had no medical or family history of cardiovascular and autonomic nervous system disorders. Additionally, the participants were required to abstain from alcohol, cigarettes, and caffeine for 24 h prior to the experiment. A consent form was signed by each participant to comply with the research ethics regulations, and the experimental protocol for measuring cardiac activity was explained to all participants.

3.2 Implementation of Textile Electrodes

The potential difference generated by the electrochemical reaction occurring in the boundary layer between the human body and electrode is referred to as the electrode impedance, which serves as electrical noise in the signal acquisition. When the electrode comes in contact with the stratum corneum of the skin, the contact impedance decreases with the increase in the contact surface area. In this study, we increased the stickiness between the skin and electrode using the structure of the electrode to increase the contact surface area. This in turn reduced the electrical noise, and high-quality signals were obtained.

The structure of the electrode was divided into variables of two aspects, namely the size and configuration (flat or convex). The dimensions of the electrodes implemented were 1×1, 2×2, and 3×3 cm. In terms of the configuration, we implemented satin-embroidered three-dimensional (3D) convex electrodes by inserting a piece of conductive rubber in the back of the electrode, whereas the flat electrodes were embroidered in a flat shape using a common satin embroidery method. Six types of cardiac activity sensing electrodes were fabricated by combining the aforementioned three sizes and two configurations. The 280-denier conductive yarn (PE-Silver plated Nickel hybrid yarn) was used as the electrode material, which was computer-embroidered on nylon fabric. To obtain the ECG signals using the modified Lead II method, we fabricated six types of triangular induction electrode sets, each including one ground (RLD) electrode and two induction electrodes (RA, LL), as shown in Fig. 1. Although the configurations of the electrodes differed in the electrode sets, the distance between the electrode centers was identical, and the length of the wire was maintained constant.

3.3 Experimental Environment and Protocol

Figure 2(a) depicts the positions (V1 to V6) of the clinical chest induction electrodes. The electric field-based contact textile electrodes for conventional cardiac activity monitoring are positioned between V3 and V4 [8]. The six induction electrode sets were attached to the same chest

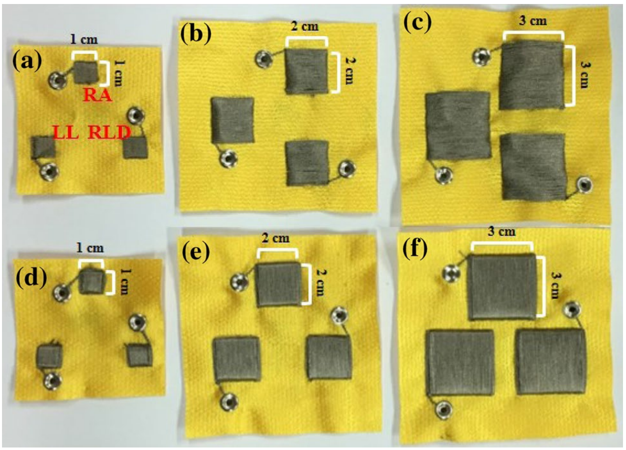

Fig. 1 Six types of textile electrodes fabricated using the modified Lead II method. **a** flat electrode (1 × 1 cm); **b** flat electrode (2 × 2 cm); **c** flat electrode (3 × 3 cm); **d** convex electrode (1 × 1 cm); **e** convex electrode (2 × 2 cm); and **f** convex electrode (3 × 3 cm)

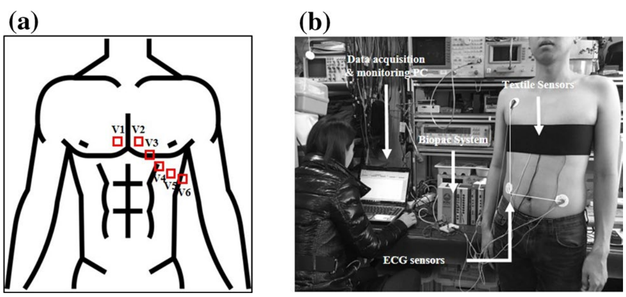

Fig. 2 **a** Positions of six leads of ECG and **b** the experimental setup of cardiac activity measurement

band, and all participants wore the chest band such that the electrodes were in contact with the same positions on the body. In other words, the center of RA in the modified Lead II method was positioned horizontally on the nipple line, whereas the center of RLD was positioned on the centerline of the body. Furthermore, to verify the reliability of the ECG signals obtained using the electrodes employed in this study, the clinical Ag/AgCl electrodes were attached to the body using the standard Lead II method. The corresponding ECG signals were simultaneously measured and used as reference signals. In the Ag/AgCl electrodes, RA, LL, and RLD were attached to the right shoulder, lower left of the torso, and lower right of the torso, respectively. The ECG signals of all participants were measured when standing still and quickly walking in place for 20 s; they were continuously measured twice for each textile electrode type (Fig. 2b). Subjects were required to limit their movements during the experiment.

3.4 Data Acquisition and Signal Processing

The experimentally detected ECG signals were measured at a sampling rate of 1 kHz using a Biopac system (Biopac MP150, AcqKnowledge; Biopac System Inc., Goleta, CA, USA) and amplified using a Biopac ECG100 (Biopac Systems Inc., Goleta, CA, USA). The measured raw signals were filtered using a bandpass filter (Butterworth bandpass: order = 10, passband: 5–15 Hz), and the ECG spectrum was analyzed using a fast Fourier transform (FFT) method in the range of 1–500 Hz. We calculated the signal power ratio (SPR) of the ECG signals obtained through each electrode using Eq. (1) to perform a quantitative comparison and analysis of cardiac activity acquisition performance based on the structure of textile electrodes. All signal processing and data analyses were performed using EEGlab, a MATLAB toolbox (2020b, Mathworks Inc., Natick, MA, USA).

$$SPR = log\left(\frac{\sum_{f=5Hz}^{15Hz} P_{signal}}{\sum_{f=1Hz}^{500Hz} P_{signal}}\right) \quad (1)$$

In other words, we calculated the ratio of the signal power of the entire signal in the 5–15 Hz band, where the ECG signals are distributed as SPR. Based on this, the ECG signal acquisition results were normalized to control the intervention of the noise variable generated by the difference in the unique characteristics of each participant. Additionally, a statistical analysis was performed for the calculated SPR, and the performance difference of cardiac activity acquisition was comparatively investigated for the six textile electrodes.

4 Results

4.1 Participants' Samples

The experimental results were displayed using AcqKnowledge 4.2, and the waveforms of the acquired ECG signals were qualitatively compared and analyzed. Figure 3 depicts the parts of the ECG signal waveforms of Subject 1. The waves at the top portion of the graph are the reference signals detected from the clinical Ag/AgCl electrodes, whereas those at the bottom are the ECG signals detected from the fabricated textile electrodes. In terms of signal size, Fig. 3 indicates that the size of the signals detected from the textile electrodes is generally smaller than that of the reference signals.

In the case of convex electrodes, all three sizes exhibited highly stable signal waveforms, and the peak points of the detected and reference signals concurred well. Conversely,

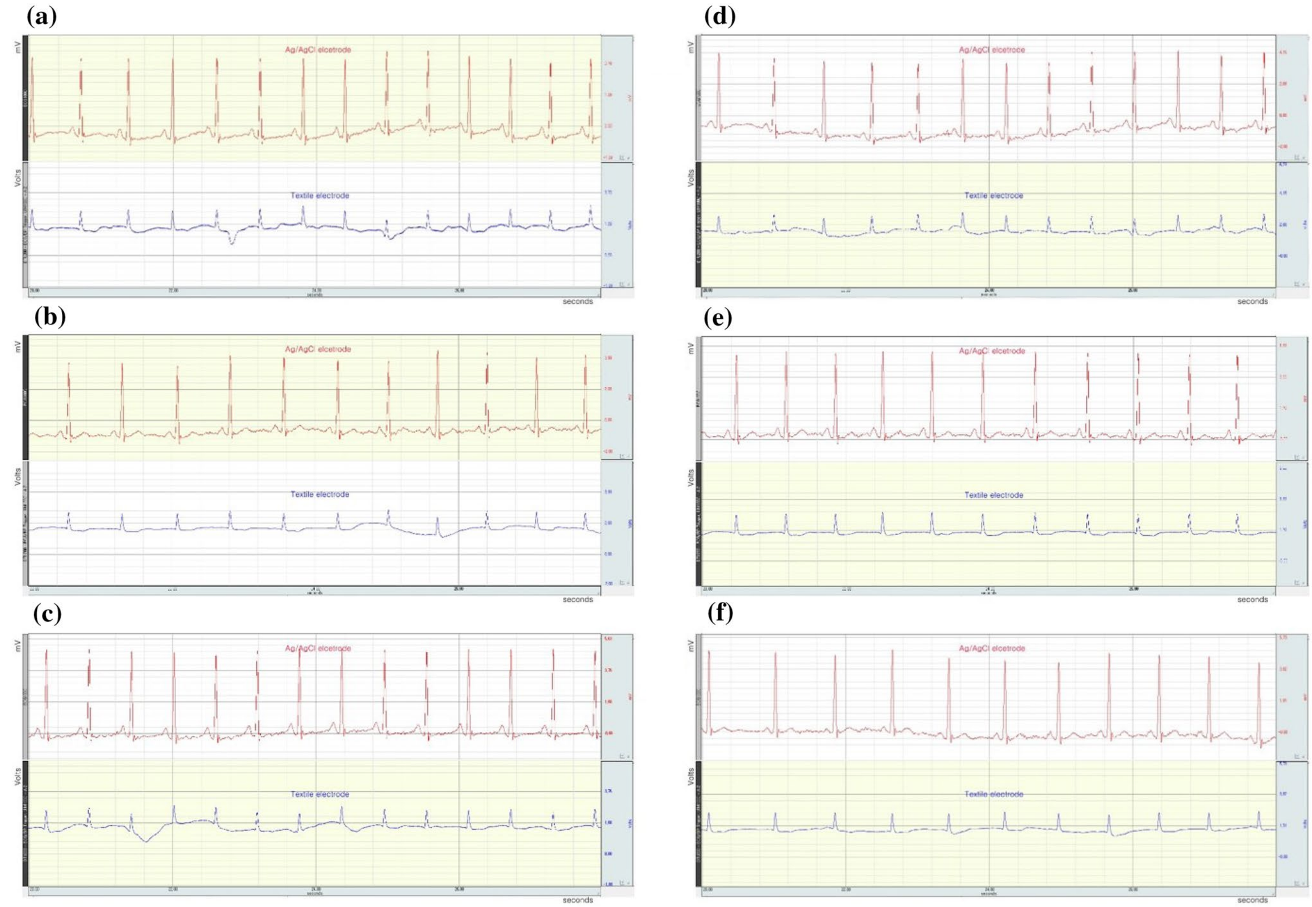

Fig. 3 The ECG signal of Subject 1 based on the type of textile electrode (top: Ag/AgCl, bottom: textile electrode): **a** flat electrode (1 × 1 cm); **b** flat electrode (2 × 2 cm); **c** flat electrode (3 × 3 cm); **d** convex electrode (1 × 1 cm); **e** convex electrode (2 × 2 cm); and **f** convex electrode (3 × 3 cm)

in the case of flat electrodes, although the detection of ECG signals was qualitatively confirmed, the waveforms were relatively unstable and the size of the acquired signals was small. Consequently, the overall signal quality was poor. Furthermore, unlike convex electrodes where the size did not have a significant effect on the signal quality, the quality of signals in flat electrodes varied depending on the size of the electrodes. In summary, the results of the qualitative analysis indicate that convex electrodes are more effective than flat electrodes for ECG signal detection.

4.2 Non-parametric Statistical Analysis

We used SPR as a parameter to perform the Kruskal–Wallis test, a non-parametric difference test based on a normality test ($p > 0.05$), to check if any difference exists in the performance between the six types of electrodes. The analysis results revealed a significant main effect among the six electrodes ($p < 0.001$). No significant interaction between the size and the configuration was found ($p > 0.05$). Additionally, the Bonferroni post-hoc test was performed to check the combinations of electrodes that exhibit these differences. The post-hoc test results indicate that a convex electrode of 1 × 1 cm performed better than flat electrodes of 2 × 2 and 3 × 3 cm with $p < 0.01$ in both cases. Furthermore, a convex electrode of 2 × 2 cm performed better than flat electrodes of 1 × 1, 2 × 2, and 3 × 3 cm, with $p < 0.05$, $p < 0.001$, and $p < 0.001$, respectively. Moreover, a 3 × 3 cm convex electrode exhibited superior performance over flat electrodes of 1 × 1, 2 × 2, and 3 × 3 cm with $p < 0.001$ in all cases. Table 1 and Fig. 4 present the comparison and the analysis results.

A non-parametric t-test, namely the Wilcoxon test, was conducted to test for differences in performance between the textile electrode configuration methods (flat and convex). Table 2 and Fig. 4 present the analysis results. A significant difference was observed in the SPR values of the ECG signals between the textile electrode configuration methods ($p < 0.001$). In other words, the quality of signals was better in the convex electrodes than that in the

Table 1 Comparison of six types of textile electrodes and post-hoc test results

Electrode type	n	Mean (SD)	Kruskal–Wallis test	p-value	Bonferroni post-hoc test
1 × 1 cm (convex)	40	− 11.62 (2.03)	60.644	8.947e−12	1 × 1 cm (flat) < 2 × 2 cm (convex)
1 × 1 cm (flat)	40	− 14.28 (7.07)			2 × 2 cm (flat) < 1 × 1 cm (convex)
2 × 2 cm (convex)	40	− 10.70 (1.83)			2 × 2 cm (flat) < 2 × 2 cm (convex)
2 × 2 cm (flat)	40	− 16.14 (5.06)			1 × 1 cm (flat) < 3 × 3 cm (convex)
3 × 3 cm (convex)	40	− 10.62 (2.33)			2 × 2 cm (flat) < 3 × 3 cm (convex)
3 × 3 cm (flat)	40	− 16.06 (3.99)			3 × 3 cm (flat) < 1 × 1 cm (convex)
					3 × 3 cm (flat) < 2 × 2 cm (convex)
					3 × 3 cm (flat) < 3 × 3 cm (convex)

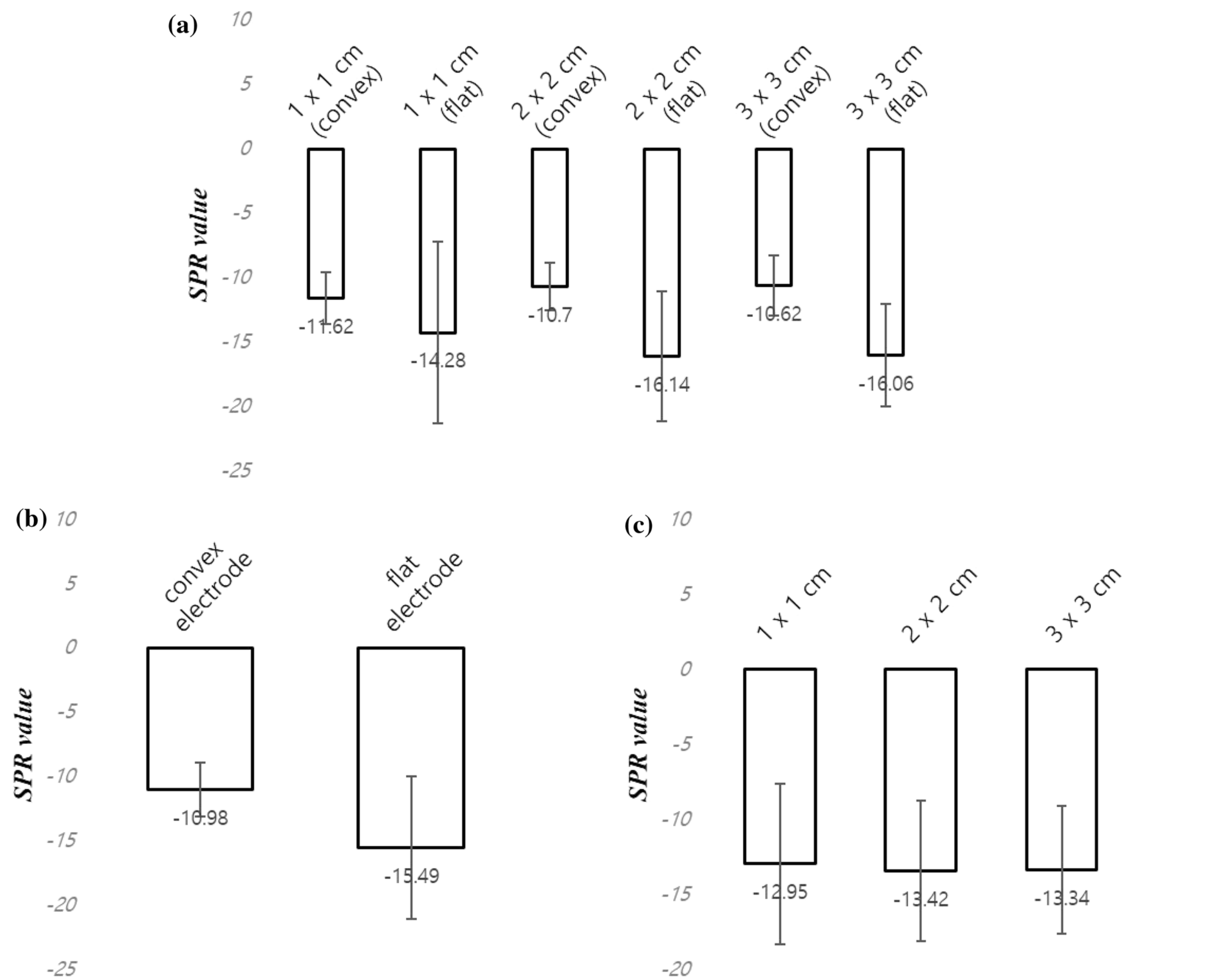

Fig. 4 Boxplots of sensing performance **a** for each textile electrode type, considering **b** the configuration method and **c** size of the textile electrodes

Table 2 Comparison of textile electrodes based on the flat or convex configuration

Config	n	Mean (SD)	Wilcoxon test	p-value
Convex electrode	120	− 10.98 (2.10)	7510	4.831e−14
Flat electrode	120	− 15.49 (5.54)		

flat electrodes. This is because, unlike the flat electrodes, the three-dimensionally structured electrodes attach better to the positions on the skin, which is suitable for ECG signal detection. Consequently, the contact surface area increased, which in turn reduced electrical noise and detected higher quality signals.

Table 3 Comparison of textile electrodes based on the three manipulated sizes

Size	n	Mean (*SD*)	Kruskal–Wallis test	p-value
1×1 cm	80	−12.95 (5.33)	0.0462	0.9772
2×2 cm	80	−13.42 (4.67)		
3×3 cm	80	−13.34 (4.24)		

Table 3 lists the results of the Kruskal–Wallis test conducted to analyze the variations in sensing performance based on the size difference of the textile electrodes. As indicated in Fig. 4, no significant difference was observed between the three sizes ($p > 0.05$). In other words, a research hypothesis assuming that the contact surface area increases proportionally if the electrode size increases, facilitating the detection of higher-quality signals, was rejected. We observed that the contact surface area does not increase by simply increasing the size of the electrode. Conversely, the contact surface area increases when the electrodes are in a configuration where they are well-attached to the skin in appropriate positions. Therefore, ECG signals detected through such configurations are of high quality.

4.3 Correlation Analysis and Bland–Altman Plot

The quality of the measured ECG signals is significantly affected by muscle movement, which exhibits a lower performance than the resting state [27]. Therefore, the effect of muscle activity on the performance of signals during movement should be considered. This study verified the performance of textile sensors in the quick-walking condition.

The SPRs extracted from the six textile sensors, namely 1×1 cm (F_1), 2×2 cm (F_2), and 3×3 cm (F_3) in the flat type and 1×1 cm (C_1), 2×2 cm (C_2), and 3×3 cm (C_3) in the convex type, were compared with the SPR value obtained from the ECG sensor (ground truth). We considered the correlation analysis (Pearson correlation coefficient (r)), mean error value (*ME*), and Bland–Altman plot (difference plot) for the comparison. The correlation coefficient ranging from − 1 to 1 was defined by the linear relationship between two or more variables. Correlation coefficients of 0.00–0.09, 0.10–0.39, 0.40–0.69, 0.70–0.89, and 0.90–1.00 represent negligible, weak, moderate (good), strong, and extremely strong correlations, respectively [28]. A Bland–Altman plot (mean–difference plot) is a method of plotting data to assess the concurrence between two different measurements. This plot includes the x and y axes, which correspond to the mean and difference values between two difference measurements, respectively. If most of the values obtained from two difference measurements are within $\overline{d} \pm 1.96s$ (a 95% confidence level: $\overline{d} + 1.96s$ and $\overline{d} - 1.96s$) in a Bland–Altman plot, the two quantitative measurements are considered to exhibit adequate concurrence. The symbols $\overline{d}$ and s represent the mean difference and standard deviation, respectively [27, 29]. All statistical analyses were performed using SPSS Statistics v.21.0 (IBM Corporation, Armonk, NY, USA).

When the correlation analysis results were compared with those of the ground truth, SPR from the textile sensor exhibited a good, strong, or extremely strong positive correlation (0.536 to 0.910) and low mean error (3.74 to 5.88) with the ECG sensor of all six electrodes. Herein, $r = 0.536$ ($p < 0.05$) and $ME = 3.74 \pm 2.23$ for F_1; $r = 0.569$ ($p < 0.05$) and $ME = 4.40 \pm 2.91$ for F_2; $r = 0.733$ ($p < 0.05$) and $ME = 5.88 \pm 1.87$ for F_3; $r = 0.691$ ($p < 0.05$) and $ME = 4.60 \pm 2.04$ for C_1; $r = 0.769$ ($p < 0.05$) and $ME = 4.64 \pm 2.26$ for C_2; and $r = 0.910$ ($p < 0.05$) and $ME = 5.06 \pm 1.20$ for C_3. Furthermore, the correlation coefficient of C_3 was the highest, and *ME* of F_1 was the lowest, as depicted in Fig. 5.

As illustrated in Fig. 6, the results of the Bland–Altman analysis confirmed that SPRs measured from ECG and the six textile sensors exist within the 95% limits of agreement (± 1.96 *SD*). In all textile sensors, the Bland–Altman plots indicate a good agreement based on a 95% confidence level between the SPRs from the ECG and textile sensors.

5 Discussion and Conclusions

In this study, we implemented six textile electrodes for detecting ECG signals, considering different sizes and configuration methods of the electrodes. We experimentally investigated the structural conditions of textile electrodes that are more suitable for obtaining ECG signals. The results indicate that the convex or flat electrode configuration significantly affects the quality of the contact-type ECG signals. Additionally, the experimental results validate that the convex electrodes of all sizes outperform the flat electrodes, verifying that the quality of signals is more affected by the electrode configuration than the size. Higher quality ECG signals can be obtained using convex textile electrodes regardless of the electrode size. Additionally, correlation coefficients for pairs of ECG and textile sensors were in the range of 0.536 to 0.910, indicating a good, strong, or extremely strong positive correlation. Mean errors were in the range of 3.74 to 5.88, representing a low difference between two measurements. The Bland–Altman plots validate that adequate concurrence was observed between the two types of measurements in all textile sensors. Based on the results of the correlation coefficient, mean error, and Bland–Altman plots, we determined that the textile sensor in convex 3×3 cm (C_3)

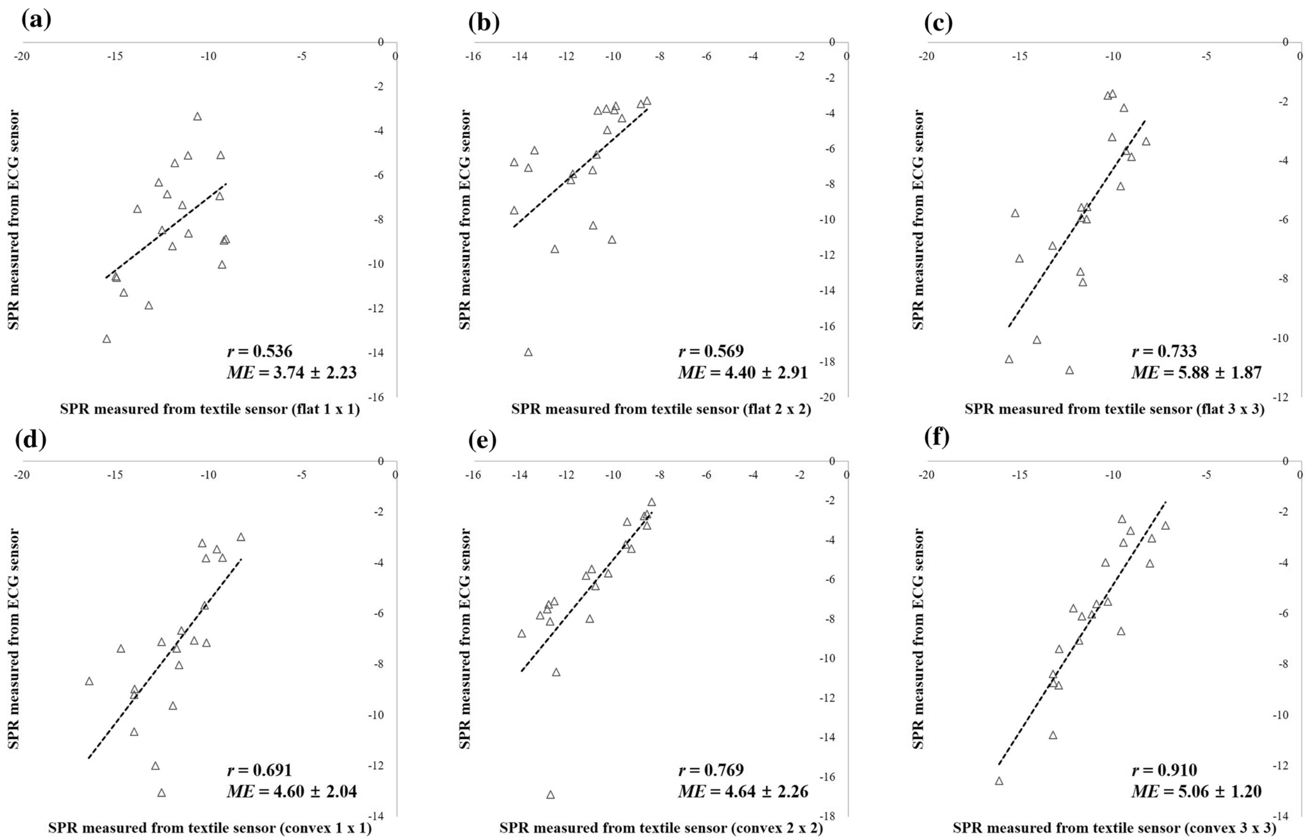

Fig. 5 Analysis results of correlation and mean error between the signal power ratio (SPR) values measured from ECG and six textile sensors: **a** F_1; **b** F_2; **c** F_3; **d** C_1; **e** C_2; and **f** C_3 ($p < 0.05$)

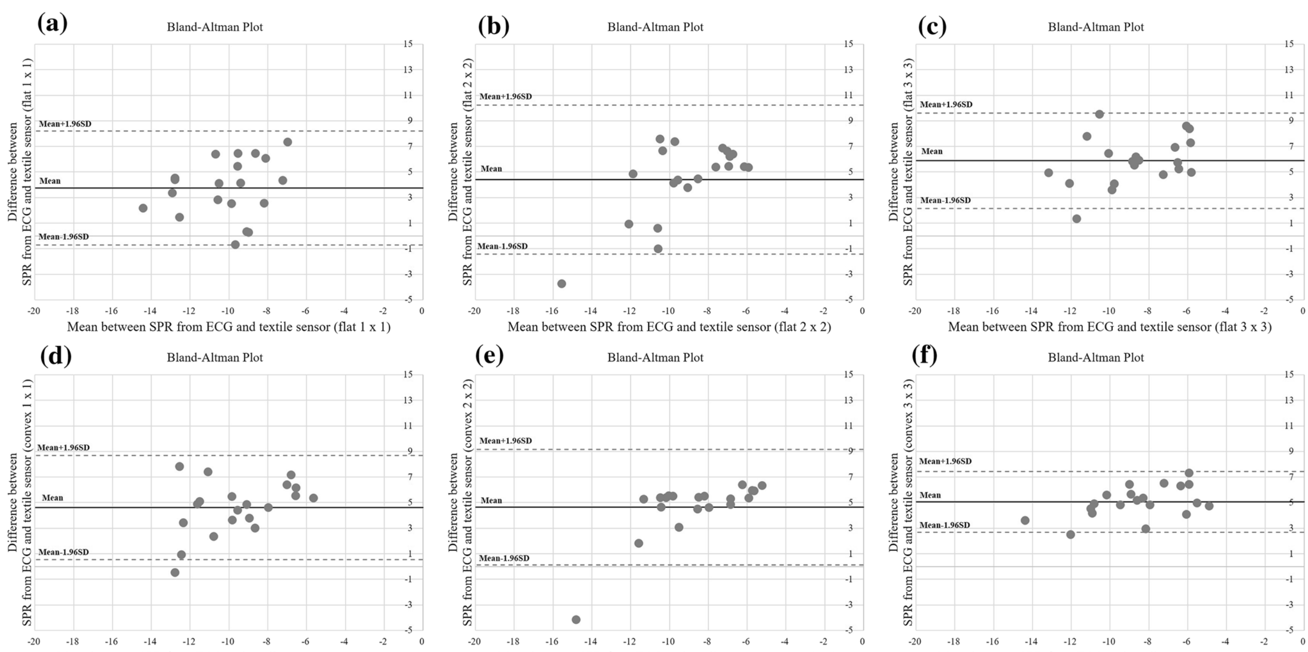

Fig. 6 Representative Bland–Altman plots for SPR obtained from ECG and the six textile sensors: **a** F_1; **b** F_2; **c** F_3; **d** C_1; **e** C_2; and **f** C_3. The solid central line in each plot represents the mean difference between the two measurements, whereas the upper and lower dotted lines represent the 95% limits of agreement (± 1.96 *SD*, $n = 40$)

type exhibits the best performance. These results indicate the possibility of a textile sensor replacing an ECG sensor.

As stated in Sect. 2.2, Song et al. [7] implemented four textile electrode types in a jacquard weaving method using silver-based metallic yarn, which were then applied to contact-type ECG signal measurements to analyze the results qualitatively. According to their findings, the performance of ECG signal sensing was the best when convex electrodes coated with conductive paste were used. The results of our study, which verify that contact textile electrodes implemented in a convex shape can obtain higher quality ECG signals compared to flat electrodes, are consistent with the aforementioned results. This confirms that a 3D structure can reduce the impedance of contact textile electrodes. Furthermore, we considered both the configuration and size variables of textile electrodes to analyze their structural condition for high-quality signal acquisition; however, no difference was observed in the sensing performance based on size. Nevertheless, the convex configuration of the textile electrodes exhibited a major effect on the quality of the ECG signals regardless of the electrode size. Therefore, future investigations are necessary in terms of the textile electrode structure, focusing on the configuration rather than the size. Furthermore, the skin impedance was extremely large compared to the electrode impedance, and the skin impedance increased more in the case of dry-type electrodes causing difficulty in measurements. In the future, we intend to determine a method for reducing the impedance of dry-type electrodes.

Previously, Cho [9] and Koo et al. [17] investigated the effect of garment structures on the signal quality when obtaining ECG signals. Therefore, we intend to combine a garment structure suitable for the ECG signal acquisition with the convex textile electrode structure to analyze if high-quality ECG signals can be obtained even during motion. Additionally, we plan to develop upgraded textile electrodes with more research variables, capable of detecting high-quality ECG signals. Based on this, we intend to implement a garment-type platform for integrating convex textile electrodes and investigate methods for enhancing the performance of smart garment technology that facilitates high-quality ECG signal acquisition without time and space constraints. Thus, we believe that the ECG signal acquisition technology can serve as an important indicator of body condition monitoring for preventive measures in sports and medical treatment purposes.

Author contributions Conceptualization, J.K.K., H.-S.C., and J.-W.L.; methodology, J.K.K., S.P. and H.-S.C.; software, J.-H.Y.; validation, J.K.K. and S.P.; investigation, S.P. and H.-S.C.; data curation, S.P. and J.-H.Y.; writing—original draft preparation, J.K.K. and H.-S.C.; writing—review and editing, S.P. and S.-H.L.; visualization, S.P. and J.-H.Y.; supervision, S.-H.L. and J.-W.L. All authors have read and agreed to the published version of the manuscript.

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Declarations

Conflict of interest The authors declare no conflict of interest.

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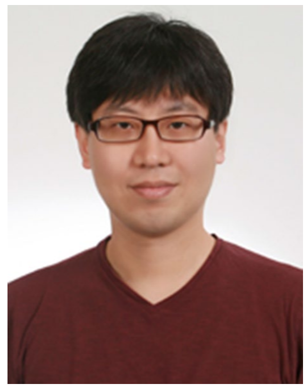
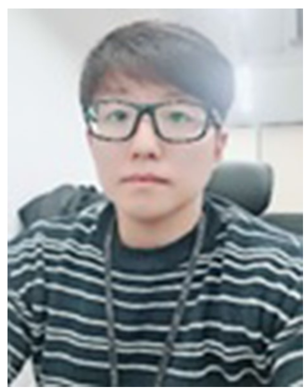


Jae Kwan Kim He received the B.S. and M.S. degrees in mechanical engineering from the Yonsei University, Seoul, Korea in 1998 and 2000, respectively. His research interests include artificial intelligence, machine learning, brain-computer interface, biomedical engineering, and process planning.



Sangin Park He received the B.S. and Ph.D. degrees in emotion engineering from the Sangmyung University, Seoul, Korea in 2012 and 2016, respectively. His research interests include brain-computer interface, non-contact measurement, neuroscience, cardiovascular physiology, and emotion engineering.

Hyun-Seung Cho She received the Ph.D. degree in Cognitive Science from the Yonsei University, Seoul, Korea in 2004. Her research interests include smart clothing and wearable technology.

Jin-Hee Yang She received the Ph.D. degree in Clothing and Textiles from the Yonsei University, Seoul, Korea in 2015. Her research interests include smart clothing and wearable technology.

Soo-Hong Lee He received the B.S. degree in mechanical engineering from the Seoul National University in 1981 and the M.S. degree in mechanical engineering design from the Seoul National University in 1983. He completed his Ph.D. from the Stanford University, California, USA in 1991. He is currently a Full-time Professor at the School of Mechanical Engineering, Yonsei University, Seoul, Korea. His research interests include intelligent CAD, knowledge-based engineering design, concurrent engineering, product design management, product lifecycle management, artificial intelligence in design, and design automation.

Jeong-Whan Lee He received the B.S. and M.S. degrees in electrical engineering in 1992 and 1994, respectively, and the Ph.D. degree in electrical and computer engineering from the Yonsei University, Seoul, Korea in 2000. He worked as a member of the technical staff, M_Application Project Team and a Project Manager, Ubiquitous-health Project Team at Samsung Advanced Institute of Technology, Kyounggi, Korea, from 2000 to 2004. Since September 2004, he has been a Faculty Member of the School of Biomedical Engineering, Konkuk University, Chungju, Korea. His research interests include biomedical signal processing and instrumentation, wearable/wireless devices for pervasive healthcare, sensing of vital signs/signatures, heart-rate variability, and biological signal measurements by electromagnetic waves and microwave radio-thermometry.