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EEG Measurement Using Dry Electrodes Comprising Two-layered Conductive Silicone with Different Carbon Concentrations*

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***Abstract*—Electroencephalography (EEG) typically involves the use of wet electrodes, which are inconvenient and impair signal quality. In this study, we propose a new electrode comprising two-layered conductive silicone with different carbon concentrations as a non-metallic flexible EEG electrode. The proposed electrode was shown to function as an EEG electrode while maintaining its superior flexibility compared to non-metallic dry electrodes, such as carbon-containing polymer electrodes. Moreover, we hypothesized that the signal quality varies with carbon concentration and contact area, and thus compared the signal correlation coefficient and SNR with that of the wet electrode for 20 sets of electrodes: 15 sets with the carbon:silicone weight ratio ranging from 1.7% to 3.1%, and 5 sets with the electrode diameter in the range 8–18 mm. We observed that the signal quality of the proposed electrode was closest to that of the wet electrode when the carbon:silicone weight ratio was 2.6%–2.8% and the diameter was 8–10 mm. Our results suggest that the proposed electrode is a viable gel-free non-metallic flexible electrode and is suitable for daily use.**

Index Terms—Electroencephalography, conductive silicone, soft sensor, EEG electrode, non-metallic dry electrode

I. INTRODUCTION

Since the discovery of the ability to monitor the brain's electrical activity from the scalp for the first time in 1929, electroencephalography (EEG) has since been extensively studied as a non-invasive measurement method of brain activity. In recent years, while various other methods have been invented, EEG has received renewed interest because it is inexpensive, the equipment is easy to introduce, and it can be performed on a wider range of subjects than other methods. The method for analyzing EEG signals has been improved; however, the method for acquiring EEG signals has not changed much [1].

Typically, wet electrodes are used for EEG measurement. They need to be professionally operated for sticking to the scalp and are known to cause a sense of discomfort [2]. This problem has not been resolved. However, in recent years, gel-free dry electrodes have been developed, which significantly retain signal quality compared to that of conventional wet electrodes. However, many of these electrodes are not suitable for daily-use devices because either metallic or non-contact electrodes are required [3][4]. Dry electrodes are now being developed using active electrodes and other new materials. These electrodes are suitable for daily-use devices to measure and use EEG in real time. As a daily-use device using bioelectricity, there is a myoelectric hand. Most commercially available surface electromyography (sEMG) prosthetic hand systems often adopt dry electrodes due to the running costs and user inconvenience caused while contacting the skin. However, many such electrodes are metallic and can thus directly contact the skin for measuring both EEG and sEMG [1][5]. These metallic electrodes cannot be used for a long time due to their rigidity, and some users are allergic to certain metals [4]. There are examples of sEMG prosthetic hand systems that use flexible metal cloth electrodes, but many infant users were reported to have experienced discomfort [6]. Although there are numerous proposals for mechanisms that mitigate stiffness in the EEG electrode, non-metallic materials are desirable as sEMG electrodes. Polymer electrodes containing carbon are used as non-metallic EEG electrodes, but they leave a mark on the skin due to their inflexibility [7]. We have been developing soft and conductive electrodes for sEMG on the human forearm, that cause less discomfort and stress to the skin [8][9][10][11]. Our developed electrode uses two layers of silicone that exhibit conductivity due to added carbon. This electrode overcomes the problem associated with gels, metals, and the rigidity of conventional sEMG electrodes. Therefore, similar problems in EEG measurement can be mitigated by this electrode if it can perform EEG measurements.

II. DEVELOPMENT OF THE PROPOSED ELECTRODE

A. Process of developing an electrode for sEMG

Conventional electrodes used to measure sEMG are wet electrodes. In recent years, metallic electrodes have often been employed in man–machine interfaces, since gel-based electrodes are not suitable for long-term measurement [12]. However, it is difficult to maintain stable electrical contact because stiff metal electrodes do not assume the shape of the skin [10]. Therefore, we have been developing dry electrodes that use soft conductive materials to solve this problem [8][9][10][11]. The electrode developed in 2017 employed silicone that exhibited conductivity by blending carbon black [9]. However, the subjects expressed discomfort regarding the

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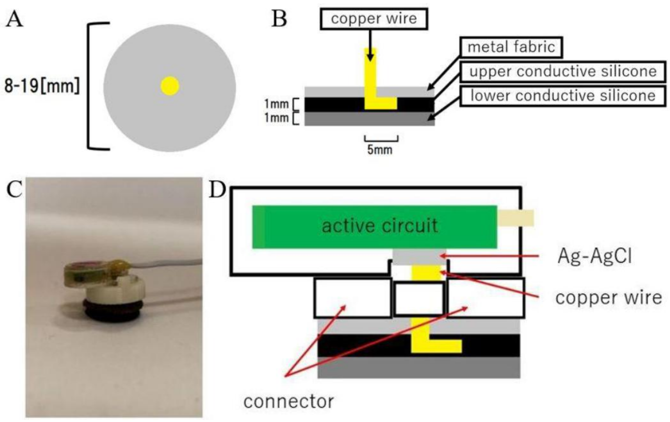

Fig. 1 (A) Schematic view from above of the proposed electrode. There are 5 types of diameters between 8-19mm. (B) Cross-sectional view from the side of the proposed electrode. This electrode consists of copper wire, metal fabric, and two types of conductive silicone. (C) The sensor used in experiment with proposed electrode attached by connector. (D) The details of the sensor.

structure of this electrode when the gold-coated copper wire came into contact with the skin. Thus, another electrode was developed in which the skin contact was silicone [8]. This electrode employed two layers of silicone with different carbon weight ratios, and the gold-coated copper wire was embedded within this. At present, our group employs the one with a carbon weight ratio of 2.6% to the layer that contacts skin and 4.0% to the other layer with the wire [8].

B. *Materials*

The silicone (TSG-E30, Tanac Co., Ltd, Japan.) that is used as a material for the electrode complies with Specification and Standards for Food, Food Additives, etc., the official notification from Ministry of Health, Labor and Welfare of Japan. There are five members in this notification: food, additives, utensils and packaging containers, toys, and cleaning products. The safety of the proposed electrode complies with the abovementioned aspects. Conductive silicone is produced by blending the silicone with carbon black (EC600JD, Lion Specialty Chemicals Co. Ltd., Japan.). As shown in Fig. 1 (B), the conductive silicone of the proposed electrode has a two-layer structure. The layer that contacts the skin is “lower”, and the layer that covers the copper wire is “upper”. The upper conductive silicone has a large carbon weight ratio (4%) to reduce the resistance to copper wire. Lower conductive silicone has different carbon weight ratios (15 patterns from 1.7% to 3.1%) in order to obtain the optimum value for measuring EEG from experiments. Since these two layers of conductive silicone are the same material, they are strongly bonded to each other. Also, these have low electrical resistance. The proposed electrode is sufficiently flexible because the maximum elongation of the silicone is 840% [11].

III. Verification on EEG

The proposed electrode showed reliable results for sEMG measurement on the forearm. However, the thickness and characteristics of the subcutaneous tissues between the signal source and the electrode for sEMG and EEG measurement are different. Therefore, the impedance characteristics of the two are different. To confirm that the signals acquired are satisfactory for EEG, it is necessary to verify whether the proposed electrode can acquire the same signals as the conventional wet electrode.

A. *Method*

The EEG subject of this study satisfies the following requirements.

1) Subjects are awake
2) Potential amplitude is large
3) Appearance is not momentary
4) Easy task to implement

Based on the above, α waves were identified as the EEG signal in this experiment. If α waves were acquired from the wet electrode and the proposed electrode at almost the same position and the signals in both electrodes appear simultaneously, the proposed electrode was considered to detect EEG. The subjects performed a task of opening and closing their eyes every 10 s repeatedly, since α waves distinctly appear when the eyes are closed. The criteria of evaluation were as follows:

1) The RMS amplitudes are significantly higher with closed eyes than with opened eyes;
2) The correlation coefficient at every second (2048 points) between both signals repeats a cycle of increase and decrease every 10 s;
3) The signal strength repeatedly increases and decreases every 10 s in the 10 Hz band, according to time–frequency analysis

The amplitude RMS is defined below.

$$E_{RMS} = \hat{E} = \sqrt{\frac{1}{N}\sum_{n=1}^{N}|E_n|^2} \tag{1}$$

E_{RMS} is the amplitude RMS per 10 s, N is the number of signal points measured in 10 s, E_n is the amplitude of the signal at a point, and $N \in \mathbb{N}$. This experiment used an electrode with a carbon weight ratio of 4.0 + 2.6% and a diameter of 13 mm, which was confirmed to be reliable in sEMG. The experiment was conducted with the informed consent of one healthy male.

B. *Equipment*

EEG measurements in this experiment were performed using the Active Two System (BioSemi Co., Ltd, Netherlands). A conductivity gel (SignaGel, Parker Laboratories, Inc., USA) was used to attach the wet electrode and the ground electrode. The reference electrode was placed at the top of the head (Cz position) and the proposed electrode was placed at the back (Oz position) to measure α waves. Cz and Oz are positions mandated by the International 10-10 System with anatomical implications [13]. In the experiment, an electrode tip was designed to be replaced for investigating the relationship between the carbon weight ratio and the measurement characteristics. Fig. 1(C) shows the electrode assembled with a sensor, and Fig. 1(D) shows its cross-sectional view. The data were saved and then filtered by an 8–15 Hz band-pass filter. Subsequently, the data were analyzed for the time correlation coefficient and for time–frequency using MATLAB 2018.

C. Results

Fig. 2 shows the filtered wave. Both waves can be seen to repeat similar fluctuation in their amplitude every 10 s. Fig. 3 shows the amplitude RMS of both electrodes. These significantly increase when 10-20 seconds, 30-40 seconds, 50-60 seconds, i.e. when the subject closes their eyes. Although an ideal square wave cannot be measured, the fluctuation in the correlation coefficient were seen every 10 s. Table I shows the result of t-test; the correlation coefficient appears significantly increased corresponding to when the subject's eyes were closed ($p < 0.05$). Fig. 5 shows a spectrum power map for time–frequency. It can be seen there are fluctuation in the power spectrum in the 10 Hz band, although inferior compared to those obtained with the wet electrode. Nevertheless, it was found that the proposed electrode was able to measure α waves.

TABLE I
COMBINATIONS OF THE P-VALUE OF CORRELATION COEFFICIENT EVERY 10 SEC

	Open	Closed	Open	Closed	Open	Closed
Time[s]	0–10	10–20	20–30	30–40	40–50	50–60
0–10	-	0.011	0.927	0.003	0.580	0.013
10–20	-	-	0.012	0.248	0.038	0.992
20–30	-	-	-	0.000	0.409	0.006
30–40	-	-	-	-	0.003	0.250
40–50	-	-	-	-	-	0.040
50–60	-	-	-		-	-

Gray area is for a combination of opened and closed eyes

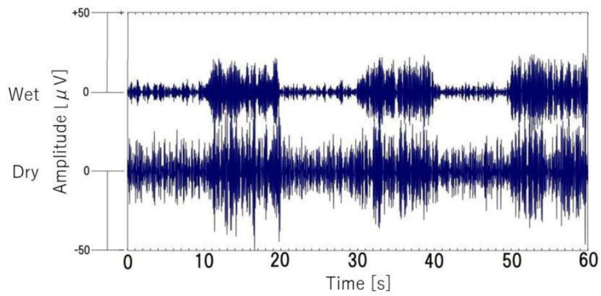

Fig. 2 The subject closed his eyes at the times 10–20 s, 30–40 s, and 50–60 s. The amplitudes increased markedly at these times.

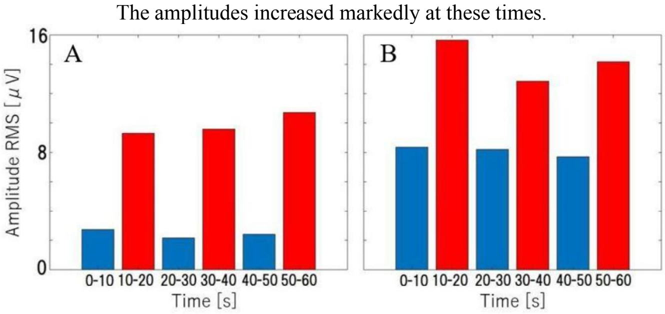

Fig. 3 The amplitude RMS measuring (A) conventional wet electrode (B) the proposed electrode. The amplitude RMS significantly increased when the subject closed his eyes. The α wave appears when the eyes are closed, as the RMS is almost identical for the same conditions.

IV. PARAMETER SEARCH EXPERIMENT

The impedance difference between two electrodes in a differential amplifier circuit needs to be low despite any difference in the contact state of the two electrodes. In the proposed electrode developed for sEMG measurement, a minimum difference in the impedance can be achieved by experimenting with different carbon weight ratios in the lower layer and electrode area. Similar experiments were performed on EEG measurement.

A. Method

Table II shows the types of electrodes manufactured for the experiment. The experiment involved the subjects opening and closing their own eyes every 10 s for about twenty electrodes. Evaluations were conducted for the following criteria:

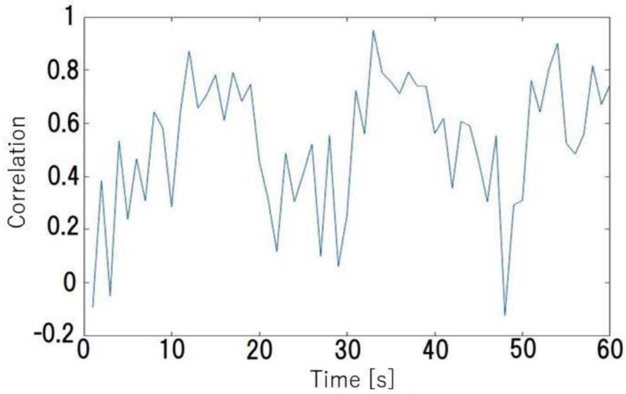

Fig. 4 The correlation coefficient repeatedly increases and decreases every 10 s. The same signal can be obtained from both electrodes when a subject closes their eyes.

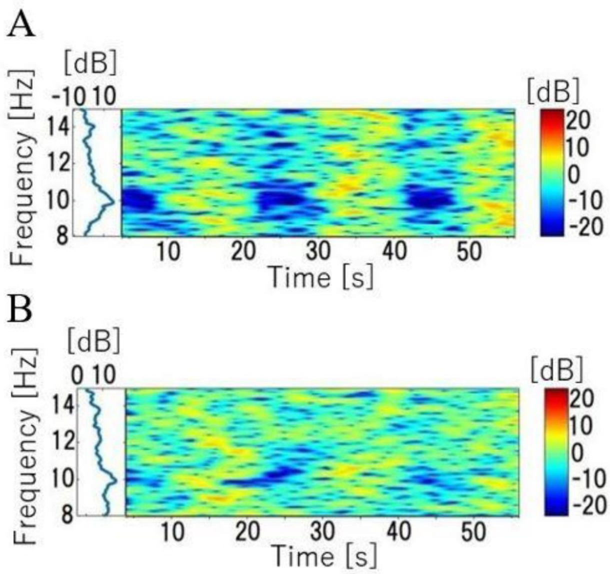

Fig. 5 Spectrum power map for time-frequency analyzing (A) conventional wet electrode (B) the proposed electrode. Both maps show three fluctuations in signal intensity at 10 Hz, which corresponds to α waves.

1) Correlation coefficient of the wet–proposed electrodes while eyes are closed;

2) Defined SNR

The SNR was expressed by the following equation;

$$\eta = \left(\hat{E}_{C_n}/\hat{E}_{O_n}\right)^2 \tag{2}$$

is the RMS amplitude for closed eyes and is that for opened (n = 1,2,3). A higher SNR implies higher performance of the electrode, as it is known from the previous experiment that the amplitude RMS is high at the time of closed eyes (only < 10). The equipment used and the position of the electrodes were the same as in the previous experiment. This experiment was conducted with the informed consent of three healthy males.

B. Results of experiments

Fig. 6(A) and (C) show the correlation coefficient of the wet–proposed electrodes for closed eyes of all subjects. The graph is the median of all trials (18 times) for one electrode and the margin of error is in the inter-quartile range. Some carbon weight ratios or electrode diameters that give higher values than others. Electrodes that have 2.6–2.8% carbon weight ratio or 10 mm exhibit a closer measurement characteristic to that of a wet electrode.

Fig. 6(B) and (D) show the SNR for all subjects. The graph is the median of all trials (18 times) for one electrode and the margin of error is in the inter-quartile range. Most of the obtained SNRs are in the range 1–2. The proposed electrode can measure EEG, but it has a higher tendency to detect stationary noise than wet electrodes; the result of the same calculation for wet electrodes is about 10.

TABLE II
NUMBER OF MANUFACTURED SETS PER LOWER STATE CARBON WEIGHT RATIO [%] AND DIAMETER OF THE ELECTRODE

Experiments	Carbon weight ratio search	Diameter search
Carbon weight ratio [%]	1.7 to 3.1	2.6
Diameter [mm]	13	8, 10, 13, 16, 19
Quantity [sets]	15	5

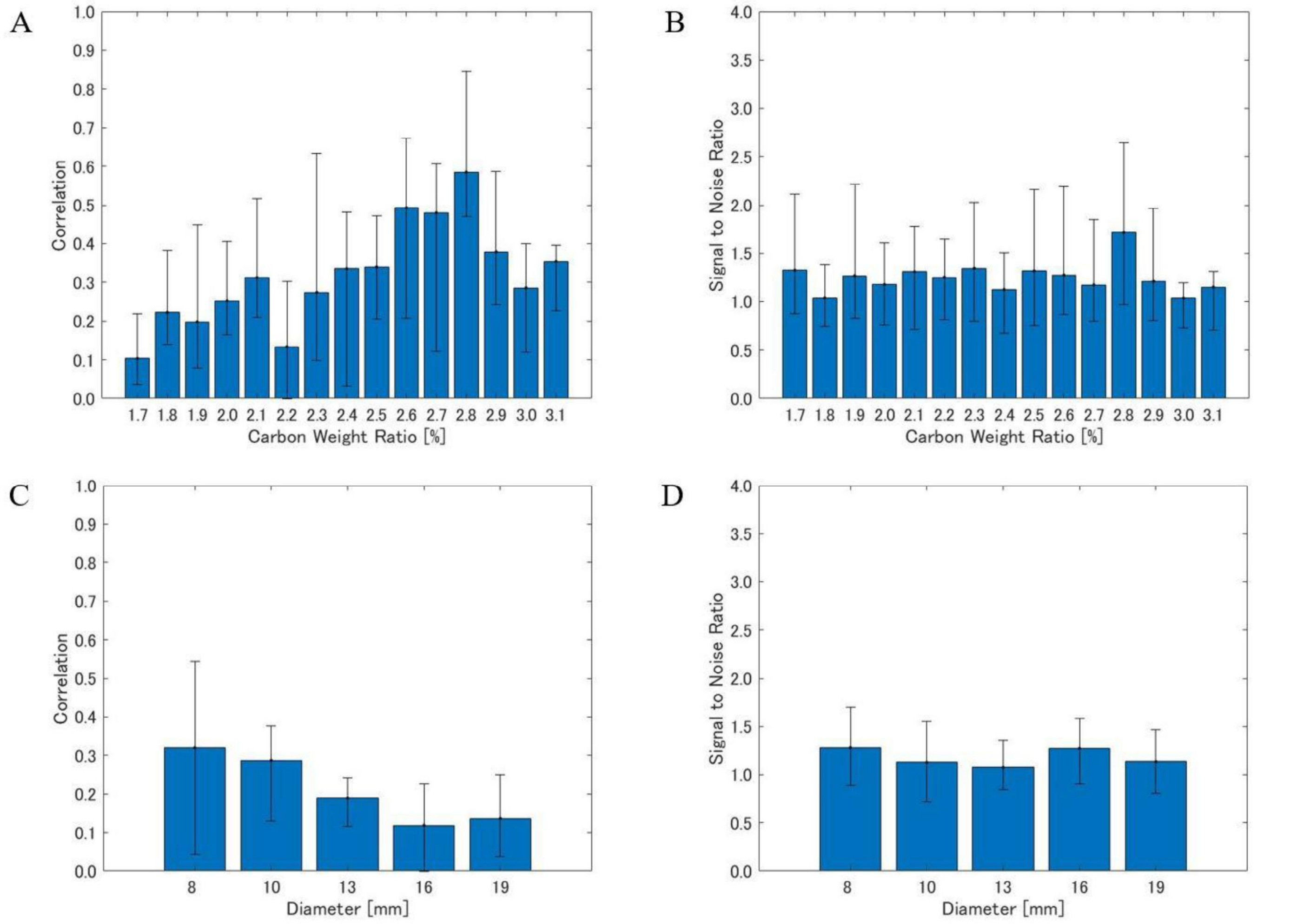

Fig. 6 Results of parameter search experiment. (A) Correlation coefficient for each carbon weight ratio (B) SNR for each carbon weight ratio (C) Correlation coefficient for each diameter (D) SNR for each diameter. The blue bar represents the median and the error bar represents the inter-quartile range. There are parameters that show performance closer to that of conventional wet electrode when comparing correlation coefficient.

V. Discussion

The proposed electrode was found to detect α waves, which are a type of EEG waves. These electrodes exhibited altered measurement characteristic as a result of changing the weight ratio of carbon or scalp contact area, and for a particular parameter, they showed similar performance to conventional wet electrodes. Moreover, Fig. 6(A) shows two peaks at 2.1% and 2.8% when changing the carbon weight ratios. Such two peaks were also seen in the performance evaluation experiment for sEMGs, due to the following reasons:

1) A carbon weight ratio of 2.8% is the limit at which the physical property as silicone can be maintained;

2) There is a carbon weight ratio with the smallest micro-electrical function inside the electrode that changes due to external forces.

The surface of the electrode is rough when the carbon weight ratio is 2.9% or more. This is considered to be a phenomenon that occurs when the carbon powder is not mixed with silicone. The roughness of the contact area has a drastic effect on the impedance of the scalp–electrode interface. Because unstable impedance leads to unstable measurement, the carbon weight ratio of 2.8%, which can maintain the physical property as silicone, exhibits the best measurement performance. Furthermore, conductive silicone is considered to be an assembly of minute conductors and insulators. In other words, these are considered to be an assembly of minute capacitors [14]. Therefore, there is a carbon weight ratio with the smallest micro-electrical function inside the electrode that changes due to external forces.

Fig. 6(C) shows that the signal quality tends to deteriorate as the electrode diameter increases, contrary to the expectation that as the electrode diameter increases, the pressing pressure will decrease and the signal quality will increase. These are considered to be caused by the influence of hair. Although different from what is expected, spatial resolution is an important factor in EEG measurement unlike in sEMG measurement, and when the signal quality is more stable as the electrode size is smaller, the spatial resolution simultaneously increases. This result is highly desirable for EEG measurements.

VI. Conclusion

This paper proposes a new dry-contact electrode for EEG, consisting of conductive as well as flexible material to ameliorate the problems associated with conventional wet and dry electrodes. We showed the following empirical characteristics of the proposed electrode:

1) The electrode can measure EEG;

2) An electrode having a carbon weight ratio of 4% + 2.8%, or a diameter of 10 mm, shows the most suitable measurement characteristics for EEG measurement.

The tendency for detecting stationary noise is higher than that of wet electrodes and is likely to be a problem in the contact state of the electrode, shown in Fig. 1(D). In this experiment, the electrode had such a structure because it was necessary to replace the conventional electrodes, but at the same time, the electrodes needed to be connected directly to the active circuit. Therefore, this problem needs to be addressed in future. If such problems are solved, the proposed electrodes can contribute to conduction of EEG measurements in daily life.

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