Novel Dry Polymer Foam Electrodes for Long-Term EEG Measurement

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***Abstract*—A novel dry foam-based electrode for long-term EEG measurement was proposed in this study. In general, the conventional wet electrodes are most frequently used for EEG measurement. However, they require skin preparation and conduction gels to reduce the skin–electrode contact impedance. The aforementioned procedures when wet electrodes were used usually make trouble to users easily. In order to overcome the aforesaid issues, a novel dry foam electrode, fabricated by electrically conductive polymer foam covered by a conductive fabric, was proposed. By using conductive fabric, which provides partly polarizable electric characteristic, our dry foam electrode exhibits both polarization and conductivity, and can be used to measure biopotentials without skin preparation and conduction gel. In addition, the foam substrate of our dry electrode allows a high geometric conformity between the electrode and irregular scalp surface to maintain low skin–electrode interface impedance, even under motion. The experimental results presented that the dry foam electrode performs better for long-term EEG measurement, and is practicable for daily life applications.**

***Index Terms*—Biopotentials, conduction gel, dry electrode, electroencephalography, skin–electrode interfaces impedance.**

I. Introduction

EEG is a kind of method to measure electrical activities of the brain by using electrodes along the scalp skin [2]. It is also a powerful noninvasive tool and can provide high temporal resolutions to reflect the dynamics of brain activities directly [3]. It has been widely used for both medical diagnosis and neurobiological researches [2], [4]–[8]. In general, the conventional wet electrodes are most frequently used for EEG measurements [2]. However, the wet electrodes require skin preparation and conduction gel to reduce the skin–electrode interface impedance [9]. These procedures usually make trouble to users easily [10]; in particular, conduction gels will inevitably leave its residues on the scalp. Conduction gel may also leak out EEG electrodes to cause short circuit between two electrodes in the close proximity, when too much gel is applied or wet electrode is pushed down on the scalp hard. Moreover, these aforementioned preparation procedures are time consuming, uncomfortable, and even painful for participants, since the skin preparation usually involves the abrasion of the outer skin layer. Repeated skin preparations and gel applications may also cause allergic reactions or infections. The EEG signal quality may degrade over an extensive time as the skin regenerates and/or the conduction gel dries [10]. There also exist some troubles as the interested measuring location covered with hairs, leading to an insufficient skin–electrode contact area.

In order to improve these limitations of conventional wet electrodes, several kinds of dry electrodes have been developed [11]–[18]. Most of these dry electrodes were made by the microelectromechanical systems (MEMS) technique [13], [16]–[18]. Griss *et al.* proposed the microspike MEMS dry electrodes to acquire the forehead EEG signals successfully [16]. However, the aforementioned MEMS dry electrode technique acquired the EEG signals in an invasive way and only on the forehead sites. Moreover, there still exist some drawbacks by using the MEMS electrodes, such as 1) attendant pains when MEMS electrodes penetrate into the skin [19]; 2) lack of physical strength during the penetration; and 3) high cost for the manufacture procedure of MEMS electrodes. Matthews *et al.* also proposed a novel zero-preparation dry electrode for EEG measurements [11], [12]. However, the electrodes with the hard substrate will lead to uncomfortable or even hurt the scalp skin when applied in daily life. Moreover, the hard substrate will also lead to the EEG signal distortion due to motion effects.

Recently, fabric-based electrodes have been widespread used for biopotential measurement [1], [20]–[23]. Beckmann *et al.* have investigated the characterization of fabric electrodes with difference fabric specifications for ECG measurement in detail [1]. Baek *et al.* fabricated a polydimethylsiloxane-based dry electrode for long-term ECG measurement (more than 7 days) [22]. It is important that the previous studies indicate that, by using fabric-based electrodes, long-term biopotential signal monitoring is available [20]. Compared to the MEMS-based

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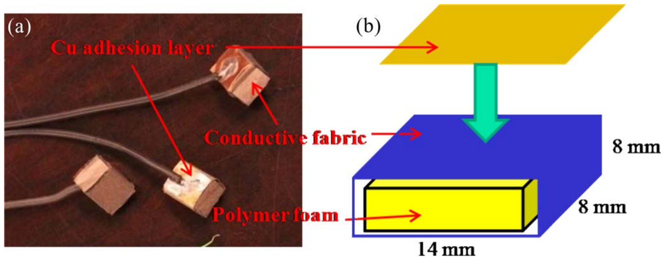


Fig. 1. (a) Top view and (b) exploded view of our proposed dry foam EEG electrode. The foam electrode was covered by the conductive fabric on all surfaces and then paste on a Cu layer.

electrodes, biopotential signal measurement by using fabric-based electrodes is performed in a comfortable and noninvasive way. However, the fabric-based electrodes are still unsuitable for measuring biopotentials on hairy sites due to the reduction of the contact area of the skin–electrode interface caused by the hairs.

In this study, a novel dry foam-based electrode was proposed for long-term EEG measurement, even on the hairy sites. The dry foam electrode was fabricated by an electrically conductive polymer foam covered by a conductive fabric. The electrically conductive polymer fabric contains partly polarizable electric characteristic to provide polarization and conductivity. The proposed dry foam electrode can be used to measure biopotentials without skin preparation and conduction gel. Moreover, the foam substrate of our dry electrode allows a high geometric conformity between the electrode and irregular scalp surface to maintain low skin–electrode interface impedance, even under motion. The rest of this paper was organized as follows. Section II introduced the design of the dry electrode. In Section III, the testing results for the proposed design were presented. In Section IV, the conclusion was drawn.

II. Design of a Novel Dry Foam Electrode

The design of our proposed dry foam electrode is shown in Fig. 1(a) and (b). It was designed to contact the skin by electrically conductive polymer foam with the compression set about 5–10%, which was made by urethane material. It was covered with a 0.2-mm-thick taffeta material made electrically conductive polymer fabric (conductive about 0.07 Ω /square) and coating with Ni/Cu on all surfaces to establish an electrical contact similar to that of the dry silver electrodes. A 0.2-mm layer of Cu was used as an adhesion layer. The size of our dry foam electrode is 14 (L) mm × 8 (W) mm × 8 (H) mm.

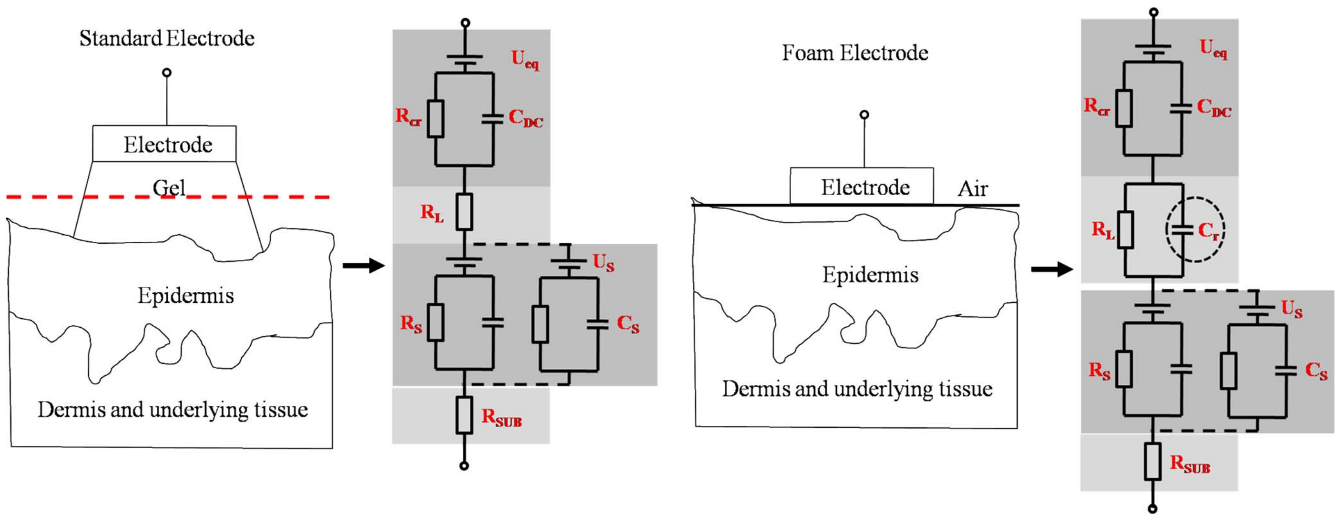
Biopotentials are electrical potentials inside human body, and Cl–, Ca+, and Na+ ions are used as charge transports in an organic system, in contrast to electrons in the leads of a sensing device. Therefore, by using electrodes to measure these biopotentials, the ion currents in the body have to be changed to electron currents in the electrode. The outer layer of the skin has a dry dielectric layer, which is called stratum corneum and will cause reduction of the transfer mechanism from ions to electrons [24]. Because the few-polarizable Ag/AgCl electrode is usually used as the conventional wet electrode, conduction gel has to be applied to moisturize the skin outer layer and change it to a highly ion-conductive layer. The equivalent circuit model of the skin–electrode interface for the conventional wet electrode is shown in Fig. 2(a) [25]. Here, C_{DC} denotes the double layer. R_{CT} and R_L are the impedances of the electrode–electrolyte interface and the electrolyte, respectively. And U_{eq} denotes the electrode potential.

Different from the conventional wet electrode, our dry foam electrode exhibits both polarization and conductivity due to the partly polarizable electric characteristic of electrically conductive polymer fabric [25], and it can provide a strong capacitive behavior at the interface [26]. The equivalent circuit model of our dry skin–electrode interface is shown in Fig. 2(b) [25]. Here, C_T is the equivalent capacitor at the interface. Therefore, biopotentials can be measured by the inductive method via C_T, and the influence of stratum corneum can be effectively reduced. Moreover, sweat and skin humidity can also form a conductive path R_L [27]. The foam substrate of our dry electrode can fit the scalp surface well to increase the contact area between skin and electrode to reduce the impedance R_L. Moreover, different to the fabric-based electrodes [25], the foam is not only used to reduce the motion force, but also potentially used to increase the fabric–skin contact area when force is applied on the electrode. It will also assimilate the motion force, rubbing and sliding of the electrode on the skin, to reduce the motion artifact and skin–electrode interface impedance.

III. Results and Discussion

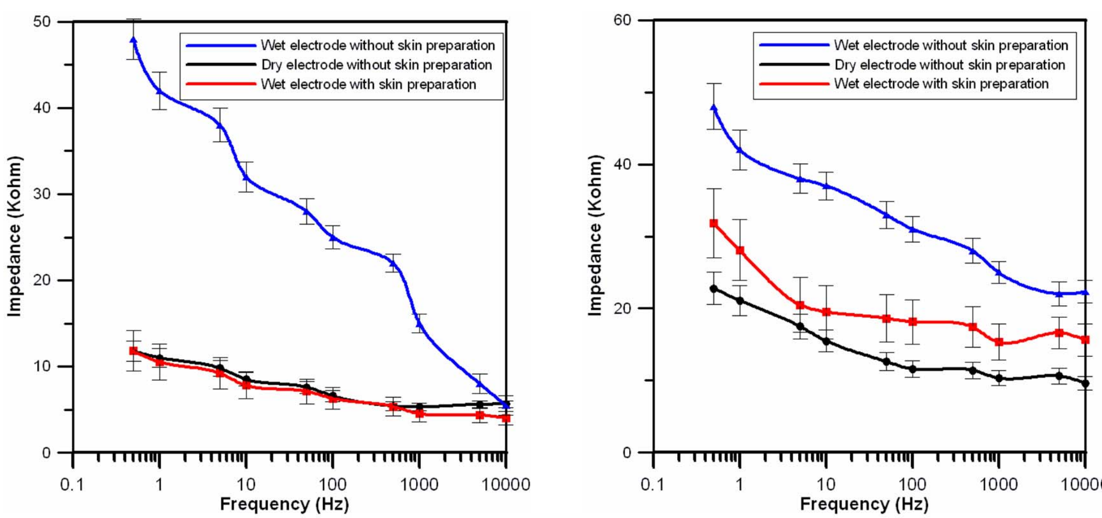
A. Impedance Variation for Long-Term Measurement

The impedance of the skin–electrode interface was analyzed by the impedance spectroscopy (LCR4235, Wayne Kerr Electronics Ltd., U.K). Two dry electrodes were placed on the forehead (F10) (4 cm apart), and then current was applied to the electrode pair to measure the impedance [26], [28]. The skin of the participant was once cleaned by gently wiping it with a 2-propanol impregnated cotton pad, which was allowed to evaporate before applying electrodes. In order to guarantee reliable and reproducible results, the test signal of the impedance spectroscopy was set to 1 V and the frequency range from 0.5 to 10 000 Hz. Nineteen tests were performed on five different subjects in this study. Fig. 3(a) and (b) shows the averaged values and standard deviations of impedance measurement results under different conditions. Here, in Fig. 3(a), the black line denotes the impedance of our dry foam electrode pair without skin preparation and conducting gel. Blue and red lines denote the impedances of conventional wet electrodes without and with skin preparation, respectively. All of the conventional wet electrodes were applied with conduction gel. The results showed that the impedance between the skin and our dry foam electrode is similar to that of the conventional wet electrode with conduction gel. Fig. 3(b) shows the impedance measurement on the hairy site (POz). It showed that, for our dry foam electrode, the impedance on the hairy site nearly equals that on the hairless skin, but that on hairless skin is even lower. Evidently, our dry foam electrode is soft enough to contact the skin properly, and the fabric layer is very stable.



(a) (b)

Fig. 2. (a) Equivalent circuit of a standard skin–electrode interface and (b) equivalent circuit of a dry skin–electrode interface [1].



(a) (b)

Fig. 3. Impedances of the skin–electrode interface on (a) forehead (F10) and (b) hairy sites (POz).

Fig. 4 shows the averaged values of the long-term impedance variation (5 h) for five subjects. The long-term impedance variation of the conventional wet electrode with conduction gel is more obvious than that of our dry foam electrode. The impedance variation of our dry foam electrode was observed in the range from 4k to 26k, and is in the acceptable range for normal EEG measurement [10], [28], [29]. Furthermore, our dry foam electrode can significantly provide better stability of the skin–electrode interface impedance because it does not need conduction gel, which is apt to drying.

B. Effect of the Contact Area of the Skin–Electrode Interface on Impedance

The effect of the contact area of the skin–electrode interface on impedance was investigated in this section. First, the skin–electrode interface impedances with different contact areas were tested. Fig. 5(a) and (b) shows the skin–electrode interface impedances with different electrode surface areas (10 mm $\times$ 7 mm, 14 mm $\times$ 8 mm, and 15 mm $\times$ 10 mm) on the forehead (F10) and hairy sites (POz), respectively. According to the results in Fig. 5(a) and (b), it is shown that the skin–electrode interface impedance will reduce both on the forehead and hairy sites when the electrode area increases. It is well known that the real contact area of the skin–electrode interface is proportional to the electrode area. Therefore, the impedance of a dry foam electrode indeed will reduce when the contact area of the skin–electrode interface increases [2], [10]. Next, the impedances of the conventional wet, fabric based, and our dry foam electrode with the same electrode area were tested. Fig. 6(a) and (b) shows the skin–electrode interface impedances between wet electrode, foam electrode (14 mm $\times$ 8 mm), and fabric-based electrode (14 mm $\times$ 8 mm) on the forehead and hairy sites. Our dry

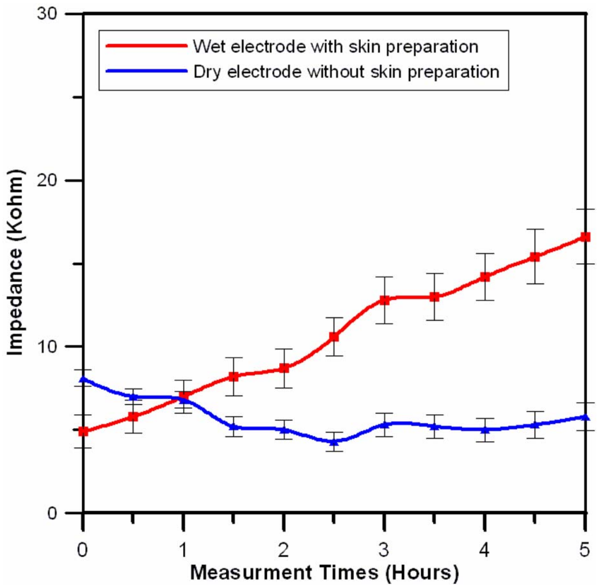


Fig. 4. Comparing the impedance variation between the wet and dry electrodes for long-term measurement on the forehead site (F10).

foam electrode obviously provides lower skin–electrode interface impedance than fabric-based and wet ones under the same electrode surface area condition [27]. This can be explained by the fact that the softness of the foam electrode substrate can help to adapt to the scalp effectively when the suitable force is applied. A larger skin–electrode contact area will lead to lower skin–electrode interface impedance [10], [20]; it is in agreement with our experimental results showing that the foam electrode has better performance than fabric-based one on both the forehead and hairy sites.

C. Comparison of Signal Quality Measured by Using Different Electrodes

The pretest experiment of signal-quality check, designed to understand distortion caused by our dry foam electrode under EEG measurement, is illustrated in Fig. 7. First, EEG data were prerecorded with 512-Hz sampling rates by using standard EEG electrodes with conduction gel, and stored in the personal computer. Next, the EEG data were fed into a programmable function generator and passed through a voltage divider to generate the simulated human EEG signal. The simulated EEG signal was further fed to our dry foam electrode, and then amplified by the EEG machine. After recording the amplified EEG signal, it was compared with the prerecorded EEG data. The signal quality can be presented by the correlation between the prerecorded EEG and EEG obtained by our dry foam electrode. In this study, the linear correlation coefficient function in MATLAB (R2007a, The MathWorks, Natick, MA) was used to evaluate the difference in EEG signal quality measured by using different electrodes. Fig. 8 shows the prerecorded EEG signal and its counterparts recorded by our proposed dry foam electrode. It showed that the averaged correlation between prerecorded EEG and EEG obtained by our dry foam electrode is high (96.65%).

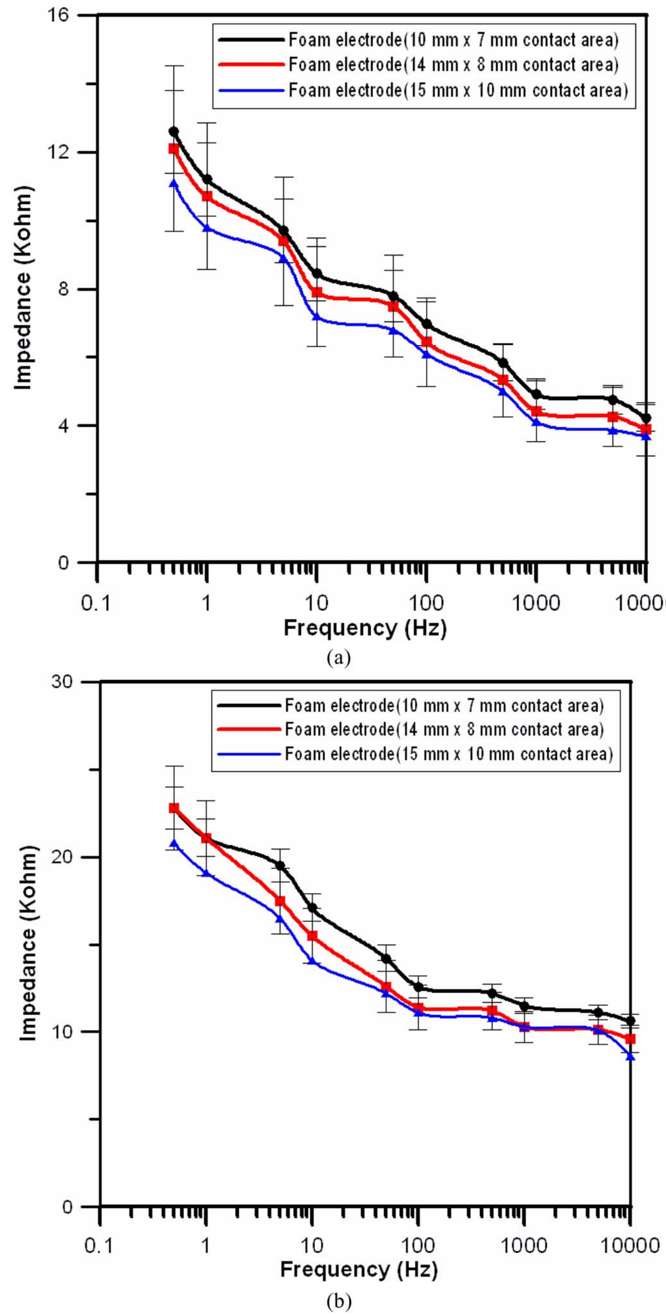
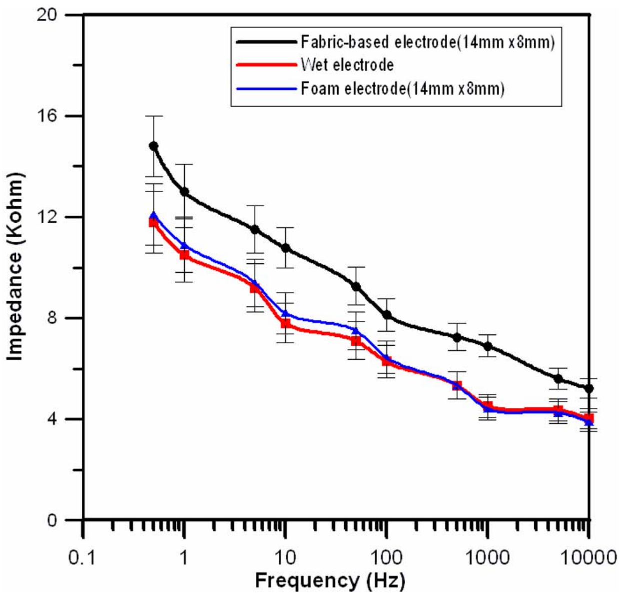
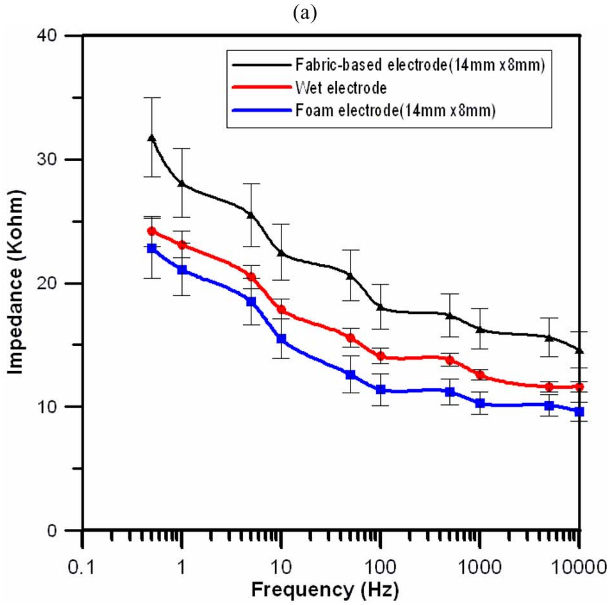


Fig. 5. Skin–electrode interface impedances of foam electrodes with 10 mm $\times$ 7 mm, 14 mm $\times$ 8 mm, and 15 mm $\times$ 10 mm electrode areas on the (a) forehead (F10) and (b) hairy sites (POz).

Next, the signal quality for the conventional wet electrode and our dry foam electrode was investigated. Fig. 9(a) and (b) shows the placements and the results of EEG measurement by using dry/wet electrode pairs in the locations of forehead (F10) and hairy sites (POz), respectively. Fig. 9(c) shows the placements and the results of electrooculography (EOG) measurement. The correlations between signals obtained by our dry foam electrode and the conventional wet electrode are typically in excess of 96.14% and 90.12% in the locations of forehead and hairy sites, respectively. For EOG measurement, the correlation is also very significant (in excess of 84.28%). Therefore, the signal quality



(a)



(b)

Fig. 6. Skin–electrode interface impedances of wet electrode, foam electrode (14 mm × 8 mm), and fabric-based electrode (14 mm × 8 mm) on the (a) forehead (F10) and (b) hairy sites (POz).

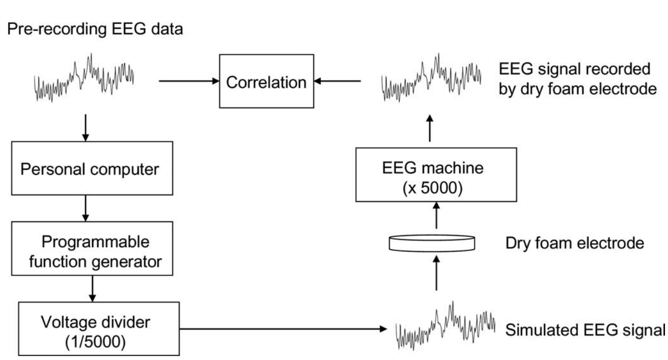


Fig. 7. Illustration of pretest experiment for signal-quality check of the proposed dry electrode.

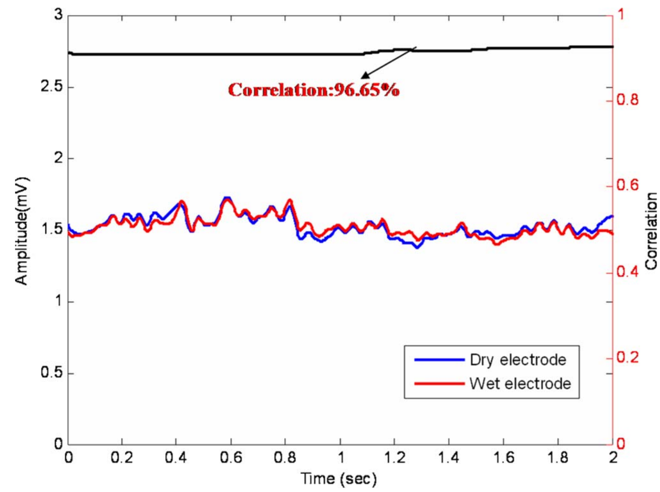
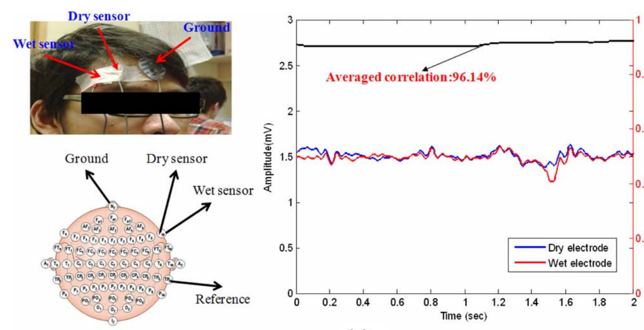
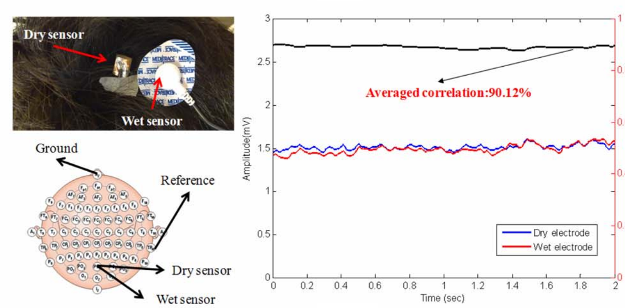


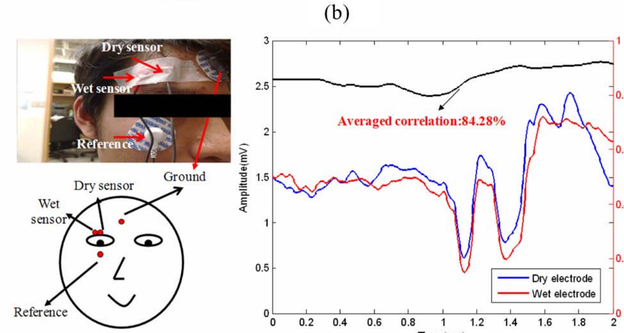
Fig. 8. Prerecorded EEG signal and its counterparts recorded by our dry foam electrode.



(a)

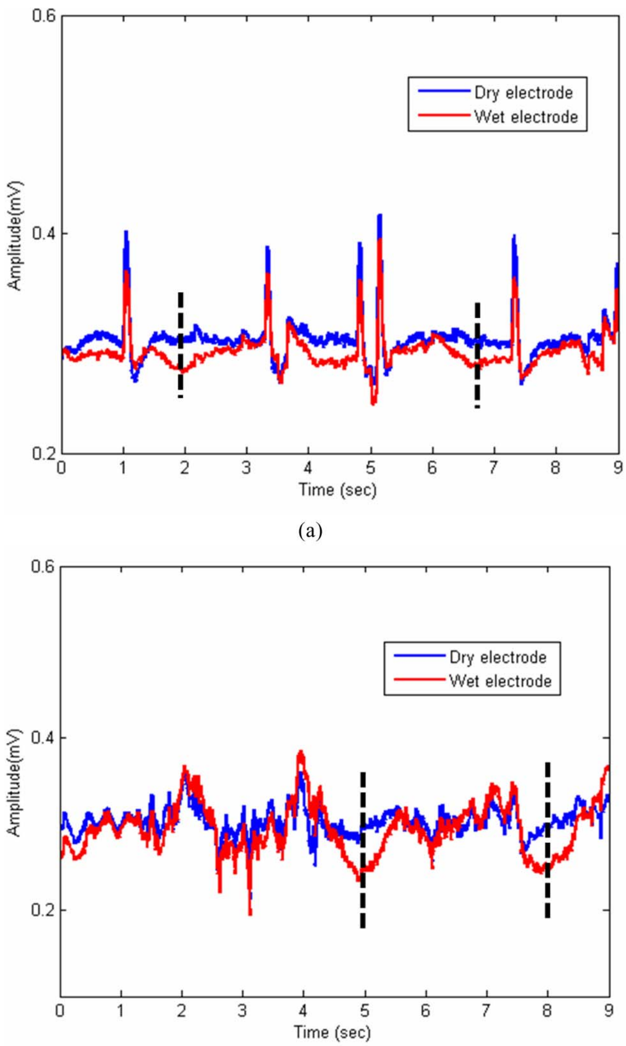


(b)



(c)

Fig. 9. Placements and results of (a) EEG measurement on forehead site (F10), (b) EEG measurement on hairy site (POz), and (c) EOG measurement.



(a)

(b)

Fig. 10. Influence of motion on EEG measurements for (a) forehead site (F10) and (b) hairy site (POz).

by using our dry foam electrode is almost identical to that of the conventional wet electrode. Based on this comparison, results showed that the proposed dry electrode (partly polarizable electrode) presents the quite different but effective way to acquire biopotentials than wet one (few-polarizable electrode). By the polarizable electric characteristic of our dry electrode, it can be regarded as a capacitive electrode, and can be used to measure biopotentials by the inductive method, not the ion change way [10].

D. *Influence of Motion Artifacts on Biopotential Measurement*

The motion artifacts were evaluated from EEG signals obtained by our wireless EEG acquisition system with different types of electrodes [10], [30], [31]. In this experiment, these electrodes were positioned on the forehead (F10) and hairy sites (POz) of the participant, and were close to each other, respectively. To evaluate the influence of motion on EEG measurement, the subject was instructed to increase the intensity of the walking motion until the EEG signal measured by the test electrodes was affected [10], [20], [29]. For the forehead site, the influence of motion for the conventional wet electrode is unobvious during first 15-min period; however, after 17 min, the conventional wet electrode presented more artifact than our dry foam electrode, though this may be dependent on the geometry of our dry foam electrode housing. Fig. 10(a) and (b) shows the EEG measurement on the forehead and hairy sites under the movement, respectively. It showed that the influence of motion of our dry foam electrode is significantly smaller than that of the conventional wet electrode, especially at the 2 and 6.5 s in Fig. 10(a) and at the 5 and 8 s in Fig. 10(b) with block dashed lines marked. This can be explained by the fact that our dry foam electrode can maintain the contact effectively even under the walking motion due to its mechanical softness and a better adhesion to the scalp. By attaching the dry foam electrode with a little pressure, its elasticity will stabilize the contact both horizontally and vertically. Wet electrodes suffered from moving charge artifact more than dry electrodes [1], [10], [20].

IV. CONCLUSION

A novel dry foam EEG electrode is developed, fabricated, and experimentally validated in this study. The major merits of this dry foam electrode include the following: 1) without skin preparation and conduction gel, it can be applied for long-term EEG measurement; 2) it is able to adapt to irregular scalp surface and even the hairy sites to maintain low skin–electrode interface impedance; and 3) the fabrication process is of low cost (about U.S.$ 0.3/unit). Experimental results showed that the proposed dry foam electrode can provide lower skin–electrode interface impedance than conventional wet ones on the hairy sites. Moreover, for long-term EEG measurement, our dry foam electrode can provide more stable impedance variation due to no drying influence of using conduction gel. Compared to other fabric-based electrodes, the softness of the foam substrate in our foam electrode can also increase the contact area of the skin–electrode interface to maintain lower impedance. The EEG signal quality acquired by using our dry foam electrode was consistent for all subjects, and the variation of EEG signal quality is very stable, even under motion. Overall, our proposed dry foam electrode provides potential for routine and repetitive EEG measurements, although its biocompatibilities still needed further validated. It indeed provides a novel prototype of a dry electrode for clinical and research applications.

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