A Novel 16-Channel Wireless System for Electroencephalography Measurements With Dry Spring-Loaded Sensors

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***Abstract*— Understanding brain function using electroencephalography (EEG) is an important issue for cerebral nervous system diseases, especially for epilepsy and Alzheimer's disease. Many EEG measurement systems are used reliably to study these diseases, but their bulky size and the use of wet sensors make them uncomfortable and inconvenient for users. To overcome the limitations of conventional EEG measurement systems, a wireless and wearable multichannel EEG measurement system is proposed in this paper. This system includes a wireless data acquisition device, dry spring-loaded sensors, and a size-adjustable soft cap. We compared the performance of the proposed system using dry versus conventional wet sensors. A significant positive correlation between readings from wet and dry sensors was achieved, thus demonstrating the performance of the system. Moreover, four different features of EEG signals (i.e., normal, eye-blinking, closed-eyes, and teeth-clenching signals) were measured by 16 dry sensors to ensure that they could be detected in real-life cognitive neuroscience applications. Thus, we have shown that it is possible to reliably measure EEG signals using the proposed system. This paper presents novel insights into the field of cognitive neuroscience, showing the possibility of studying brain function under real-life conditions.**

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***Index Terms*— Dry sensor, electroencephalography (EEG), electroencephalography measurement system, size-adjustable soft cap, wireless data acquisition device.**

I. Introduction

STUDYING brain function has become an important issue in neuroscience [1]–[3]. The electroencephalography (EEG) imaging technique is important for probing brain activation, and it is the most widely used technique in both basic neuroscience research [4]–[6] and clinical applications [7], [8] With the increased use of EEG, the requirements for EEG data acquisition devices [9] and signal processing methods have become more stringent [10]–[12]. The EEG-based brain-computer interface (BCI) [13] system provides a reliable and efficient means of communication between users and computers. This system has recently been introduced for neuroscience [5] and rehabilitation engineering [14] applications, including motor imagery [15]–[19], drowsiness detection [20], [21], and sleep analysis [22], [23].

Current EEG systems are not sized appropriately for real-life use, as their bulky size and wiring limitations restrict the available range of BCI experiments and the corresponding applications. In addition, conventional wet sensors are often used for EEG measurements, but these sensors require preparation of the skin or the application of conductive electrolytes at the skin-sensor interface, which can be time-consuming and uncomfortable for the user. Moreover, the conductive gel may cause a short circuit between nearby sensors when it is applied excessively, and in cognitive experiments, drying of the conductive gel in wet sensors can result in poor readings.

To overcome the limitations of conventional wet sensors, such as skin preparation, different types of dry sensors have been developed [24]–[32]. Some of these dry sensors are based on micro-electromechanical systems (MEMS) [26], [29], [30], [32], which acquire the EEG signals from the forehead [29]. There are several drawbacks using dry MEMS sensors, including the high manufacturing cost and the hard substrate, which is uncomfortable to wear. Other types of dry sensors are made using fabric-based sensors [33], [34], which are a more comfortable option than dry MEMS sensors. However, fabric-based sensor measurements are not suitable for use on hairy sites (i.e., parietal and occipital sites). Until now, dry sensors integrated with wearable and wireless EEG systems have not been available.

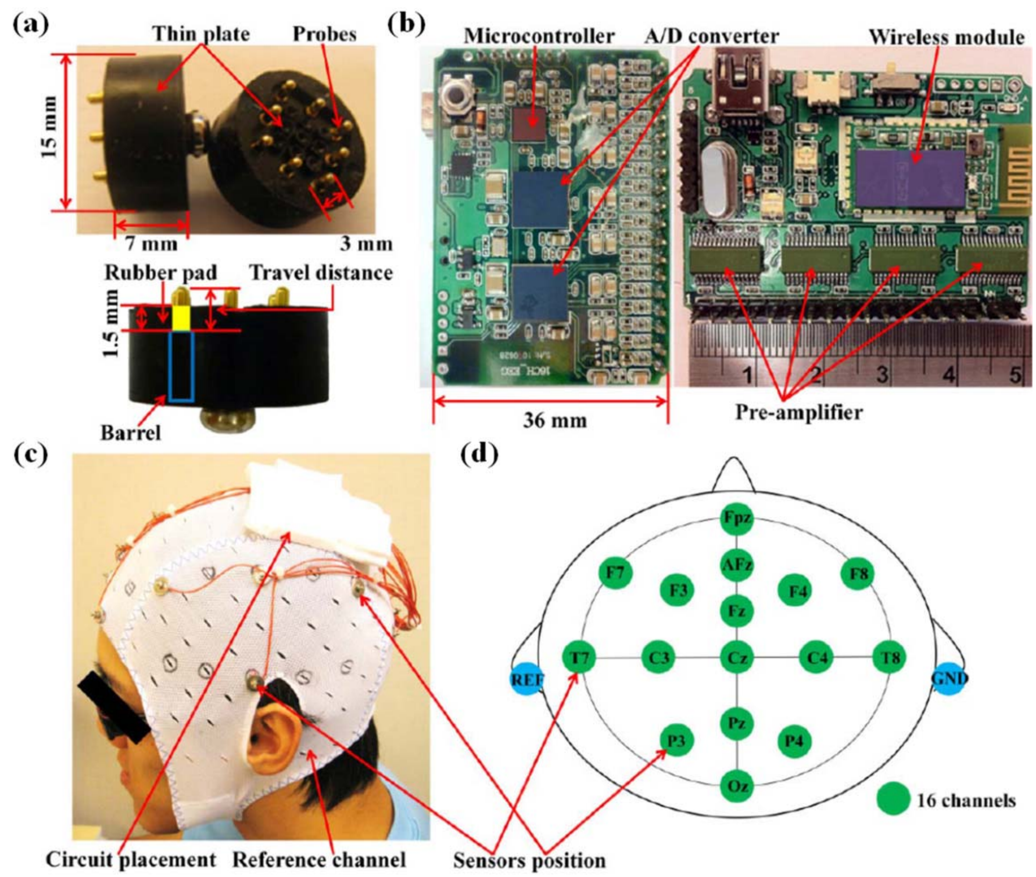

Fig. 1. Proposed design for the 16-channel EEG system with dry sensors. (a) Dry EEG sensor with a 15 mm diameter, a 7 mm depth and 8 probes. The travel distance of each probe is 3 mm. There is a unique rubber pad around the bottom surface of the sensors. (b) Wireless EEG acquisition system with a preamplifier, an ADC, a microcontroller, and a wireless module. Each circuit board is 36 mm in width. (c) Size-adjustable soft cap with 16 dry EEG sensors. The placement of each sensor is in accordance. (d) Standard 10-20 EEG system.

In this paper, a wearable, wireless 16-channel EEG system with dry EEG sensors was developed, consisting of dry spring-loaded sensors, a wireless acquisition system and a size-adjustable wearable soft cap. The dry sensors can be utilized without the application of a conductive gel, even on hairy sites. The sensors provide good electrical conductivity for effective acquisition of EEG signals. In contrast to traditional EEG measurement systems that use dry sensors, the proposed system requires reduced skin preparation and benefits from highly accurate EEG signals. Thus, the wireless and wearable EEG measurement system developed here has the potential to be used in cognitive engineering applications [35].

II. Materials and Methods

The fundamental components of the proposed system are shown in Fig. 1(a)–(d), including the dry spring-loaded sensors, a wireless EEG acquisition system, and a size-adjustable wearable soft cap, all in accordance with the international 10–20 system for sensor placements [35].

A. Design of Dry Spring-Loaded Sensor

The dry spring-loaded sensors were designed with eight probes, as shown in Fig. 1(a). These probes were designed to contact the skin and maintain electrical conduction: they are coated with gold on all surfaces to establish an electrical contact similar to that of conventional wet sensors. Building on our design from a previous paper [36], here, we propose the addition of a unique rubber pad around the bottom surface of the sensors, as indicated in Fig. 1(a). This pad can significantly reduce the pain when force is applied on the sensors. To test and demonstrate this design, a dry sensor composed of the probes, a spring, a plunger, a barrel, and the rubber pad was constructed. The top of the probe has a spheroid shape and is coated with gold to enhance the conductivity. Gold is chemically stable, biocompatible, and does not easily react with other substances. Moreover, gold's high conductivity, high resistance to oxidation and resistance to environmental degradation (i.e., resistance to other non-chlorinated acids) justify the extensive use of gold materials in the electronics and biomedical industries. The spring force of the sensor was $\sim$23 g, which is the level required for EEG signal measurements on the scalp [36]. Depending on the location of spring contact with the scalp, the spring could either increase or decrease in length.

In contrast to conventional wet sensors, dry sensors exhibit the electronic characteristics of electrically conductive materials. They obtain high-quality signals without skin abrasion or

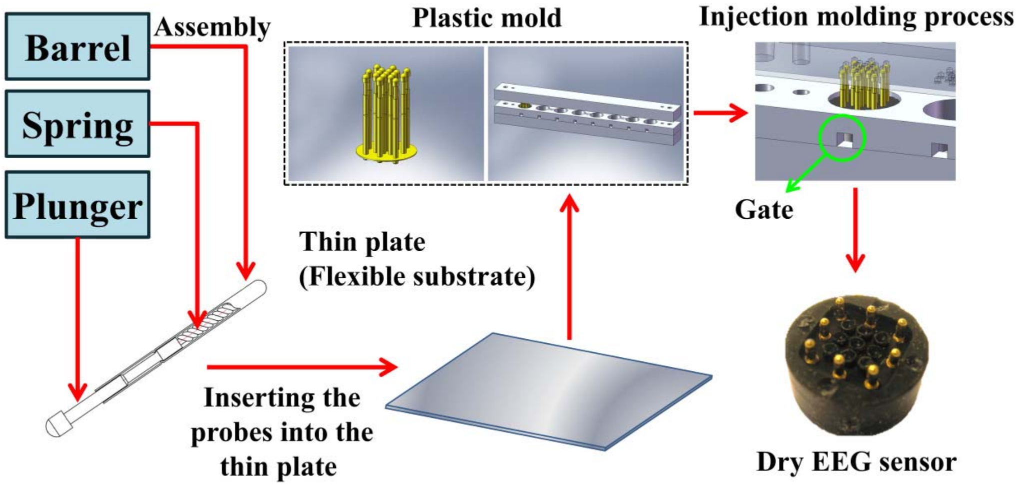

Fig. 2. Assembly process for the dry sensors, including injection-molding and packaging processes.

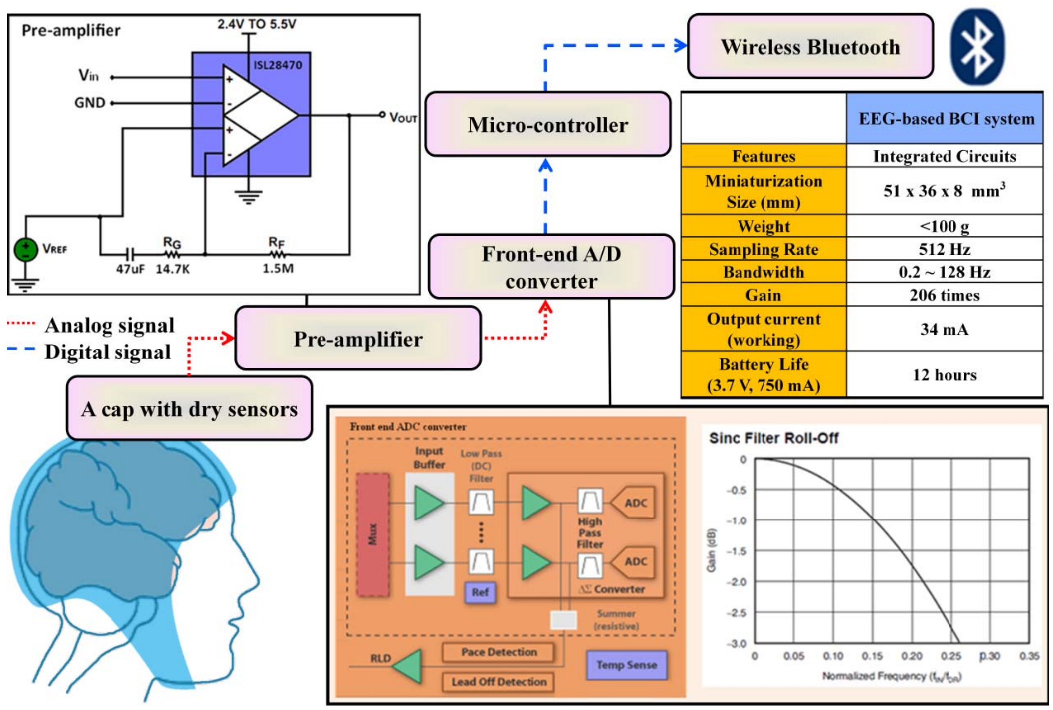

Fig. 3. Wireless and wearable 16-channel EEG system with dry sensors.

preparation. Moreover, unlike fabric-based sensors [37], [38], the spring-loaded sensors allow a high level of geometric conformity between the sensor and the irregular scalp surface due to the flexibility of the probes when applied to the scalp. This flexibility also can increase the skin-sensor contact area on hairy sites.

B. Manufacturing of Dry EEG Sensors

The manufacturing process for the dry EEG sensors is shown in Fig. 2. Eight probes are inserted into a piece of thin copper plating that is applied to the flexible base of the sensor. After insertion, the eight probes on the copper plate are all conductive. When force is applied to the sensor, the flexible substrate permits high geometric conformity to the irregular scalp surface. The spring provides buffering effects, enabling the dry EEG sensor to contact the scalp when force is applied. The flexibility of the spring increases the comfort when the sensor contacts the scalp. After fabricating and inserting the probes into the flexible substrate, an injection-molding process is used to integrate the flexible base with several probes. The probes with the elastic base are fixed into the plastic mold. Similar to the thin plate and spring contact probes, the sensors also remain flexible after the injection molding process [36].

C. EEG Acquisition Module

A typical EEG signal ranges from 10 to 100 μV in amplitude when measured from the scalp. EEG signals measured through sensors on the scalp are easily affected by

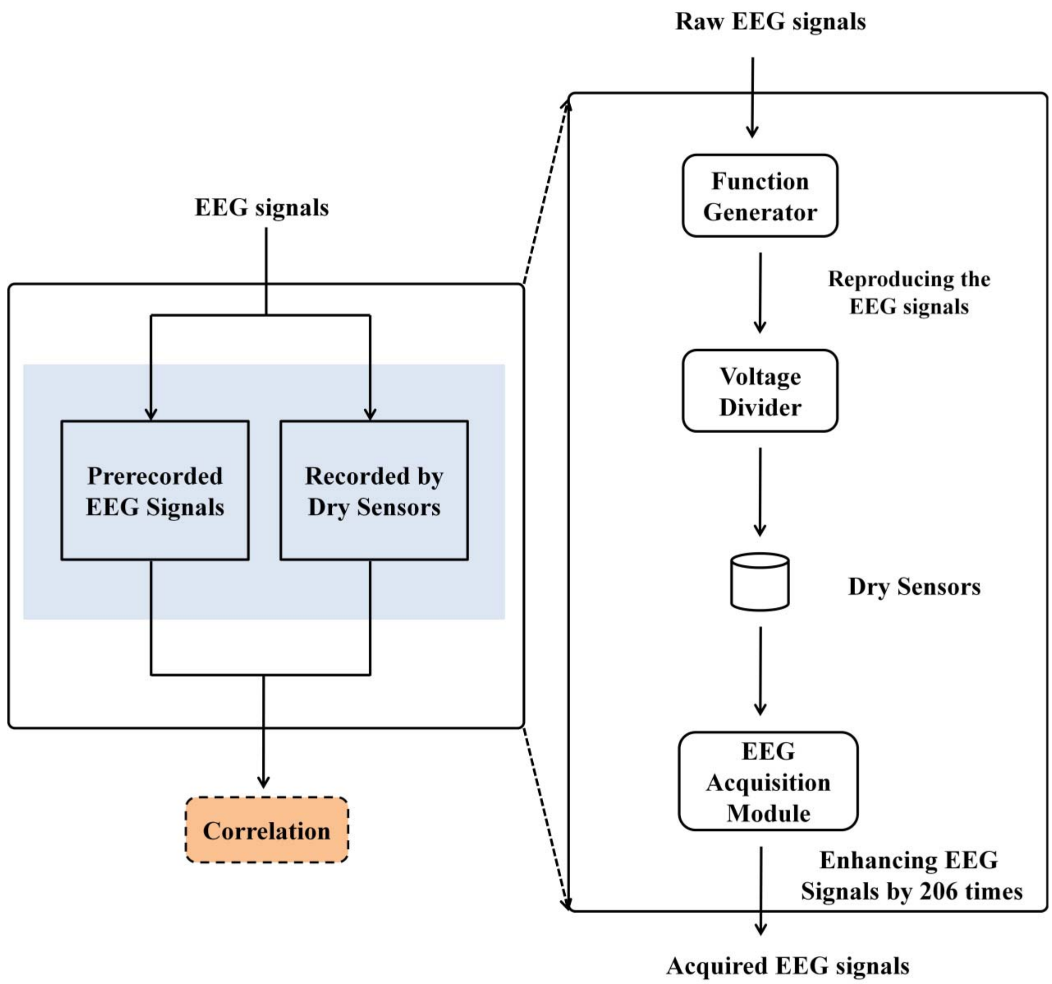

Fig. 4. Testing the accuracy of the signal from the dry sensors.

artifacts indirectly related to brain activation [39], [40], such as electromyography and electrooculography. These artifacts are irrelevant physiological signals in this experiment and may significantly obscure the EEG signals of interest. The 16-channel EEG acquisition module was designed to measure true EEG signals, as shown in Fig. 1(b). The acquisition module consists of four major units: 1) a preamplifier unit; 2) a front-end analog-to-digital converter (ADC) unit; 3) a microcontroller unit; and 4) a wireless unit. The wireless 16-channel integrated circuit-based acquisition module described here measures approximately $51 \times 36 \times 8$ mm^3 and can be embedded into our system. When measured by the dry EEG sensors, EEG signals are first amplified by the preamplifier unit (ISL28470, Intersil, USA), which amplifies the voltage difference between the reference and EEG electrodes and simultaneously rejects common-mode noise (i.e., power line noise). An instrumentation amplifier was used as the preamplifier because of its extremely high input impedance and high common-mode rejection ratio (CMRR). The instrumentation amplifier improves the CMRR and amplifies the EEG signals such that microvolt-level signals can be detected successfully.

The gain of the preamplifier unit is set to 103 V/V, and the cut-off frequency is regulated to 0.2 Hz by a high-pass filter. The transfer function of this preamplifier circuit is as follows:

$$V_{\text{out}} = \left(1 + \frac{R_F}{R_G + 1/sC}\right) V_{\text{in}} \tag{1}$$

$$\frac{V_{\text{out}}}{V_{\text{in}}} = \left(1 + \frac{R_F}{R_G + 1/sC}\right) \tag{2}$$

$$\frac{V_{\text{out}}}{V_{\text{in}}} = \left(1 + \frac{R_F}{R_{\text{eq}}}\right) = \left(1 + \frac{1.5 \times 10^6}{14.7 \times 10^3 + 1/j\omega \times 47 \times 10^{-6}}\right). \tag{3}$$

The preamplifier circuit, shown in Fig. 3, has two amplifiers: one that is connected to the input voltage (V_{in}) and the ground and another that is connected to the feedback of V_{out} and reference voltage (V_{REF}). Thus, using the superposition theorem [41], [42], the transfer function of the preamplifier circuit is as shown in (1). The values of the transfer function (e.g., $R_{\text{F}} = 1.5$ MΩ, $R_G = 14.7$ KΩ and equivalent impedance of 47 μF) are shown in (2). Equation (2) can be reorganized into the form of a high-pass filter with input signals of frequency ω, as presented in (3). The high-pass filter is regulated to 0.2 Hz and consists of a resistor (resistance R_G) and a capacitor connected in series. Therefore, the gain of the preamplifier unit is 103 V/V [i.e., $(1 + 1.5 \times 10^6/14.7 \times 10^3)$].

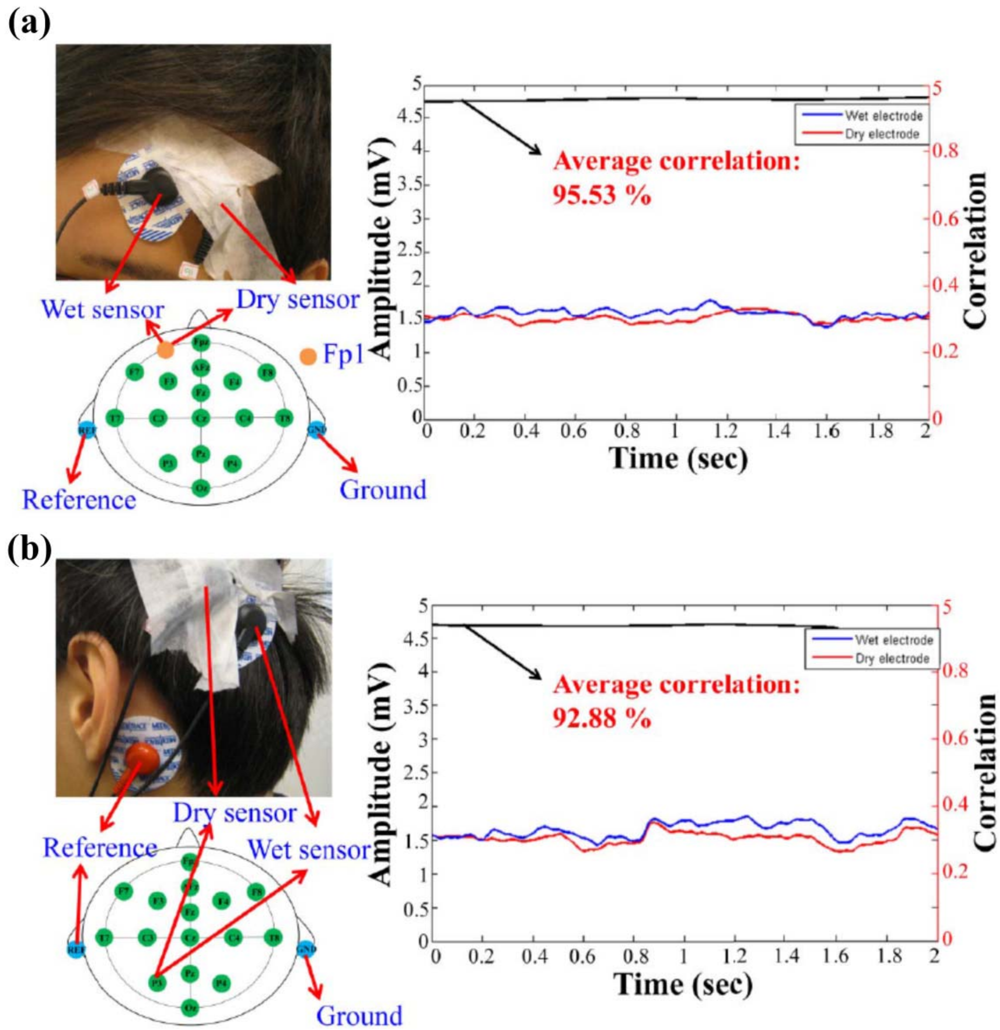

Fig. 5. Comparison of the signal quality between the dry and wet sensors. EEG measurements from (a) frontal sites (Fp1) and (b) hairy sites (P3) are shown.

The front-end ADC (ADS1298, Texas Instruments, USA) is used to digitize the amplified EEG signal. The minimum input voltage of the ADC ranges from −1.94 to 1.94 mV, and the maximum ranges from −23.30 to 23.30 mV. The least significant bit voltage is 0.286 μV. The simplified design of this system reduces the space requirements and power consumption compared to other systems. The front-end ADC digitizes the analog EEG signals with a sampling rate of 512 Hz, and a sinc filter removes the frequencies above 128 Hz, as shown in Fig. 3. The microcontroller unit (MSP430F5522, Texas Instruments) was used to regulate the signal sampling rate, magnification, and noise reduction. The processed EEG signal from the ADC was reduced to 60 Hz noise by the microcontroller unit using a moving average. The microcontroller unit set the default gain of the ADC unit to 2 V/V. Therefore, the total gain of the EEG signal was set to 206 V/V (i.e., 103 $\times$ 2 V/V). Adjusting the gain of the ADC unit to the maximum (12 $\times$), the total gain of the EEG signal is 1236 V/V (i.e., 103 $\times$ 12 V/V). After removing the noise and amplifying the EEG signal, the EEG signal was transmitted to the computer interface by a wireless module, specifically a Bluetooth module (HL-MD08R-C2, HotLife Electronic Technology Co., Ltd., Taiwan). The Bluetooth module supports a high band-width transmission with its high baud rate (i.e., 921 600 b/s), according to the Bluetooth v2.1+ enhanced data rate specification. Power for the board is supplied by a commercial 750 mAh Li-ion battery with a 3 V output voltage, which can also supply power for the EEG acquisition circuit and can be continuously operated for over 12 h.

D. BCI System

Standard EEG systems have multiple channels (i.e., 64 or 128 channels) available for measuring brain activity, with sensors organized on an elastic head cap according to the international 10–20 [43] system. Such a cap is suitable only if the sensors are covered with a conductive gel. To solve this problem, an easy-to-use, size-adjustable soft cap with dry sensors is proposed here. The EEG size-adjustable soft cap is fitted with 16 dry sensor sites, as shown in Fig. 1(c). The cap is composed of an elastic material, providing a more comfortable fit and more flexibility, enabling the experimenter to place the sensors in close contact with the user's scalp, which is

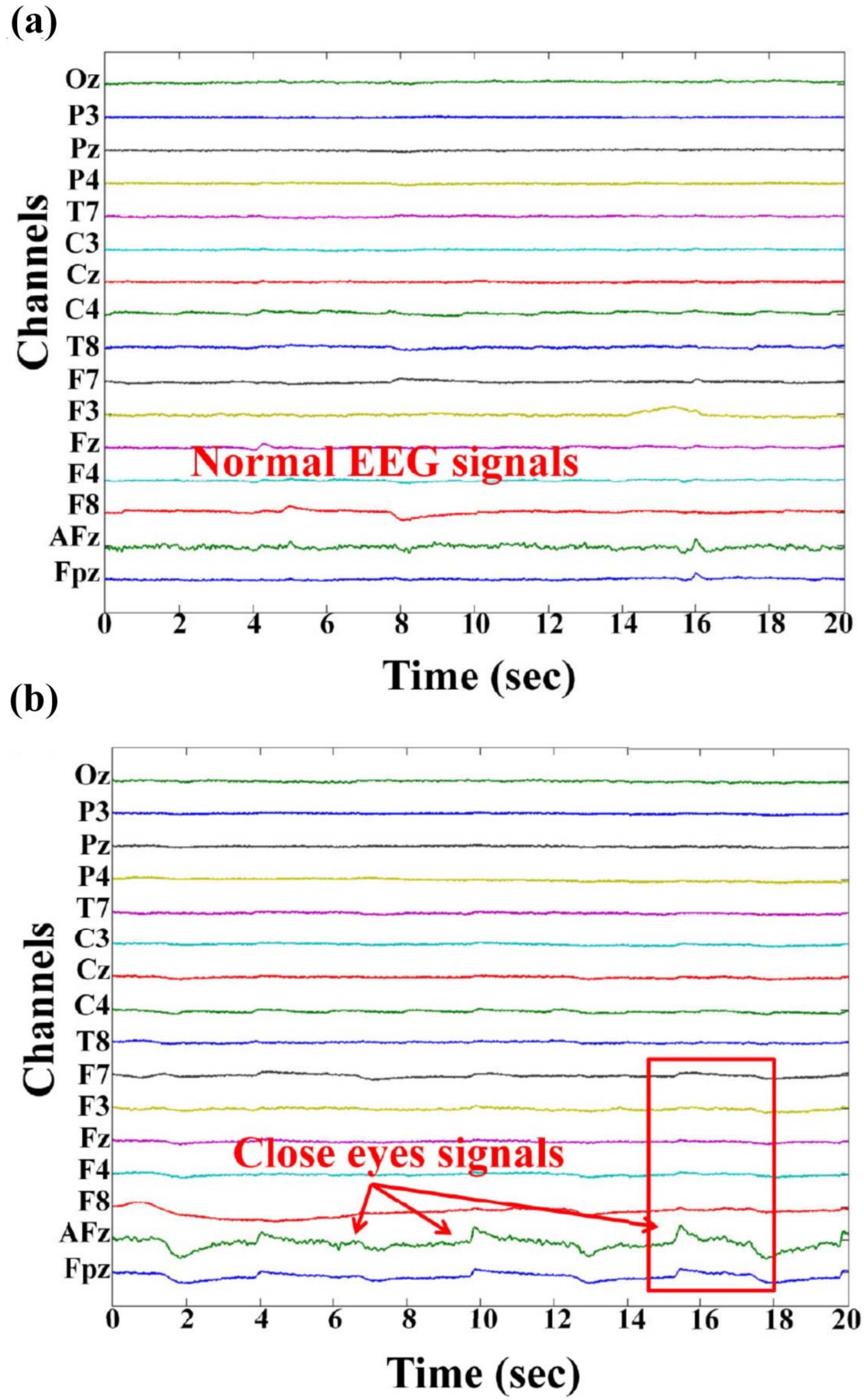

Fig. 6. 16-channel EEG system was used to measure EEG signals from hairy sites using the dry sensors. The data show measurements of (a) normal EEG signals and (b) EEG signals made with the eyes closed.

typically an irregular surface. The inner layer of the cap holds in place the universal joints that connect to the dry sensors on the scalp. This arrangement provides multiple angles of contact with the scalp surface, thus providing stable EEG signals. The outer layer of the cap, comprised of elastic fiber and Velcro, provides great flexibility for covering the heads of various users. The 16 dry sensors are located on the cap according to international 10–20 system, as shown in Fig. 1(d), with

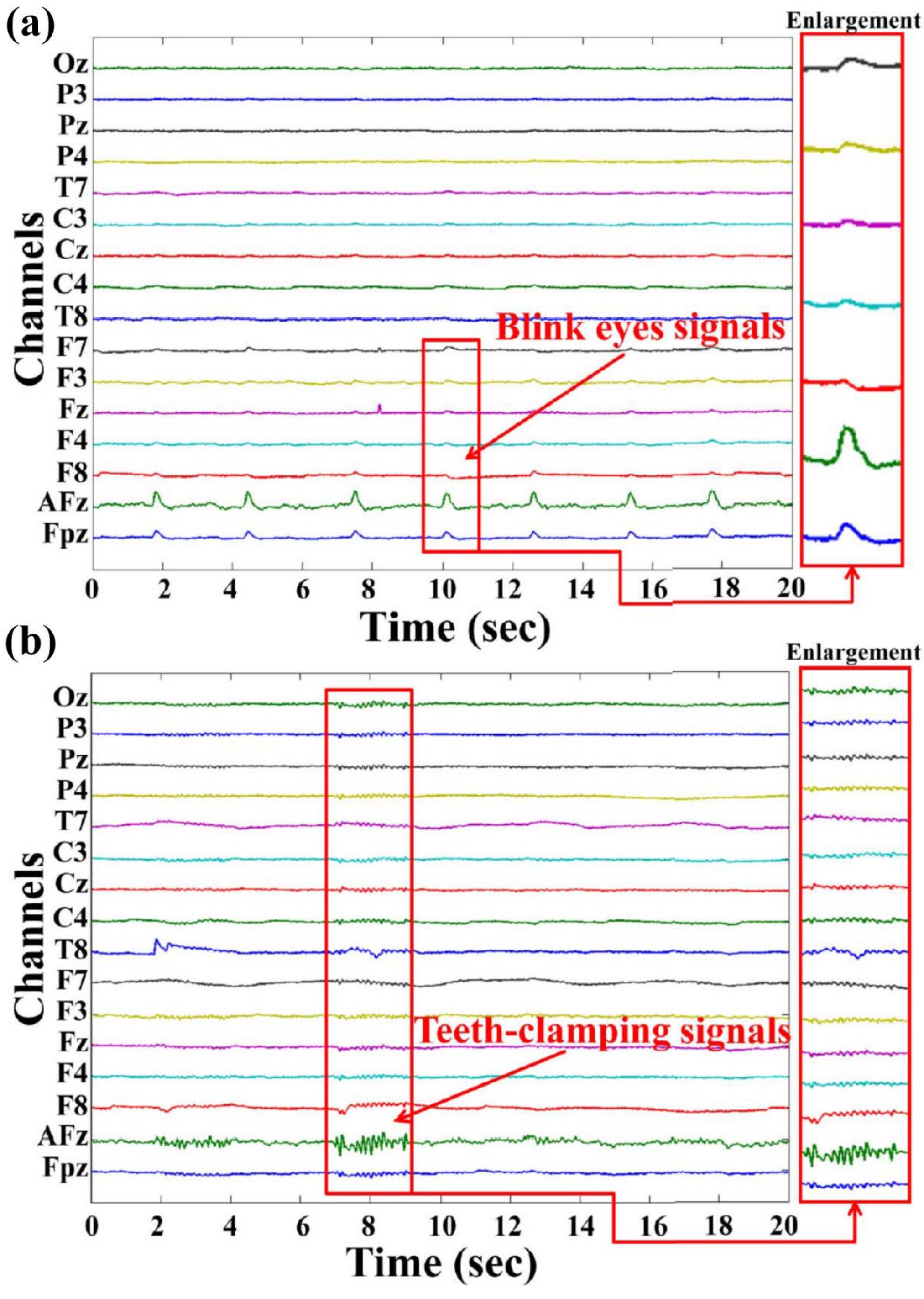

Fig. 7. 16-channel EEG system was used to measure EEG signals on hairy sites using dry sensors. The data show measurements of (a) signals during an eye blink and (b) signals during teeth clenching.

sites Fpz, AFz, F8, F4, Fz, F3, F7, T7, T8, C4, Cz, C3, P4, Pz, P3, and Oz included.

III. Results and Discussion

The experiments presented here consisted of three major stages. In the first stage, a validation experiment was used to verify the signal quality, as shown in Fig. 4. EEG data were prerecorded using a conventional EEG electrode with a conductive gel. These data were fed into a programmable function generator and passed through a voltage divider, thus generating simulated human EEG signals. The simulated EEG signals were then fed to a dry electrode, and the output data of the dry electrode were recorded. prerecorded data were used to provide a set of standard EEG patterns for repeated experiments so that the performance of the dry electrodes could be objectively evaluated [9], [36], [38]. Therefore, the physiological meaning of the prerecorded EEG data was not interpreted except to validate the proposed dry sensors. The aim of this validation process was to identify any distortion caused by the dry EEG sensor during EEG measurements.

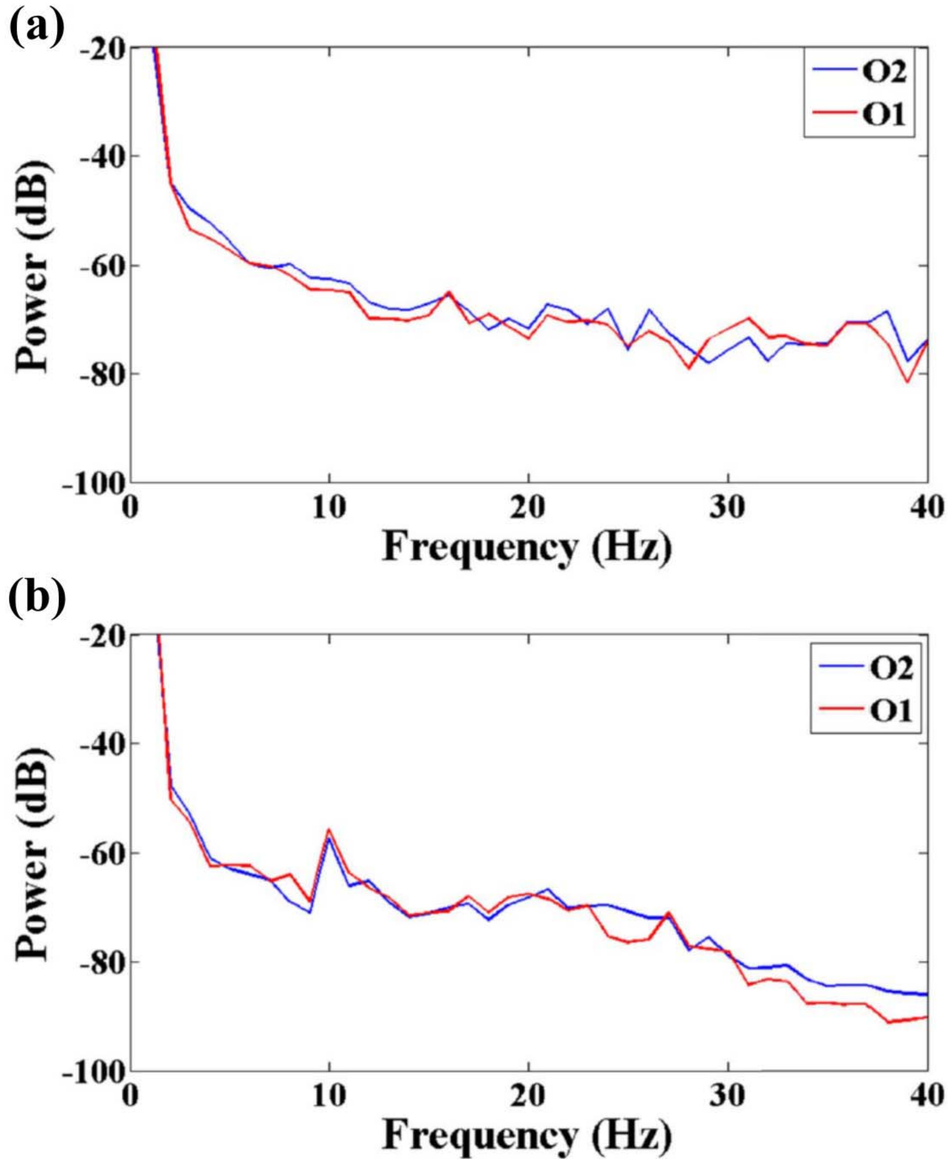

Fig. 8. Results showing the difference between the normal state and the eyes closed state. (a) Subject at rest, showing normal EEG signals from the O1 and O2 channels. (b) Subject with the eyes closed, showing alpha activity in the EEG signal measured from the O1 and O2 channels.

In the second stage, a user sat comfortably in front of a monitor wearing both dry and wet sensor simultaneously. The correlation between the conventional wet EEG electrode and the dry EEG sensor was investigated. Finally, after demonstrating the precision of the signals measured by the dry EEG sensors through the circuit, the newly developed wireless and wearable EEG cap with 16 dry sensors was used to measure a normal EEG, an EEG with the eyes closed, an EEG during an eye blink, and an EEG during teeth clenching, without the use of the conductive gels or skin preparation.

Fig. 4 shows the design of the validation experiment to test the signal quality of the dry sensors. EEG signals were prerecorded using wet electrodes as described above and then transmitted to the data acquisition device. The secondary EEG signals that were recorded by the dry sensors were also transmitted, and the correlation between the signals from the dry and wet sensors was determined. The prerecorded EEG signals and the signals from the dry EEG sensor were highly correlated at 96.83%.

Fig. 5(a) shows the results of the simultaneous EEG measurements made using both dry and wet sensors located on the forehead (site Fp1). The EEG signals recorded by the wet and dry sensors were highly correlated at 95.53%. In addition to this correlation, the data show that the signal quality from the dry sensor and readout circuit was stable and reliable compared to the wet sensor. Fig. 5(b) shows the results of EEG measurements made using the wet and dry sensors on a hairy site (P3). The correlation of 92.88% on a hairy site is significant.

According to these experimental results, the 16-channel dry sensor system described here can be used for measuring EEG signals with high signal quality, especially on hairy sites. We next measured a series of EEG signals: normal signals, eye-blink signals, signals with the eyes closed, and signals due to teeth clenching. The normal EEG signals that were measured by the proposed system are shown in Fig. 6(a). The EEG signals could be observed from frontal (i.e., Fpz, AFz, F8, F4, Fz, F3, and F7), temporal (i.e., T7 and T8), central (i.e., C4, Cz, and C3), parietal (i.e., P4, Pz, and P3), and occipital (i.e., Oz) brain regions. Due to the scaling of the plot in the figure, the signal variations appear relatively small, but the raw EEG data were clear and reliable. EEG signals with the eyes closed were also measured by the proposed system, as shown in Fig. 6(b), and were detectable at the frontal sites (i.e., Fpz, AFz, F8, F4, Fz, F3, and F7). In this measurement, the alpha wave was larger. Thus, the signals obtained from the Fpz, AFz, F8, F4, Fz, F3, and F7 sites were more significant than those obtained from the temporal (i.e., T7 and T8), central (i.e., C4, Cz, and C3), parietal (i.e., P4, Pz, and P3), and occipital (i.e., Oz) areas. Fig. 7(a) shows the 16-channel EEG-system measurement of signals during an eye blink. Because the motion of blinking occurs physically near the frontal area, the signals from blinking eyes are significant in the frontal zone (i.e., Fpz, AFz, F8, F4, Fz, F3, and F7). Therefore, during an eye blink, the signals were more obvious on the frontal site relative to other sites (i.e., central, temporal, parietal, and occipital). Fig. 7(b) shows the signal due to teeth clenching, during which the whole head (i.e., frontal, central, temporal, parietal, and occipital) had significant signal variations. Fig. 8(a) and (b) shows the power spectra of the EEG data collected by the dry sensors in this paper, showing characteristic low frequency bands (1–30 Hz). The EEG activity from a subject at rest [Fig. 8(a)] shows the activated reactions caused by holding the eyes open for a few seconds. Because the general alpha frequency band of the EEG signal is distributed between 8 and 12 Hz, the experimental results in Fig. 8(b) fit the trend in the alpha domain.

Here, we have shown positive results from measuring EEG signals with the proposed system and its dry sensors. Our experimental results have shown that dry sensors are capable of recording EEG signals via the EEG measurement system. The signal correlation between measurements performed with dry and wet sensors at the same locations was high. These results are significant with respect to the EEG measurement system because the dry sensors can be utilized without using conductive gel on hairy sites. In addition, these sensors can effectively acquire EEG signals (i.e., normal, closed eyes, blinking, and teeth-clenching signals) in frontal (i.e., Fpz, AFz, F8, F4, Fz, F3, and F7), temporal (i.e., T7 and T8), central (i.e., C4, Cz, and C3), parietal (i.e., P4, Pz, and P3), and occipital (i.e., Oz) areas. In contrast to traditional EEG measurement systems, the use of dry sensors allows users to feel more comfortable and experiments to be performed more quickly.

IV. Conclusion

In this paper, a wearable EEG system with dry spring-loaded sensors is proposed to transfer the EEG signals wirelessly to the computer. The developed system contains a size-adjustable soft cap, dry spring-loaded sensors, and a 16-channel acquisition circuit. The experimental results show that the proposed EEG measurement system with dry sensors can provide good signal quality on hairy sites compared to conventional wet sensors. Unlike the conventional system with wet sensors, the proposed system can be used to measure EEG signals without the use of conductive gel and skin preparation processes. Due to the soft substrate in the dry sensors and the spring-loaded probes, the design ensures that the dry sensors fit on the scalp tightly. The soft cap is suitable for different head sizes (i.e., small, medium, or large) for basic cognitive experiments. The quality of the EEG signal measured with the dry sensors approached that of the signal quality from the wet sensors. Thus, researchers can use the EEG system with dry sensors developed here to reliably investigate human cognitive states in real-life conditions.

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