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Loop Fabric EEG Textrode for Brain Activity Monitoring

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Abstract. Recently, metallic dry EEG electrodes have been introduced to overcome the limitation of wet electrodes, as the conductive gel used causes skin irritation and dries out over time. However, the metal dry EEG electrodes have a high weight and a stiff structure that limits them from wearable application. In this work, we have developed a textile-based EEG electrode (textrode) from silver-plated polyamide loop fabric washable up to 100 cycles. The new EEG textrode collects quality signals comparable to commercial dry Ag/AgCl EEG electrodes. The signals were detected at all major EEG bandwidths. In addition, the new textrodes showed a lower skin-to-electrode impedance (-19.23%) than the commercial dry electrode and a higher signal-to-noise ratio (+27.4%). Therefore, these novel textile-based electrodes can be used to monitor brain activity for wearable applications, especially when long-term monitoring is required.

1. Introduction

The invention of the high-conducting yarns approaches, which can be applied to any textile technology, brought novel electrode types for a wide range of technical and medical applications. Textile electrodes have been successfully developed to recognize bioelectric impulses from the human body. A typical example is the emergence of Electroencephalogram (EEG) textile electrodes to monitor brain activity.

EEG is a method for monitoring the brain by placing metal electrodes on the scalp which measure electrical potentials that arise outside of the head due to neuronal action within the brain [1]. The commercial gel-dependent metallic electrodes used for EEG acquisition can cause skin irritation. Moreover, the gel gets dehydrated over time which leads to an increase in skin-to-electrode impedance and more noise. To overcome the limitations of wet electrodes dry metal discs and needle spikes were recently introduced. Dry electrodes had several advantages in comparison with wet ones, such as the electrode–skin interface impedance, signal intensity, and size of the electrode [2]. However, their weight and flexibility could not be still suitable for long-term monitoring and wearable purposes [3–5]. This caused an emergence of textile-based EEG electrodes like the copper-plated textile fabric EEG electrode that gives similar signals to standard wet EEG electrodes [6] and knitted soft textile electrodes for EEG monitoring made from nylon, conductive fibers, spandex, and polypropylene [7]. But, there was no evidence reported that showed the aforementioned textile-based EEG electrodes retained their textile texture after the electrode construction [8]. In earlier research [4,5], we produced a textile-based EEG electrode made of PEDOT:PSS/PDMS-printed cotton fabric that has the properties of ordinary textiles. The stability of the PEDOT:PSS, on the other hand, was proven to be a concern over time. Here, we

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have developed the EEG electrode directly from a conductive textile loop fabric which is washable and flexible. The electrical and physical property of the electrode is the same as the fabric. In this paper, the acronym ‘textrode’ is used for textile electrodes.

2. Materials and Methods

A washable, up to 100 cycles, electro-conductive silver-plated polyamide Velcro® loop tape with a surface resistance of 1.4 Ω/sq (obtained from Ardafruit, USA) was used to construct a 2 cm diameter circular EEG textrode (Figure 1a). The textrode was designed using a layered structure by placing the loop fabric on foam to ensure the conductive fabric is sufficiently pressed against the skin on the active sides of the electrode. An elastic bandage was also used to keep the electrodes from slithering.
Eight textrodes were placed on the frontal, temporal and parietal (Fp1, Fp2, F7, F8, T3, T4, A1, and A2) according to the 10-20 EEG placement system (Figure 1b); the 10 and 20 in the system means the distance between the textrodes with respect to the dimension of the head. The numbers in the indicated positions represent the corresponding location of the forehead. Even numbers denote the right location and odd numbers the left. The diagram in Figure 1a illustrates the position of the textrodes. EEG signals were collected using an OpenBCI Board assisted by OpenBCI GUI software conducting 5 minutes of static EEG measurement as shown in figure 1b at 30 fps, and 1-50 Hz BP Filt. 2457.759. An OpenBCI Board with a built-in ADS1299 [9] for impedance testing was also used to assess skin-to-electrode impedance according to Tseghai et al. [10], shown in Figure 1c. Besides, a textile head phantom was used to determine the signal-to-noise ratio (SNR) of the textrode according to Tseghai et al. [11], shown in Figure 1d. The SNR was calculated using (1).

$$\mathrm{SNR\ (dB)} = 2\log\left(\frac{\mathrm{DVS}}{\mathrm{UVS}-\mathrm{DVS}}\right) \tag{1}$$

Where, DVS and UVS undenoised and denoised voltage signals, respectively.

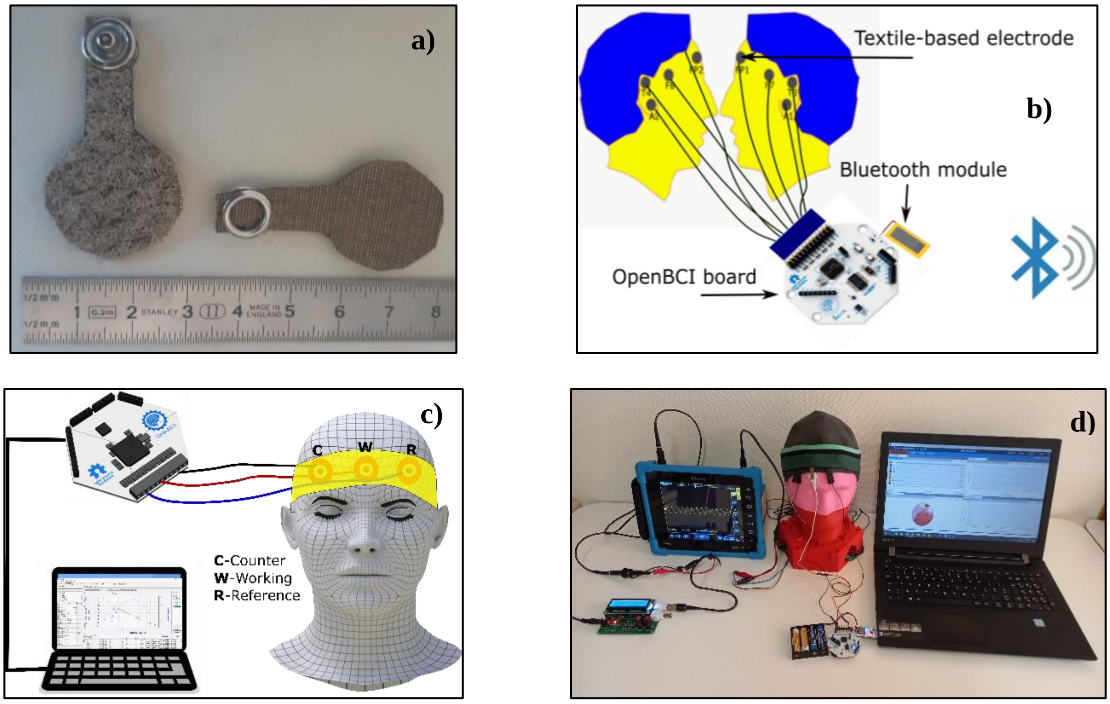

Figure 1. (**a**) EEG textrode; (**b**) textrode placement and EEG measurement; (**c**) skin-to-electrode impedance measurement; (**d**) synthetic wave generation and injection to head phantom to assess the signal-to-noise ratio (SNR)

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3. Results and discussion

3.1. Skin-to-electrode impedance

The loop fabric EEG electrode gave a significantly smaller skin-to-electrode impedance than the dry Ag/AgCl electrode where the f-ratio value is 39.74 and the p-value is less than 0.01 at a 95% confidence interval according to One-Way ANOVA. The average skin-to-electrode of the loop fabric textrode in four minutes measurement was found 2457.76 Ω (-19.23%), R2 = 0.96 as shown in Figure 2. This value is much lower than the impedance required to collect bio-signal i.e. less than 5000 Ω [12]. The impedance of the textrode linearly drops with time which could be due to sweat generation. The linear equation that shows the impedance with respect to time is shown in (2). These results confirm that the developing EEG electrodes to monitor brain activity from such conductive textile loop fabric could potentially replace the gel-based standard Ag/AgCl wet electrodes and commercial dry electrodes.

$$Z = -68.354t = 2765.4 \tag{2}$$

Where Z = skin-to-electrode impedance (Ω) and t = time (s)

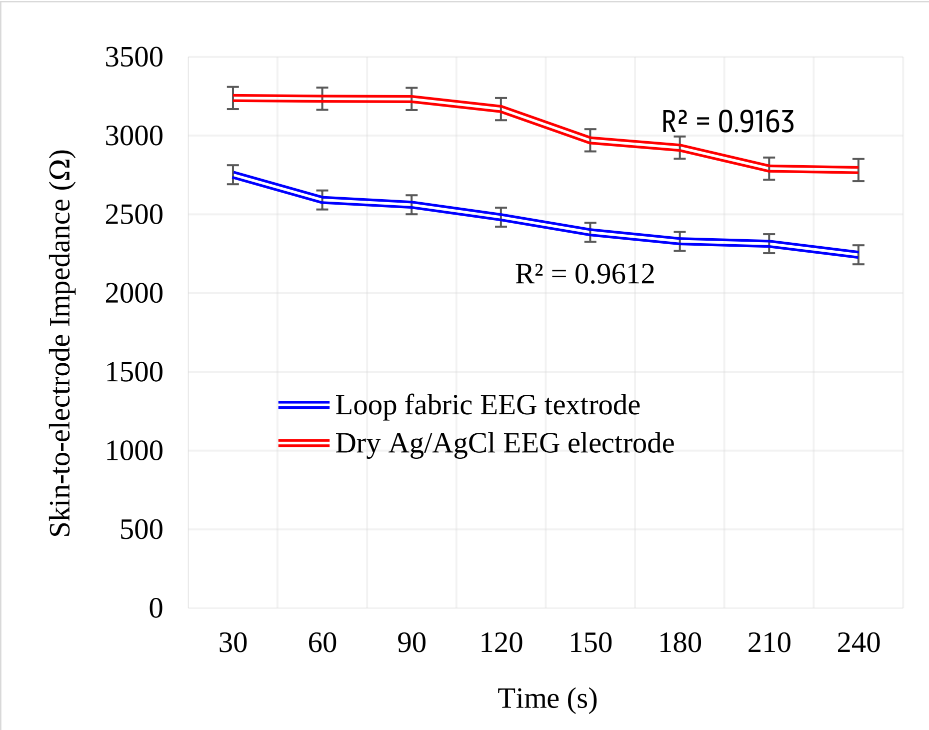

Figure 2. skin-to-electrode impedances at alpha bandwidth

3.2. Electroencephalogram Signal

An EEG waveform with 2.03 µVrms (+19.4%) to 50.5 µVrms (+27.5%) peak-to-peak voltage has been collected from the textrode during 5 minutes of static EEG measurement using an OpenBCI board at 30 fps, and 1-50 Hz BP Filt. The waveform acquired from the dry Ag/AgCl electrode at the same parameters was 1.70 µVrms to 39.6 µVrms. Therefore, the EEG signal quality obtained by the textrode is strongly comparable to the standard electrode. The EEG signal was also clear at an amplitude of 0.1 µV to 1 mV and an EEG band frequency of 1 to 60 Hz. The real-time trace of EEG and Fast Fourier Transform (FFT) plots from the textile electrode, amplitude at each frequency, and the ten-second-average Vrms amplitudes are shown in Figures 3a, 3b, and 3c, respectively.

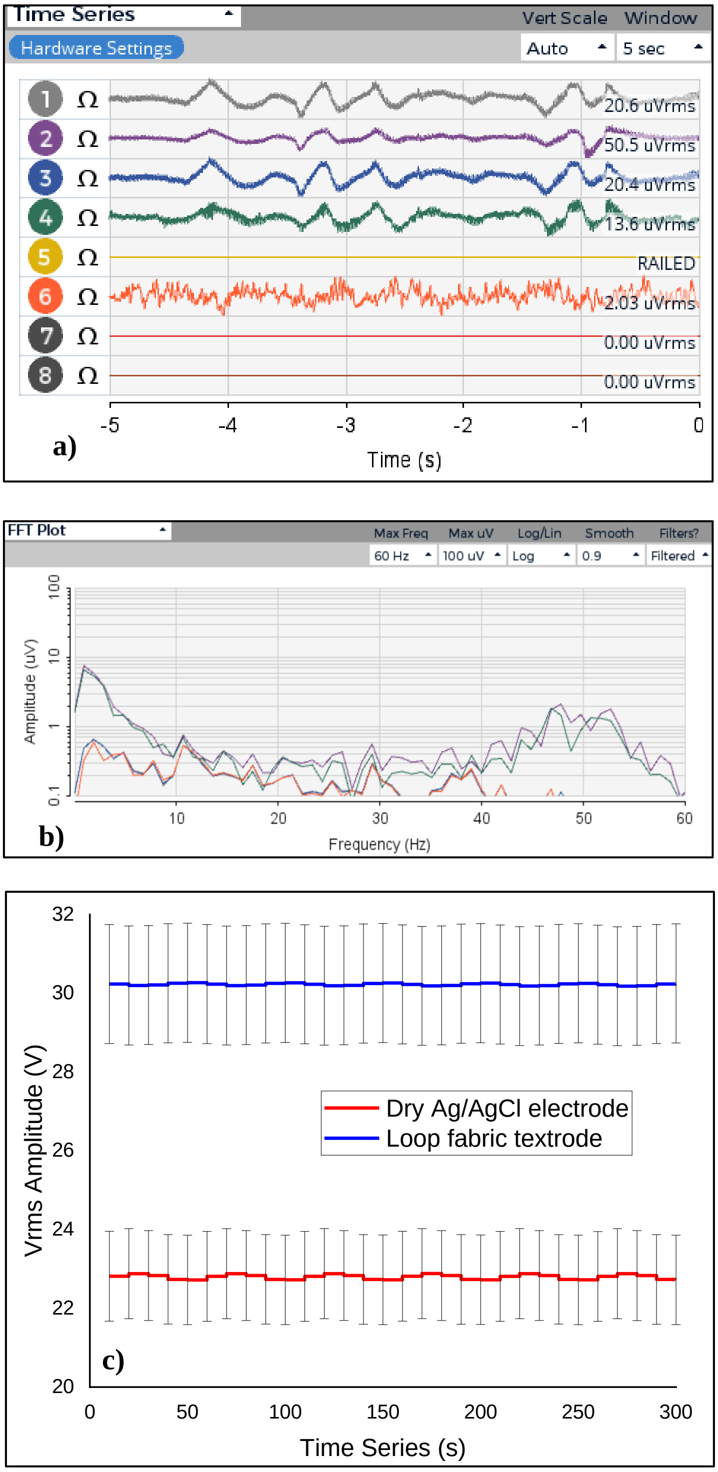

Figure 3. (**a**) Real-time trace of EEG signals; (**b**) FFT plot from the EEG textrode; (**c**) Ten-second-average Vrms amplitudes

3.3. Signal-to-Noise Ratio (SNR)

The textile electrode collects EEG signals with a higher signal-to-noise ratio (+27.4%) than the commercial dry electrodes as shown in Figure 4. This shows that the new textrodes are promising.

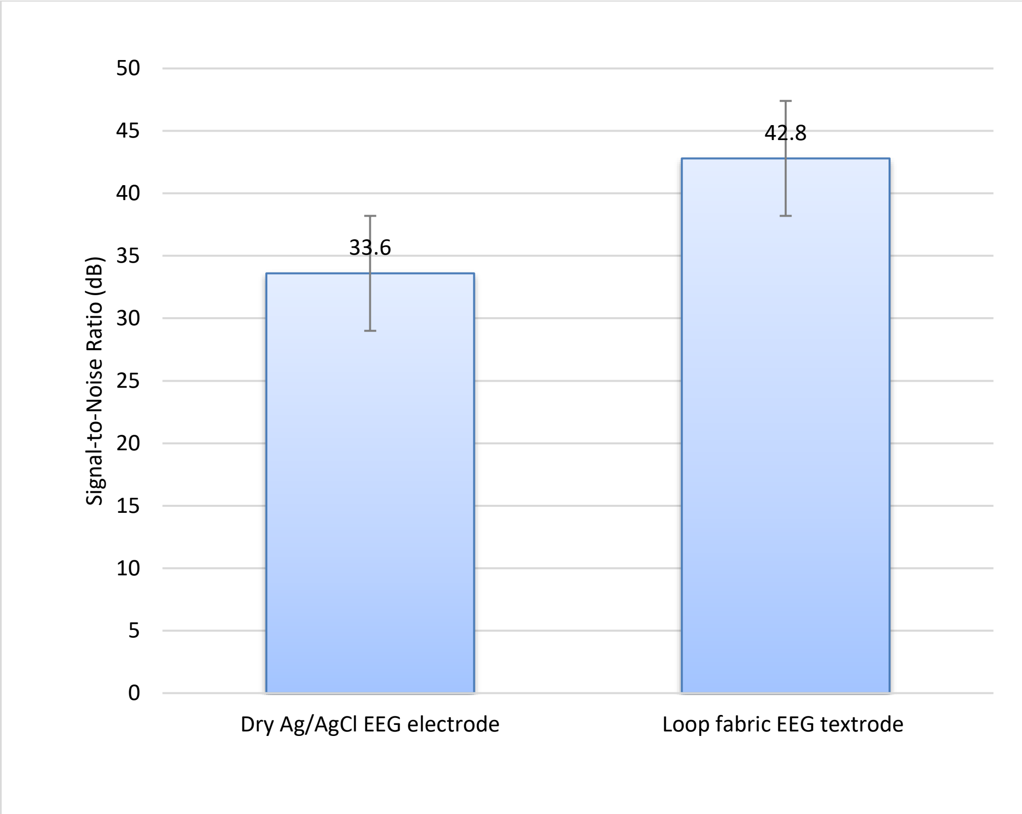

Figure 4. Signal-to-Noise ratio of the EEG electrodes

4. Conclusion

The demand for more comfortable has driven the advancement of different formats of dry electrodes that can overcome the limitations of wet ones. The wet electrodes used for EEG acquisition can cause skin irritation and the gel dries over time which leads to an increase in skin-to-electrode impedance and more signal-to-noise ratio. For that reason, dry metal EEG discs and needle spikes were just introduced. However, their high weight and structural rigidity could make them not suitable for wearable purposes. As a solution to overcome the aforementioned problems, a flexible electro-conductive textile material has been used to develop a washable textile-based electrode that can detect EEG signals comparable to the standard Ag/AgCl electrodes. Moreover, the new textile-based electrode gives much lower skin-to-electrode impedance and a much higher signal-to-noise ratio than commercial dry electrodes. The use of textile-based electrodes could overcome the associated problems and would fill the gap in both wet and dry metal electrodes as they do not need gel to link to the skin.

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