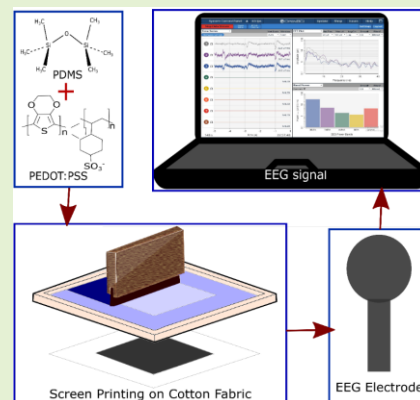


Dry Electroencephalography Textrode for Brain Activity Monitoring

Granch Berhe Tseghai, Benny Malengier, Kinde Anlay Fante, and Lieva Van Langenhove

Abstract—The advancement in smart materials allows researchers to seek smart textiles for wearable health monitoring. Here, a washable and flexible textile-based dry electroencephalography (EEG) electrode that can detect brain activities has been developed. The EEG electrodes were constructed from an electrically conductive cotton fabric with $67.23 \Omega/\text{sq}$ produced through printing PEDOT:PSS/PDMS conductive polymer composite on cotton fabric via screen printing. The mechanical properties like flexural rigidity and tensile strength of the conductive fabric were compared against the bare base material and a PEDOT:PSS-printed fabric. The result from an SEM revealed a uniform printing of the PEDOT:PSS/PDMS on the fabric. The signal-to-noise ratio of the textrode was higher than the Ag/AgCl dry electrode i.e. $17.32 (+3.1\%)$ which open the door for long-term EEG monitoring. Moreover, the electrode can give clear and reliable EEG signals up to 15 washing cycles, 60 bending cycles, 5 multiple uses, and 8 hours of continued use.

Index Terms—brain activity, dry EEG electrode, e-textile, PEDOT:PSS/PDMS, textrode.



I. Introduction

THE use of smart textiles for health monitoring is a booming business. To make it more user-friendly, it is important to develop e-textile-based bio-potential sensors for active control of your health without compromising the comfort and bulk property of the textile. In the case of brain-related health issues, medical treatment might eventually decrease mortality and increase the quality of life, but to do so, both the consequences of the diseases and responses to treatments need to be objectively quantified. Moreover, most brain disorder patients do not recognize the commencement and also do not understand what is happening during the occurrence. Therefore, disorder detection devices that allow a rapid objective assessment and recognition of the illness frequency and treatment through closed-loop systems could potentially lessen morbidity and mortality. With this in mind, the ideal monitoring device should be safe and easy to use for patients, families, and medical staff. It must be comfortable for the users, especially during sleep, and thus preferably

wireless and lightweight. If electrodes are used, they must be as small and few as possible since a significant percentage of patients are not willing to wear electrodes on a long-term basis [1]. On top of that, the device should be unobtrusive and discrete. Uncomfortable cables, electrodes, lights, buttons, and sounds should be avoided as this disturbs the patient and family, even more so in the long term. This was confirmed by a recent survey evaluating patients' desires that revealed a strong preference for a seizure detection device that had little interference with daily activities [2]. The use of textile electrodes could overcome the problems associated with metal-based dry electrodes like structural rigidity and weight and would fill the gap. Textile electrodes are electrode types that are made from conductive textile fabric. These electrodes do not need a gel to achieve connection to the skin. Moreover, they are good for long-time measurements as they do not irritate the skin. In addition, they are lightweight, ductile, and it is possible to make them reusable and washable [3].

Modern applications of brain-computer interface (BCI) based EEG rely heavily on the so-called wet electrodes (e.g. Ag/AgCl electrodes) which require gel application and skin preparation to operate properly. However, these electrodes can cause skin irritation as a conductive gel is necessary to reduce electrode-to-skin impedance mismatch. Besides, it may also lead to artifact generation due to drying out of the gel over time and humidity changes. Furthermore, skin preparation and gel application are time-consuming when a high number of electrodes are required.

The demand for more comfortable and user-friendly electrodes led to the emergence of metal-based dry EEG electrodes that can overcome the aforementioned limitations associated with wet electrodes. However, metal-based dry electrodes have disadvantages of higher electrode-to-skin

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impedance and susceptibility to movement artifacts. In literature, it is reported that because of the accumulation of perspiration sweat under the electrodes, after a settling time, the impedance greatly decreases, and artifact noise becomes lower than for wet electrodes [4]. Other studies reported that dry electrodes have several advantages in comparison with wet ones, such as signal intensity and smaller size [5]. However, metal-based dry electrodes are heavier and have a rigid structure which makes them unsuitable for wearable applications, especially when long-term monitoring is needed. This led to the emergence of textile-based dry EEG electrodes.

Recently, some researches on textile-based dry electrodes have been reported: a passive electrode based on porous titanium (Ti) and PDMS for long term EEG recording [6]; a porous ceramic-based 'semi-dry' electrode that has tips that can slowly and continuously release a tiny amount of electrolyte liquid to the scalp, which provides an ionic conducting path for detecting neural signals [7]; wearable, less visible ear-EEG recording [8]; copper plate fabric textile EEG electrode that can give similar signals as commercially available EEG [9]; knitted soft textile electrodes for EEG Monitoring made from nylon, conductive fibers, Spandex and polypropylene [10]; a textile electrode using electrically conductive polyurethane (PU) foam developed through a coating of inherently conductive polyaniline (PANI) polymer on PU foam [11] and a 3D printed dry electrode made by an insulating acrylic-based photopolymer [12]. All these works reported that the signals acquired were comparable to the standard wet EEG electrodes. However, there is no scientific information or experimental evaluation on the property of the electrodes as a textile material such as the tensile strength, flexural rigidity, coating thickness, and weight [13]. Therefore, it is not known if the textile substrates own characteristics of normal textile materials and the important factors for wearable long-monitoring like flexibility and light weightiness used could not be compromised. The use PEDOT:SS for EEG electrodes has been successfully explored. For instance, Leleux et al. reported a PEDOT:PSS-based electrode cups that give better performance than gold cup electrodes [14]. Ferrari et al. also reported a PEDOT:PSS-based EEG tattoo that collects signals qualities comparable to standard Ag/AgCl electrodes [15]. Here, both mechanical and electrical properties of the electrodes have been assessed as textile and electronic materials. Moreover, the effect of electrode size and shape and its performance under multiple and continuous uses, as well as bending and washing, have been studied.

II. EXPERIMENTAL DESIGN

A. Materials and Chemicals Selection

A 140 gram per square meter (GSM) knitted cotton fabric obtained from (UGent MaTCh laboratory, Belgium) was chosen as a textile substrate. A high-conductivity grade PEDOT:PSS PH1000 Clevious conductive polymer obtained from (Ossila Ltd., UK) and a biocompatible Poly(Dimethylsiloxane (PDMS) elastomer obtained from (Polyscience, Inc., UK) were used to produce a conductive polymer composite. 1,2,3,4-Butanetetracarboxylic acid

(BTCA) obtained from (Sigma-Aldrich, Inc, Germany) was also used as a fixing agent to improve wash fastness.

B. Conductive Fabric Development

A 1:4 PDMS to PEDOT:PSS was mixed at room temperature using a 2 mm diameter circular rod until a homogenous PEDOT:PSS-PDMS paste was obtained. Then, 10% of the weight of paste BTCA was added. The PEDOT:PSS/PDMS paste was next screen printed on a knitted cotton fabric [16]. Next, the printed-fabric was exposed to 70 °C for 10 minutes to dry and 150 °C for 3 minutes to cure. The schematic illustration of the screen printing and the actual conductive fabric are shown in Fig. 1(a) and (b), respectively.

C. Conductive Fabric Characterization

The increase in weight that could be also considered as the increase in GSM, was mathematically determined as a difference of the conditioned weight before and after printing. The thickness was determined according to ISO 5084:1996(E) and printing thickness was determined from the difference in thickness before and after printing. Bending length was measured according to BS 3356:1990 and its respective flexural rigidity was calculated using (1).

$$G = MC^3 \quad (1)$$

where, G = Flexural Rigidity (mg cm), C = Bending Length (cm), M= Mass/Area of Specimen (g/m²).

The strength and elongation at break were tested according to ISO 13934-1. The surface topology and appearance of yarns within the fabric before and after printing were also tested using FEI Quanta 200 FFE-SEM at an accelerating voltage of 20 kV. The non-conductive sample was printed with gold using Balzers Union SKD 030 sputter printer prior to analysis. FTIR test was also conducted to verify the presence of PEDOT:PSS/PDMS conductive polymer composite on the knitted cotton fabric after printing.

Next, the electrical characteristics, specifically the sheet resistance and resistivity of the conductive fabric used to construct EEG electrodes was measured using a portable four-probe MR-1 Surface Measuring Instrument. The measurement was carried out at five different positions of a 5 cm x 5 cm conductive fabric and we recorded the average. The electrical analysis was also done after washing the fabric, washing was carried out according to ISO6330:2012(E) Type A (mild).

In addition, the textile-based EEG electrodes were subjected to 25 washing cycles according to ISO6330:2012(E) Type A (mild) test method and were bend to a 5 mm radius. The impact of this on the EEG signal quality has been studied. Also, the effect on the signal quality of multiple and continued use of the electrodes has been studied.

D. Textile-Based Electrode (Textrode) Design

Circular electrodes with 1 cm, 2 cm, and 4 cm diameters were constructed from the printed fabric (Fig. 1(b) to examine the effect of electrode size on signal quality. In addition, the effect of electrode shape on signal quality was studied. For this purpose, circular, rectangular square, and equilateral triangular electrodes of each π cm² surface area were constructed and compared with each other.

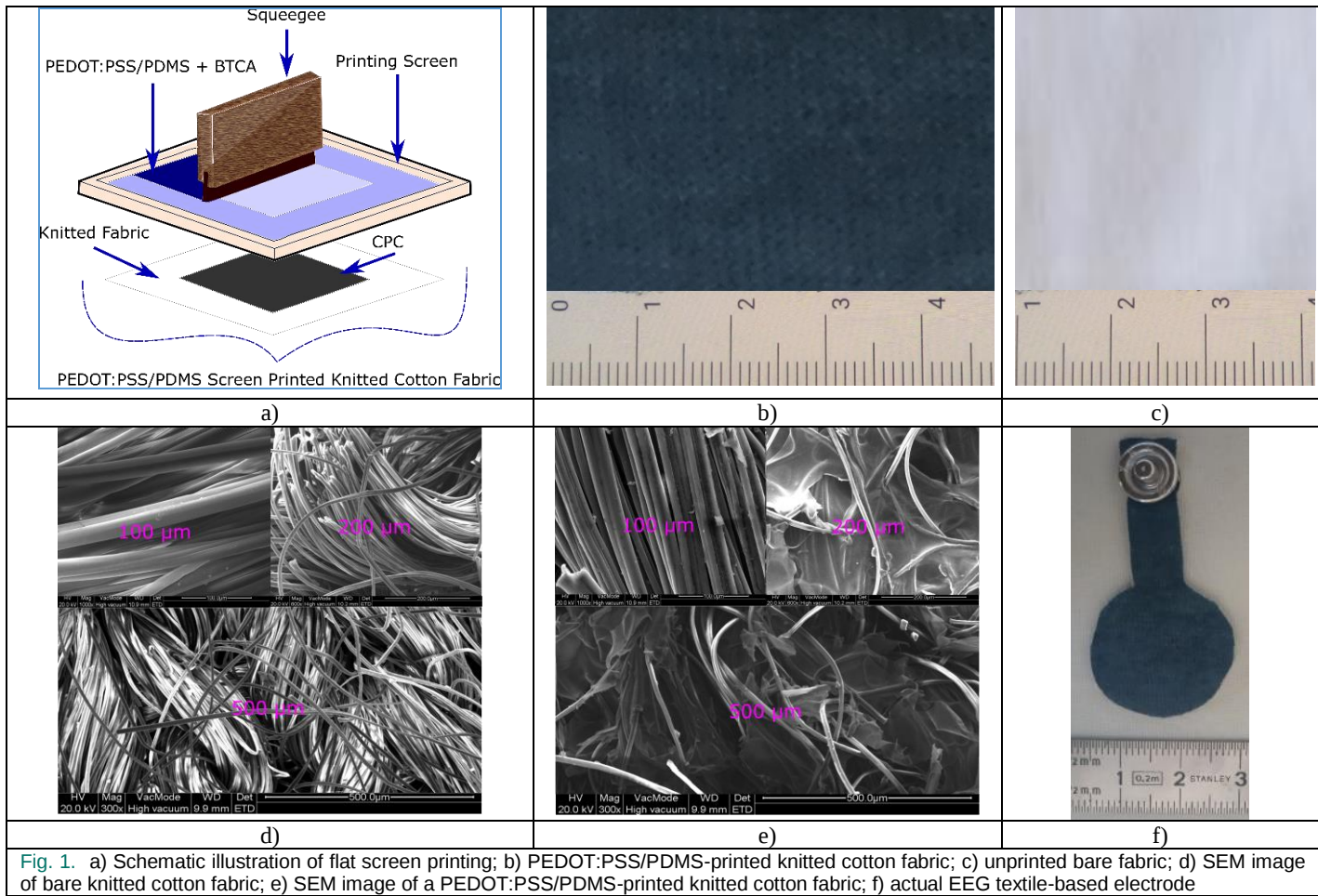


Fig. 1. a) Schematic illustration of flat screen printing; b) PEDOT:PSS/PDMS-printed knitted cotton fabric; c) unprinted bare fabric; d) SEM image of bare knitted cotton fabric; e) SEM image of a PEDOT:PSS/PDMS-printed knitted cotton fabric; f) actual EEG textile-based electrode

E. Impedance Measurement

The skin-to-electrode impedance was measured using the OpenBCI board according to the illustrated setup shown in Fig. 2(a). This measurement setup gives a root mean square (RMS) voltage. Therefore, the actual average impedance of each electrode was calculated using (2). In the equation, the induced current i.e. 6 nA, and a load resistance i.e. 5 kΩ are considered.

F. EEG Measurement

Five electrodes, three active and two reference electrodes, were placed on the frontal head according to the 10-20 EEG placement system [17, p. 1]. The 10 and 20 in the system indicate the distance of the electrodes from each other in proportion to the size of the head. The numbers in the indicated positions represent the corresponding position of the forehead. Even numbers represent the right position and odd numbers the left. The diagram in Fig. 3(a) illustrates the position of the electrodes.

The EEG waveforms were recorded using an OpenBCI board and displayed on a laptop with the OpenBCI GUI software. A tight-fitting headband constructed from an elastic bandage was used to hold the electrode in the required

positions according to the 10-20 system, preventing them from sliding. A non-conductive foam was placed between the elastic bandage and the electrode to ensure the electrode is sufficiently pressed against the skin. The schematic illustration of the EEG measurement setup used in this work is shown in Fig. 3(a). The collected signals were also analyzed using EEGLab software. A Florida Research Instrument Inc product reusable Ag/AgCl standard dry Electrode (TDE-20-15 reusable EEG electrode) obtained from OpenBCI was used as a comparison electrode.

G. Signal To Noise Ratio (SNR) Analysis

A synthetic sine wave (360 mv peak to peak voltage, 9.925 Hz frequency, 50 ms time) was generated using Micsig TO1104 Handheld Tablet Digital Oscilloscope and then injected into a ballistic gelatin phantoms. To impersonator events, the EEG phantom signal parameters were set in the alpha wave range and varying the amplitude to mimic a neurological event. Then EEG wave was measured the textrode and Ag/AgCl dry electrode with an OpenBCI board. Then, the quality of signals collected was analyzed mathematically in terms of Signal-to-Noise Ratio (SNR) using (3).

$$\text{Actual Average Impedance } (\Omega) = (\text{RMS Voltage} \times 2\sqrt{2} \text{ (V)}) / (\pi \times \text{Current (A)}) - \text{Load Resistance } (\Omega) \quad (2)$$

$$\text{SNR (dB)} = 10 \log(\text{Peak to Peak Voltage Signal}) / (\text{Peak to Peak Voltage Noise}) \quad (3)$$

III. RESULT AND DISCUSSION

A. Mechanical Property Analysis

The gram per square meter (GSM) of the fabric increased only by 100 GSM (71.4%). The increase in weight can also be considered as the solid add-on of the conductive polymer composite. The thickness measurement based on ISO 5084:1996 (E) showed an increase from 0.5 mm to 0.65 mm (+30%). The significant change in weight and thickness might lead us to believe that the textile aspect of the fabric was greatly changed. This was not the case, however, as the bending length test according to the BS 3356:1990 test method revealed a very small increase from 2.45 cm to 2.61 cm (+7%) through its respective flexural rigidity (1) increased from 205.88 to 426.71 mg cm (+107%), which is a consequence of the added weight. This new value indicates the knitted fabric now behaves as a twill ¼ cotton fabric or a twill 2/1 kermel/viscose of the same approximate GSM [18]. The effect of the presence of PDMS on flexural rigidity was studied in previous work [19]. In that study, the flexural rigidity of a PEDOT:PSS/PDMS conductive polymer composite printed fabric has been achieved lower than the same fabric printed with pristine PEDOT:PSS. Therefore, the flexibility of the conductive fabric used for the construction of the EEG electrodes is also better than a pristine PEDOT: PSS printed fabric. Most importantly, the current conductive polymer composite printing caused an improvement in tensile strength i.e. from 68.1 to 116.4 N, and a drop in the tensile strain at break i.e. from 113 to 86.5%.

The surface morphology analysis using SEM showed an evenly smooth fabric surface with fewer protruding yarn loops after printing and the yarn loop interstices were filled up by the conductive polymer composite. SEM images of the fabric before and after printing are shown in Fig. 1 (d) and (e), respectively. Therefore, from the mechanical analysis, we can say that the conductive cotton fabric retains its textile texture.

As shown in Fig. S1 (Supporting information), the FTIR spectra of the bare cotton fabric (Fig. 1(c)) and PEDOT:PSS/PDMS-printed cotton fabric (Fig. 1(b)) showed cellulose characteristic FTIR peaks, i.e. O–H (3331.86 and 3331.85 cm^{-1}) [20], C–H (2889.82 and 2867.77 cm^{-1}) [21] and C–O (1052.58 and 1049.59 cm^{-1}) [22] stretching vibrations, respectively, as does the knitted cotton substrate used. In the presence of BTCA, the O–H stretching is disappeared and the C–H and C–O stretching vibrations are shifted to 2879.28 and 1054.27 cm^{-1} , respectively, which could be due to the esterification of cellulose with BTCA. The peaks in the spectrum at 830.98 and 882.89 cm^{-1} attributed to the C–S stretching vibrations [23] of the thiophene ring of the PEDOT. The peaks at 1313.44 and 1401.71 cm^{-1} attributed to the S=O symmetric stretching [24] of the sulfonate of the PSS. The peak at 1162.98 cm^{-1} attributed to the Si–O–Si stretching [25] confirms the presence of the PDMS. At 1707.04 cm^{-1} attributed to C=O [26] appeared which confirms the presence of conjugated ester bonds BTCA. This bond did not appear in raw cotton, while for the PEDOT:PSS sample only a small 1723.58 cm^{-1} peak attributed to C=O [27] is present. Finally, the FT-IR test confirms the presence of PEDOT:PSS/PDMS and BTCA on cotton fabric. Finally, the FT-IR test confirms the presence of PEDOT:PSS/PDMS and BTCA on cotton.

B. Electrical Property Analysis

The surface resistance and resistivity of the PEDOT:PSS/PDMS-printed knitted cotton fabrics measured at five different positions of the fabric (top-left, bottom-left, top-right, bottom-right and center of a 25 cm^2 printed fabric) via a four-probe MR-1 Surface Measurement Instrument was 67.08 , 67.19 , 67.09 , 67.78 and 67.03 Ω/sq and 1.34 , 1.33 , 1.33 , 1.34 , and 1.32 Ω cm, respectively. Therefore, the average sheet resistance and resistivity with their standard deviation and margin of error are 67.23 ± 0.27 ($\pm 0.41\%$) Ω/sq and 1.33 ± 0.007 ($\pm 0.55\%$) Ω cm, respectively, at a 95% confidence interval (CI). This indicates that printing uniformity and distribution of the conductive component over the fabric surface were good.

Moreover, the conductive fabric stayed conductive up to 20 washing cycles. The sheet resistance of 20 times washed conductive fabric at five different positions was 68.13 , 67.81 , 67.25 , 67.47 and 67.96 Ω/sq and the average with its standard deviation and margin of error is thus 67.72 ± 0.32 ($\pm 0.47\%$) Ω/sq . Though a very small increase in sheet resistance was observed ($+0.73\%$), the difference was found to be not significant providing 5.32 f-ratio and 0.05 p-values at a 95% CI based on One-way ANOVA. This good washing fastness is due to the use of BTCA as a fixing agent.

C. Skin-to-Electrode Contact Impedance

At the beginning of the measurement, the skin-to-electrode contact impedance of the textrode was higher than the Ag/AgCl dry electrode. But, it becomes lower after 3 minutes as shown in Fig. 2(b), the average numerical results of a five replica are provided in Table S1 (Supporting Information). These results could be because perspiration can wick to the textrode but not to the Ag/AgCl dry electrode i.e. metal. The one-way ANOVA at a 95% CI showed the skin-to-electrode contact impedance of the textrode and Ag/AgCl dry electrode is significantly different after three minutes providing an f-ratio value of 20.97 and p-value of < 0.05 . Moreover, the skin-to-electrode contact impedance of the textrode also significantly dropped at a 95% CI after three minutes. The One-way ANOVA test before and after three minutes resulted in an f-ratio value of 165.59 and a p-value of < 0.05 . Most importantly, the skin-to-electrode contact impedance stayed strongly uniform in the textrode as shown in Fig. 2(b). Thus, the textrode could have a potential advantage over the Ag/AgCl dry electrode for long-term monitoring, especially for wearable applications. Moreover, the impedance is less than 5 $\text{k}\Omega$ [28], so it fulfills the requirement to be used as an EEG electrode. Besides, the coefficient of determination (R^2) indicated that more than 95% of the response data are around the mean.

D. EEG Signal Analysis against Standard Electrodes

The EEG signals recorded by the textrode were comparable to Ag/AgCl dry electrode. The amplitude of the signal was as high as the Ag/AgCl dry electrode. Moreover, the textrode gave a similar range of bandwidth. The EEG signals from both the textrode and Ag/AgCl dry electrode after 3 minutes of measurement are shown in Fig. 3(b) and (c), respectively.

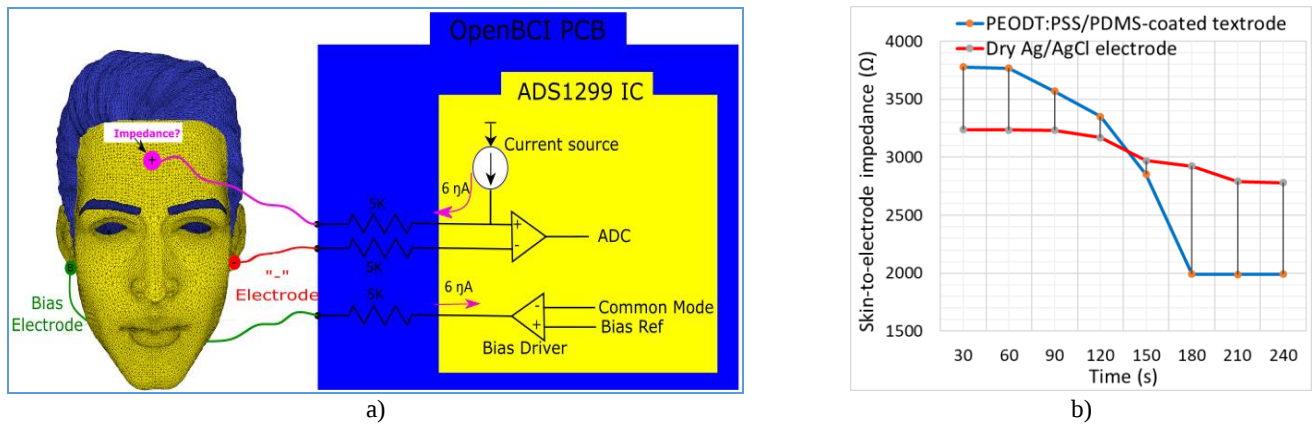
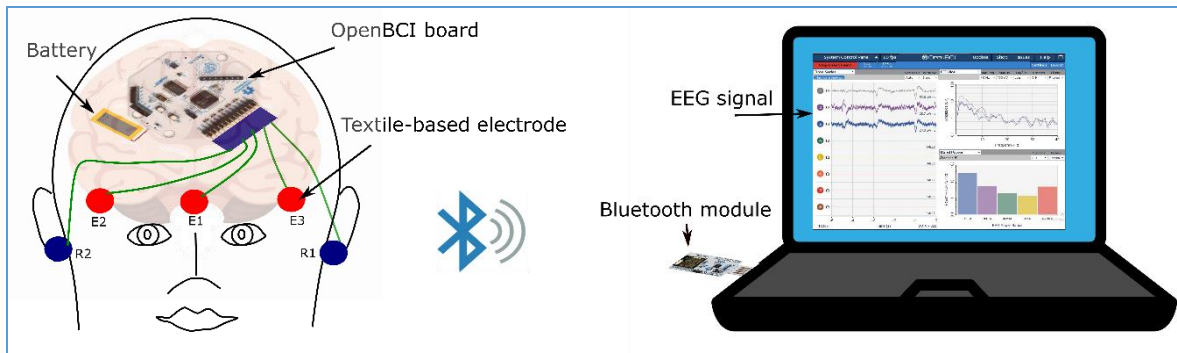
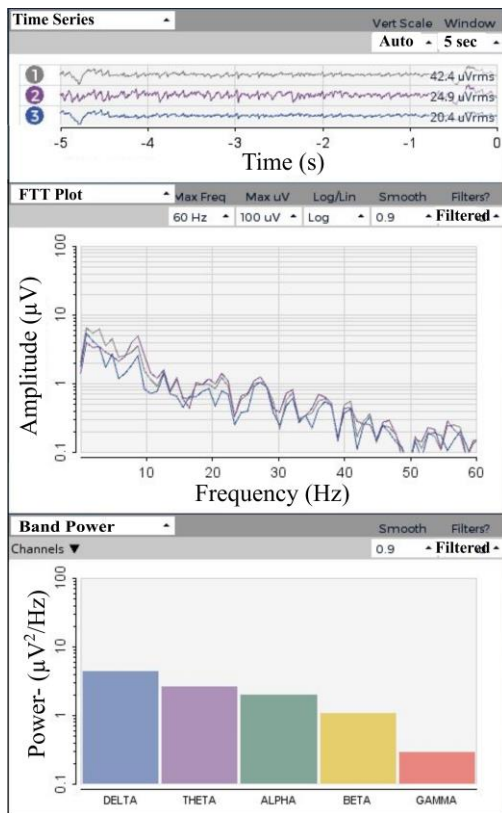


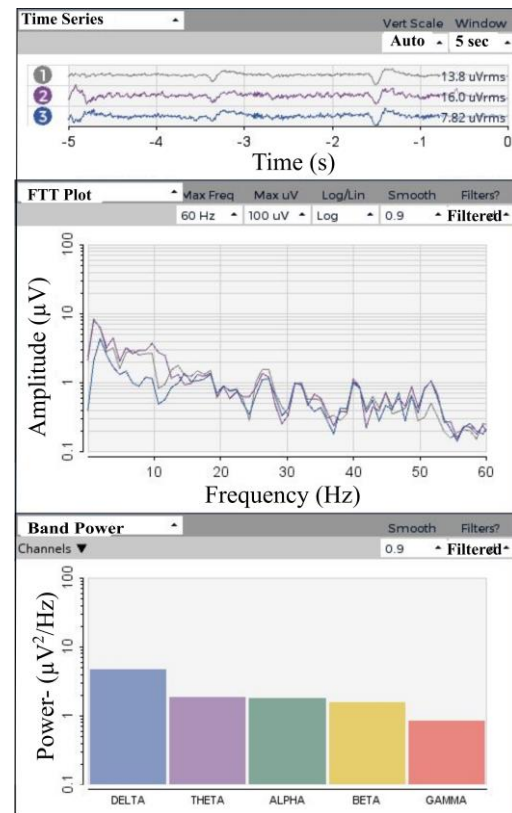
Fig. 2. a) Schematic illustration of skin-to-electrode impedance measurement using OpenBCI board; b) skin-to-electrode impedance condition of the EEG electrodes over time, PEDOT:PSS/PDMS-printed textrode and dry Ag/AgCl dry electrode



a)



b)



c)

Fig. 3. a) Schematic illustration of the EEG measurement setup; b) EEG signal from silver-silver chloride commercial dry electrode tested at 60 Hz notch and 1-50 Hz bandpass filter; c) EEG signal from textile-based electrode tested at 60 Hz notch and 1-50 Hz bandpass

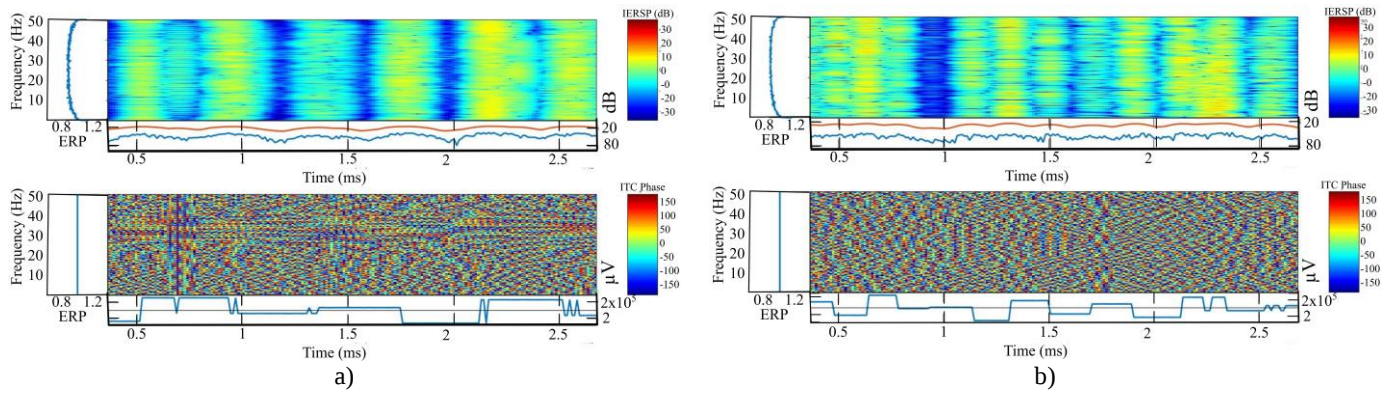


Fig. 4. ITC phases: a) Ag/AgCl dry electrode CDEs; b) textile-based electrode

The output graph of inter-trial coherence (ITC) from the textrode and the Ag/AgCl dry electrodes were compared using EEGLAB software. ITC shows the degree of a tendency for the data phase at each frequency to be in the same phase in each trial. It is a measure of phase consistency over trials, where a value close to 0 shows high variability of phase angles across trials, and a value close to 1 shows a low variability. The ITC plot can be generated from an EEGLAB software that in turn gets generated as in (4) according to spectral and coherence estimates on EEG recordings [29].

$$ITC(f, t) = \frac{1}{n} \sum_{k=1}^n \frac{F_k(f, t)}{|F_k(f, t)|} \quad (4)$$

where, F and t denote frequency and time, respectively. In each plot, the frequency range and the time range are placed in the y-axis and x-axis, respectively, and a color scale is used where green is for non-significant, and red represents significant ITC at a 99% CI. Beneath each ITC plot is the averaged ERP response for that individual (in blue), in microvolts.

For this work, the amplitude scale for the ERP response is fairly similar for both the textrode and Ag/AgCl dry electrodes. To the left of each ITC plot, the average power is shown for that electrode at each frequency, while a black dotted line shows the significance threshold at each frequency relative to the baseline period at a 99% confidence level. The ITC phases of the textrode and Ag/AgCl dry electrode are shown in Fig. 4. From EEGLAB software analysis, the maximum log power spectral density for both the textrode and Ag/AgCl dry electrode was ~ 95 dB. The ERSP graph is also included in Fig. S2 (Supporting Information)

E. Signal To Noise Ratio (SNR)

The average SNR of the textrode has been found at 17.378 dB ± 0.0716 ($\pm 0.41\%$) at a 95% confidence interval which shows the values are not significantly different. Therefore, the sensing reputability of the textile-based EEG electrode is excellent. Moreover, The SNR of the textrode was higher than the Ag/AgCl dry electrode by 3.1%.

F. Effect of Bending on EEG Signal Quality

The electrodes showed good EEG signals up to 60 cycles of bending to a 5mm radius. Based on the one-way ANOVA Tukey HSD Test at a 95% CI, the difference of the means of the maximum RMS voltage amplitude up to the 60 bending cycles was insignificant giving Tukey HSD p-values of 2.34. This robustness to bending could be due to the presence of

PDMS elastomer that contributes much to the flexibility of the conductive fabric and so to the electrode. The EEG signals for three active electrodes at 60 Hz notch and 1-50 Hz bandpass filter at different bending cycles are shown in Fig. 5(b). The numbers (1, 2 and 3) in the vertical axis of figures in Fig. 5 represent the active electrodes E1, E2, and E3, respectively.

G. Effect of Multiple and Continuous Use on EEG Signal Quality

The washed textrode has been used five times and collected clear signals and similar amplitudes as the unwashed electrode used for the first time. Moreover, the electrodes were also tried for 8 hours of continuous use and no variation in the quality of signal acquisition was observed. In this test, the result was not collected for 8 hours but electrodes were simply placed for 8 hours, and then the actual measurement was done after that. The one-way ANOVA Tukey HSD Test at a 95% CI on the multiple and continuous test proved that there were no significant differences in the mean of the RMS voltage. The HSD Tukey p-values were 0.80 and 0.43 for the multiple and continuous use test, respectively. Their respective EEG signals at 60Hz notch and 1-50 Hz bandpass filter are shown in Fig. 5(c) and (d).

H. Effect of Washing on EEG Signal Quality

The textrode constructed from 5 cycles washed conductive fabric gave almost similar signals with unwashed ones. Based on a one-way ANOVA Tukey HSD Test at a 95% CI, the Tukey HSD p-value was 0.09 for 15 washing cycles. However, after 20 washing cycles, the amplitude becomes very small. The EEG signal at 60 Hz notch and 1-50 Hz bandpass filter before and after washing are shown in Fig. 5 (e). The improvement in the washing fastness is because of BTCA which was intentionally incorporated during the printing for this purpose. The PEDOT:PSS/PDMS could become fixed to cotton using the polycarboxylic acid crosslinker BTCA, a group of widely used durable press crosslinkers in the textile industry in the presence of sodium hypophosphite (SHPI) catalyst. SHPI is a widely used catalyst with BTCA. According to a US patent, the catalyst can be effective in the range of 0.3% (3 g/l) to 11% of the crosslinker [30]. For this work the median ratio i.e. 5.5% has been used.

BTCA possesses four carboxylic acid groups and one to three of the carboxylic acid groups are involved in an esterification reaction with cotton by the formation of a five-membered cyclic anhydride through the dehydration of the

adjacent carboxyl groups followed by the formation of ester links between the anhydride intermediate and the cellulose of the cotton [31].

I. Effect of Electrode Size on EEG Signal Quality

The Ag/AgCl dry electrode is 1 cm in diameter. However, at this size, the signal qualities of the textrode are not satisfactory, resulting for example during blinking in artifacts such as very wavy signals. Hence, the size was doubled. The effect of electrode size on EEG signal acquisition is shown in Fig. 5(g, h & i). Based on the one-way ANOVA Tukey HSD test at a 95% confidence level, reducing the 2cm electrode by half or doubling the size caused a significant difference in the RMS voltage amplitude of the EEG signals, providing a Tukey HSD p-value of less than 0.01. Moreover, there is also a bandwidth shift when the size is decreased by half as shown

in Fig. S4 (Supporting Information). Whereas doubling the size of the electrode did not change the bandwidth and collected clear signals.

J. Effect of Shape on EEG Signal Quality

With the shape, the signal did not show any influence on the quality when observed visually. However, the RMS voltage amplitude was found to be affected, within decreasing order the circular electrode, rectangular square, and lowest value for the equilateral triangle, with Tukey HSD p-values of less than 0.01, based on one-way ANOVA Tukey HSD Test at a 95% CI. besides, rectangular and triangular electrodes need bigger space to place on the skin as they have pointed tips. Thus, the circular electrode is preferable. The effect of electrode shape on EEG signal acquisition is shown in Fig. 5 (j, k & l).

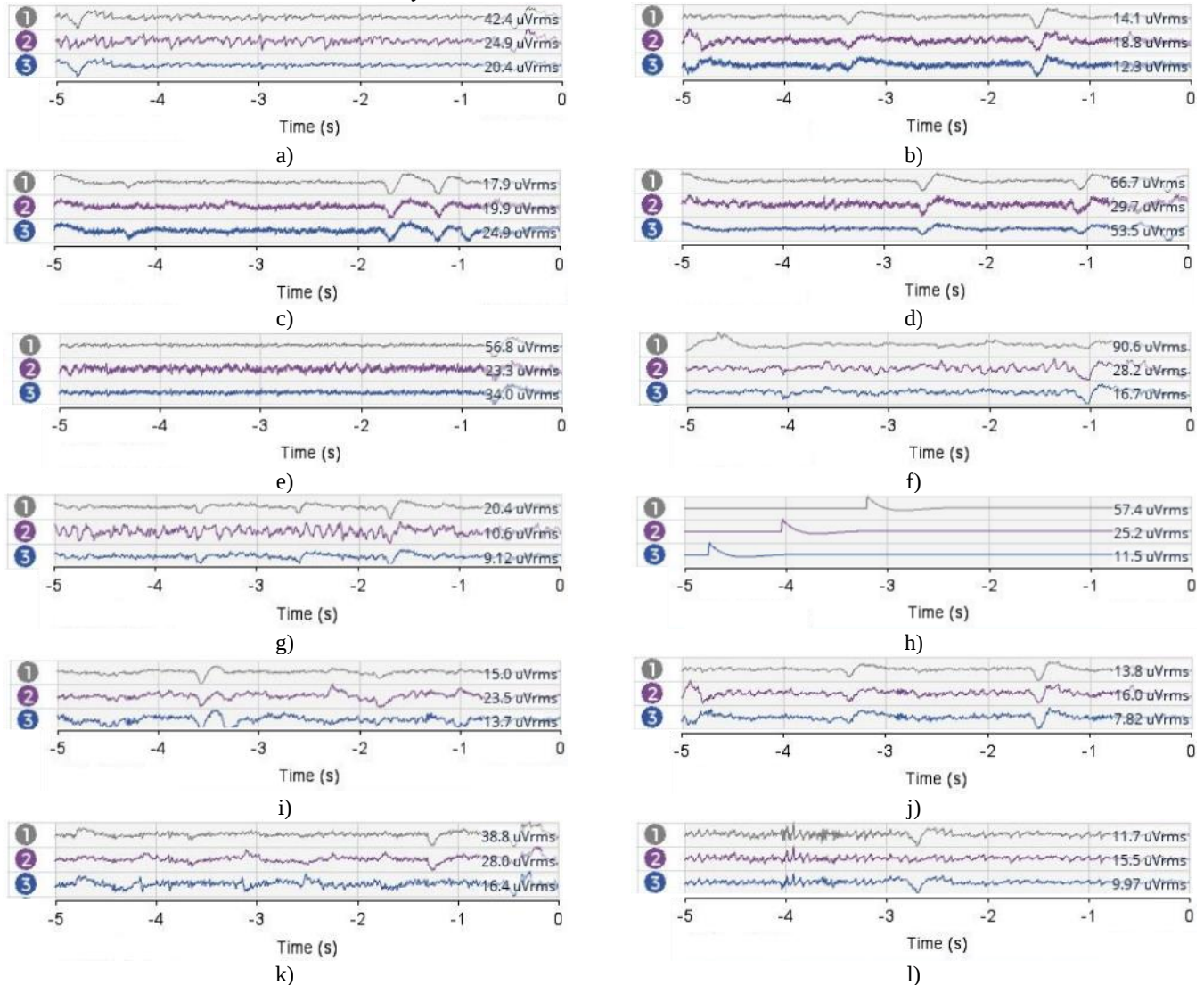


Fig. 5. Stability of signal quality to different factors tested at 60 Hz notch and 1-50 Hz bandpass filter: a) Ag/AgCl dry electrode; b) unwashed textile electrode (TE); c) unwashed TE after 60 bending cycles; d) unwashed TE at 5 multiple uses; e) unwashed TE after 8 hours continuous use; f) TE after 15 washing cycles; g) 4 cm diameter circular electrode; h) 2 cm diameter circular electrode; i) 1 cm diameter circular electrode; j) π cm² Circular electrode k) π cm² rectangular square electrode; l) π cm² equilateral triangle electrode. The fast Fourier transform plots and band powers of Fig. (a) –(l) are presented in Fig. S3, S4 and S5, (Supporting Information).

IV. CONCLUSION

The demand for more comfortable and user-friendly electrodes has led to the development of an increasing number of dry EEG electrodes that can overcome the limitations of wet electrodes, however, the commercial electrodes are metal-based and still lack in flexibility, while they are also relatively heavy, which makes them unsuitable for wearable applications. Therefore, the use of textile electrodes would overcome these associated problems. Therefore, this work explored textrodes suitable for long-term brain activity monitoring, especially for wearable applications.

In this work, we have successfully developed a machine-washable, flexible and lightweight textile-based dry electrode from PEDOT:PSS/PDMS-printed cotton fabric that collects good quality EEG signals comparable to the commercially available metal-based dry electrodes. Moreover, the impedance of the textrode is lower than that of the Ag/AgCl dry electrode after 3 minutes and gives a higher signal-to-noise ratio. This textrode showed good stability to washing, bending, and multiple and continuous uses. Most importantly, the electrode does not need a gel to achieve connection to the skin.

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