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Comb-shaped Polymer-based Dry Electrodes for EEG/ECG Measurements with High User Comfort

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***Abstract*— Soft, comfortable polymer-based dry electrodes are fabricated. Impedance and biopotential measurements are carried out to compare the performance of conventional gel electrodes with our dry electrodes. The impedance of our dry electrodes is reduced by adding more conductive additives to the polymer material. To further lower the impedance, two skin pretreatment techniques are evaluated regarding their influence on skin impedance. However, these techniques are found to have only temporary beneficial effects. Finally biopotential measurements (both ECG and EEG) are performed using our soft polymer electrodes. The ECG signal acquired with both gel and our polymer electrodes demonstrates high degree of similarity. Therefore, heart beat detection is straightforward. To enable monitoring of EEG signals with smaller amplitudes, our dry electrodes need to be combined with pre-amplifiers. Initial EEG tests show that the alpha waves are clearly identifiable with the dry electrodes when subjects close their eyes. Based on the results, combining with sophisticated signal acquisition electronics, the dry electrodes provide a high user comfort solution for high quality biopotential measurements, even on very hairy skin.**

I. Introduction

For current ECG (Electrocardiography) and EEG (Electroencephalography) biopotential recordings, gel electrodes are widely used. Nevertheless, many drawbacks of applying the gel electrodes exist, such as skin irritation, gel drying during measurement, cumbersome skin pretreatment and electrodes set-up by experts [1]. Skin irritation often happens because of the long-term electrode/gel-skin contact. Besides, the signal quality will decline during monitoring because of gel drying. For conventional EEG analysis with gel electrodes, the electrode set up is time consuming and often painful, and needs to be performed by experienced people. Furthermore, removing the gel after EEG monitoring is again time consuming.

Dry electrodes are good substitutes to avoid the disadvantages of gel electrodes. Some dry electrode EEG systems are commercially available. However, they typically use hard metal electrodes. Subjects often suffer from pain after monitoring their EEG for a while because of high pressure from the hard electrode material and poor designed shape of the electrodes as well as of the whole headset [1, 2]. Important improvements can be realized by dry electrodes fabricated using more flexible materials and/or by spring-like electrode mounting systems on the headset.

In this paper, two kinds of comb-shaped polymer-based dry electrodes are presented. The conductivity of the electrodes is investigated by measuring the impedance on phantoms (platinum metal film and electrolyte wet cloth) and human skin. Furthermore, the influences of skin hydration and the use of abrasive gel on skin impedance are studied. Lastly, our polymer dry electrodes are used to monitor ECG and EEG signals using imec's biopotential systems [3, 4].

II. Electrode Fabrication

A. *Non-conductive polymer with metal coating*

First, the electrode shapes were designed in Solid Edge. Then they were fabricated by stereo lithography using Verowhite as material. Third, a four layer metal stack (TiW /copper/TiW/gold) was coated onto the Verowhite material to make the electrodes conductive. Finally, the electrodes were mounted into the EEG headsets for EEG monitoring, see Fig. 1. This kind of electrode was used mainly to optimize the electrode design for minimum impedance on hairy skin: influences of pin number, pin shape and pin gap size have been studied.

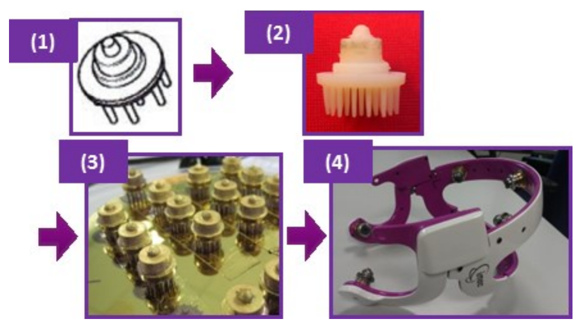

Figure 1. Non-conductive polymer with metal coating for conductivity. (1) designing by software (2) prototyping by stereolighography (3) coating with conductive layers (4) mounting into EEG headset.

B. *Conductive polymers*

EPDM (ethylene propylene diene monomer) rubber electrodes with conductive additives, such as carbon black, were fabricated by compression molding, see Fig. 2. These electrodes are designed with 2mm high flexible pins, and fabricated using a variety of polymer compositions, in order to study the influence on impedance.

Figure 2. Conductive polymer electrodes using EPDM rubber as matrix combined with conductive additives (such as carbon black).

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III. Impedance Measurement

A. Method

In order to study the impedance of the dry electrodes, the following measurement method is followed. A potentiostat (IVIUM), is used to establish a three electrodes system, which contains a reference, a counter and a working electrode, see Fig. 3 and [5, 6]. 25mV is generated from the potentiostat. The current which passes through only working (W) and counter (C) electrodes is calculated by the potentiostat internally. The reference electrode is used to acquire the voltage across the working electrode. Since the voltage and current on working electrode are known, the impedance between the electrode and skin can be calculated.

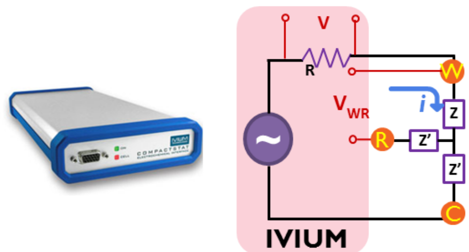

Figure 3. (1) IVIUM generated 25mV and sweeped from 1 Hz to 100kHz during the impedance measurement. (2) Equivalent electrical circuit applied for electrodes impedance measurement. W, R ,C labels stand for working, reference and counter electrodes.

It is known that skin properties change continuously. In order to study accurately the influence of the polymer composition on the impedance of dry electrodes, impedance measurements are also carried out on two phantoms, next to the skin measurements. The first phantom is a platinum coated Si wafer, and the second one is a cloth soaked in saline solution (Bausch & Lomb ReNu MultiPlus).

For the impedance evaluation, two commercially available gel electrodes are placed with 10cm distance from each other on the phantoms or the forearm. These electrodes are the counter and reference electrodes. The electrode under test (working electrode), is placed next to the reference electrode, as illustrated in Fig. 4. The measurements always started 30 minutes after the set up to accommodate for the so-called 'stabilization' of gel electrodes on skin [7]. After connecting the electrodes with the potentiostat, the measurement will start, using a signal amplitude of 25mV and a frequency changing in discrete steps from 100 kHz down to 1Hz, at equidistant intervals on a logarithmic scale. In this work, the impedance data is only reported at 10Hz to simplify the comparison procedure. Similar observations can also be made at other frequencies.

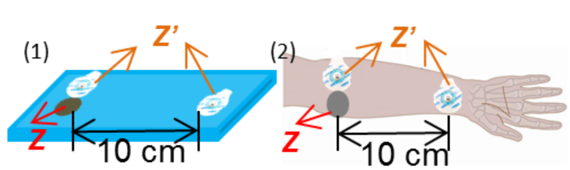

Figure 4. Electrode positions for impedance measurement on (1) phantoms (Pt metal film and electrolyte wetted cloth) and (2) human forearm. Gel electrodes (with impedance Z') were used as reference and counter electrodes. The electrodes under test (with impedance Z) were placed 10 cm away from the reference electrodes.

B. Influence of content of additives in conductive polymer electrodes

The impedance of dry polymer electrodes containing different amounts of carbon black was measured on phantoms and on 4 subjects. Fig. 5 shows that the impedance decreased with higher carbon black content. The impedance approached the gel electrode when carbon black content reaches ~49%. A similar trend is observed when the electrodes were placed on the chest on suitable positions for ECG signal recording.

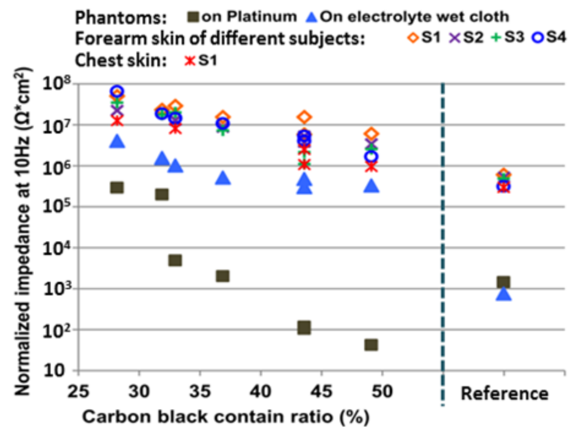

Figure 5. Dry conductive electrode impedance versus the content of conductive additives (carbon black) in the polymer, on phantoms, forearm skin and chest skin of different subjects. The gel electrode impedance is plotted at the right hand side as a reference.

C. Influence of skin pretreatment techniques

The impedance was not only related to the electrodes but also the skin condition. It is know that the most outer part of human skin, the stratum corneum, has a high impedance [8]. Consequently, abrasive gel (to remove some part of the stratum corneum) and conductive gel (for hydration) are used to lower the impedance in clinical settings [9]. In this work, gel of conventional ECG electrodes and abrasive gel (Nuprep Skin Prep Gel) were applied to check if the impedance of the conductive polymer dry electrodes also decreased by the help of these gels.

After putting a gel ECG electrode on skin for 30 minutes and then removing it, the dry conductive polymer electrodes were placed on the gel exposed area. As shown in Fig. 6, the impedance of dry conductive polymer electrodes decreased dramatically after the skin treatment. The measurement was continued until 40 minutes after applying the treatment.

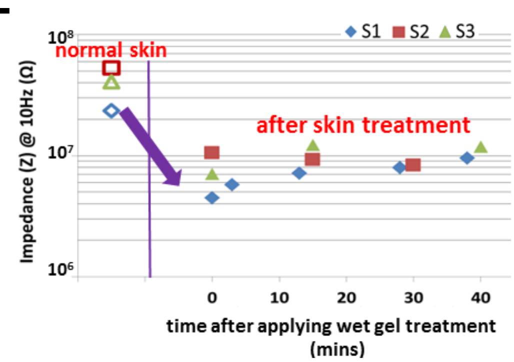

Figure 6. Dry conductive polymer electrodes impedance over time after applying skin pretreatment on different subjects.

The effect of skin hydration (and hence a lower impedance) is only temporally. For 2 subjects of our test the skin impedance was increasing already within the duration of the test (40 minutes).

Another skin pretreatment technique was applying abrasive gel on the skin for 3 minutes and then cleaning with water and drying with tissue paper. As shown in Fig. 7, the impedance of conductive polymer electrodes on one subject decreased immediately after the pretreatment. However, when observing the impedance for longer time (90 minutes), the impedance increased again, approaching the impedance value on normal skin.

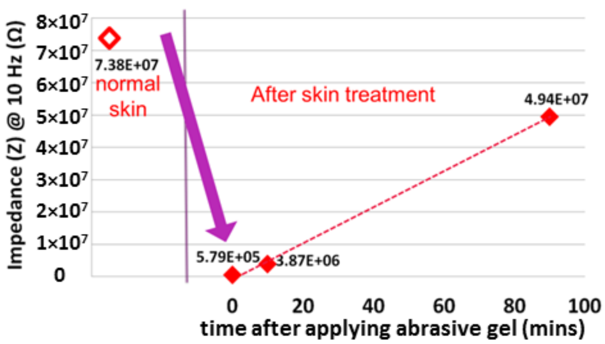

Figure 7. Dry conductive polymer electrodes impedance over time after applying abrasive gel on the skin.

These two results showed that the moisture of gel and the abrasion effect by Nuprep Skin Prep Gel were only temporary. If polymer electrodes are used to substitute the conventional wet-gel electrodes for long term measurements, the skin preparation techniques are not very useful.

IV. BIOPOTENTIAL SIGNALS MEASUREMENT

A. ECG monitoring

The impedance of the conductive polymer electrode with the highest carbon black content approached that of the gel electrode (as shown in Fig. 5), so the quality of biopotential signals acquired from these electrodes was studied.

The conductive polymer electrodes were mounted on the snap of conventional gel ECG electrodes (gel was removed) so the sticky foam parts could be used to position the electrodes on the skin. In order to compare the signals acquired from the gel and dry electrodes, wet and dry electrodes were placed next to each other on the subject's chest. Additionally, one dry electrode was placed on the bottom right chest as a bias electrode. All electrodes were connected with the imec ECG monitoring system and all signals were recorded simultaneously. Gel electrodes were placed on the skin 30 minutes before recording, to enable impedance stabilization at the contact surface.

Fig. 8 shows the raw ECG signals of both electrodes, and the ECG signal after 50 Hz signal filtering. The signal obtained with dry electrodes contains stronger 50 Hz component, which can be easily removed by filtering, resulting in similar signals for both electrodes. The R peaks and T waves could be clearly be identified.

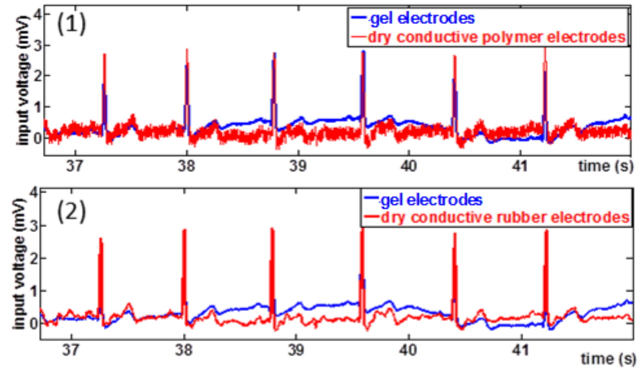

Figure 8. ECG signals recorded by gel and dry conductive polymer electrodes using a conductive polymer electrode as bias: (1) acquired data and (2) after 50Hz signal filtering.

B. EEG by commercially available headset

The non-conductive polymer electrodes with metal coating were mounted on a commercially available EEG headset. This headset was originally equipped with wet (electrolyte) sponge electrodes [10]. The dry electrodes were mounted into the headset and the conductivity of the assembly was checked by a multi meter measurement before the EEG recording started. A facial expression detection experiment was carried out by using the commercial software accompanied with the headset [11]. When making 12 different kinds of facial expressions, such as blink, right and left wink, look right and left, etc., the software tries to tell which facial expression the subject has shown, based on the signals recorded by the 16 electrodes.

The original sponge electrode and three kinds of dry electrodes were tested on a subject with long and curly hair as in Fig. 9. The subject performed each expression 20 times and the software suggested which expression was performed based on signal analysis. In Fig. 9 the result of this test is shown: the accuracy of the pattern recognition is very similar for the original wet sponge and dry electrodes. Since the expression recognition is based on the signals acquired by the electrodes, this test shows that our dry electrode achieve slightly lower performance then the commercially available one.

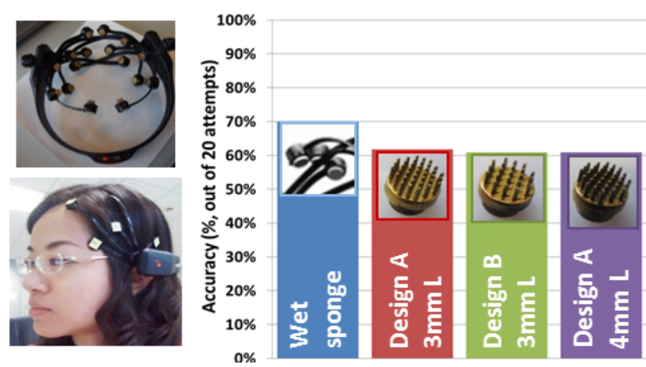

Figure 9. Accuracy of facial expression detection tests by applying sponge and different designs of non-conductive polymer with coating electrodes on commercially available EEG headset. Design A had higher pin density than design B.

It should be remarked however, that this pattern recognition is not only based on recorded EEG signals, but also on EMG, signals corresponding to the muscle movements related with the facial expressions. For further tests, it was decided to use a headset for which the raw EEG signals were visible for the user, instead of the commercial headset which did not allow studying the EEG signals directly.

B. EEG by headset with active electrodes

To test the obtained EEG signal quality with our electrodes we used an in-house developed EEG setup able to measure and provide raw EEG signals. The headset is originally equipped with Ag/AgCl electrodes, which are combined with a pre-amplifier (so-called active electrodes) in order to reduce noise [12]. For our dry electrodes with higher impedance, this pre-amplification is essential for good signal quality [13]. The signals of Ag/AgCl and dry electrodes of positions Cz and Pz were recorded when subject was sitting and relaxing with eyes open and eyes closed. The obtained signal spectral power over time is shown in Fig. 10. For both types of electrodes, the signal spectral power around 10 Hz, the frequency domain of the so-called alpha waves, was obviously stronger when the subject closed the eyes.

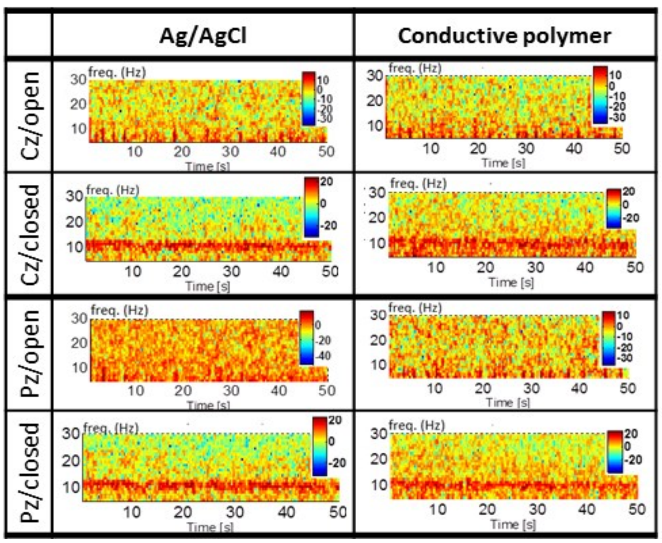

Figure 10. EEG spectral power over time at positions Cz and Pz recorded by Ag/AgCl and soft conductive polymer electrodes. Alpha-waves are clearly visible when the test subject closed his eyes for both kinds of electrodes. Color legend on the right has a unit µV2/Hz.

This test is performed on several subjects (S1, S2, S3), with similar results. The relative power of alpha waves, which is the power spectrum density (PSD) in the 5-12 Hz frequency range divided by the PSD in the 5-45 Hz frequency range, is shown in Fig. 11. The relative power is higher when subjects closed the eyes. These very promising results, in combination with the high user comfort provided by our soft electrodes, show that these dry conductive polymer electrodes have a strong potential to be applied in EEG systems.

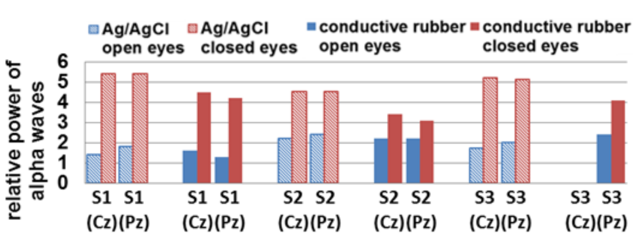

Figure 11. Relative power of alpha waves of three subjects.

V. Conclusion

Two kinds of comb-shaped polymer-based electrodes were fabricated. To characterize these electrodes, the impedance measurement and biopotential signals were performed. The dry conductive polymer electrodes with the highest carbon black content had the impedance value approaching that of the gel electrodes. The influences of skin pretreatment techniques on impedance were studied, by applying gel of conventional ECG electrodes and abrasive gel on the skin. An important impedance reduction was obtained, but the effects of skin treatment were only temporary for dry conductive polymer electrodes.

When applying the dry electrodes on imec's ECG/EEG systems, very promising results were acquired. The signals using the polymer electrodes without pre-amplification were similar to the ones using the gel electrodes. In case of EEG, pre-amplification was found to be essential for the polymer electrodes. When using such active electrode setup, alpha waves were easily detected when the subject closed the eyes. These results in combination with the high user comfort reported by all our test subjects indicate that our conductive polymer electrodes are promising candidates for high quality biopotential measurements.

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