Characterizing Contact Impedance, Signal Quality and Robustness as a Function of the Cardinality and Arrangement of Fingers on Dry Contact EEG Electrodes

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***Abstract*—Continuous monitoring of patients' electroencephalography (EEG) outside of clinical settings will be valuable for detecting the onset of medical conditions such as epilepsy, as well as for enabling patients with physically disabling conditions like amyotrophic lateral sclerosis (ALS) to communicate using a brain-computer interface (BCI). This requires the development of a wearable dry-contact EEG system that takes into account not only the signal quality but also the robustness of the system for everyday use. To this end, we investigate whether certain designs of dry electrodes lend themselves to better characteristics overall with respect to these factors. Five different metallic finger-based dry electrodes were designed and scalp electrode impedance was used to compare them under varying capping conditions, followed by an evaluation of how well they captured steady state visually evoked potentials (SSVEP). Our findings indicate that configurations with a relatively low density of fingers can more effectively penetrate through hair on the scalp and are more robust to varying conditions. This was confirmed to be a statistically significant observation through a one-sided paired t-test that resulted in a p-value < 0.004.**

I. Introduction

Dry electrodes have garnered significant interest in recent times as a favorable alternative to wet/gel-based electrodes for electroencephalography (EEG). The problems of using wet electrodes including lengthy scalp preparation times and irritation due to scalp abrasion have been well documented [1-3], and it is generally accepted that it would be unfeasible to implement a mobile, easy to use and wearable EEG system that is based on wet electrodes. There are already several existing brain computer interfaces (BCI) based on alpha rhythms, steady-state visually evoked potentials (SSVEP) and P300 [2], [4] that use dry electrodes. Such BCIs can be crucial for patient care, such as in the case of patients with amyotrophic lateral sclerosis (ALS) [5] who could regain the ability to communicate. For large scale adoption, the electrode design should perform consistently well despite variations in the scalp/hair contact and use case. With this in mind, we evaluate our electrode designs in terms of signal quality and robustness to different capping conditions.

Other works such as [3], [4] and [6] present the dry electrode design being investigated in this work: a set of metal 'fingers', meant to penetrate through the hair, arranged in a circle. These works compare dry electrode designs to conventional wet electrodes and attempt to establish correlation between them. In this work we compare different configurations of dry electrodes with each other. Moreover, to our knowledge, there has been no investigation into the optimal design for a finger-based dry electrode with regards to the number and arrangement of fingers and this work attempts to address this issue.

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II. Theory

Scalp electrode impedance can be used to compare the effectiveness of different electrode types. A lower impedance contact results in a better signal-to-noise ratio (SNR) and hence more accurate measurements of EEG. Figure 1 shows the details of the impedances involved in the contact with the scalp [3], [7]. Z_{ES} denotes the electrode scalp impedance due to the contact of the fingers. Z_S denotes the impedance of the epidermis layer of the skin, whereas Z_D denotes the impedance of the inner dermis layer. The overall impedance faced by the electrode is given by the series combination:

$$Z_{electrode} = Z_{ES} + Z_S + Z_D \tag{1}$$

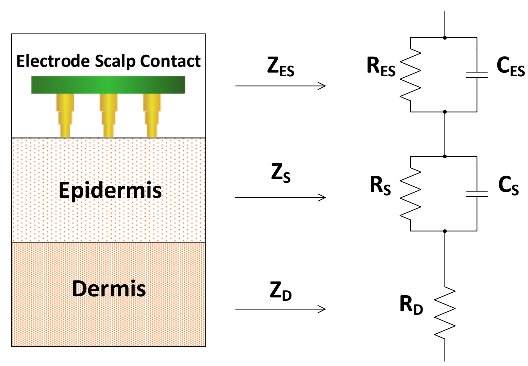

Figure 1 - Electrode skin circuit model

In this work, we investigate the characteristics of the electrode finger design that minimizes Z_{ES}, thus facilitating a better EEG recording. Since the type of finger used and the overall size of the electrodes remain constant in our designs, we hypothesize that any major differences in the performances across electrode types will be due to the effect of high resistance contact with hair for one or more individual fingers and electrode designs that avoid hair consistently will perform better.

Figure 2 shows the overall circuit model when two differential electrodes are used to measure EEG on the scalp. The positive or 'signal' electrode measures V_{sig} with respect to ground, whereas the negative or 'reference' electrode measures V_{ref} with respect to ground. If $Z_{overall}$ denotes the overall impedance between the two electrodes, we have:

$$Z_{overall} = Z_{elec,sig} + Z_L + Z_{elec,ref} \tag{2}$$

The first term $Z_{elec,sig}$ gives the impedance faced by the signal electrode, which is the same as $Z_{electrode}$ in (1). Similarly, $Z_{elec,ref}$ refers to the impedance faced by the reference electrode. In this work, the type of dry electrode used as the signal electrode is varied, but the reference electrode is always a pre-gelled adhesive patch electrode.

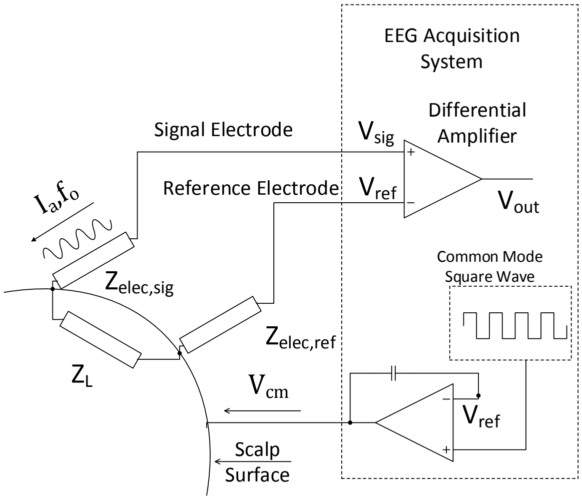

Figure 2 - Scalp impedance circuit with signal and reference electrodes

Consequently, in the model for the reference electrode, instead of the finger impedance Z_{ES} we have the impedance of the gel, Z_G. So we can re-write (2) as:

$$Z_{overall} = (Z_{ES} + Z_S + Z_D) + Z_L + (Z_G + Z_S + Z_D) \quad (3)$$

The additional term Z_L represents the impedance of the length of scalp between the two electrodes, thus completing the circuit. Using an adhesive electrode for the reference is evidently not feasible for a wearable EEG system, however for the purposes of comparing different types of dry electrodes we did not want to introduce the uncertainty of using a dry electrode for the reference as well. We assume all impedances except Z_{ES} described in (3) remain constant during experiments for the different dry signal electrodes.

Figure 3(a-e) shows the five different electrode configurations evaluated in this work. For all the electrode designs, the distance from the center to the outermost ring of fingers remains constant at 7.21mm. We used spring loaded fingers of height 4.5mm for all the designs. We also compared our designs to the state-of-the-art g.SAHARA dry electrode by g.tec shown in Figure 3(f) to ensure that the results from our custom designed electrodes are comparable to that of a commercially available one.

III. Methodology

A. Impedance Excitation Response

To estimate scalp electrode impedance, we inject current at the signal electrode, shown as I_a in Figure 2. When this applied current is a sinusoid of known frequency f_o then the frequency response of V_{out} at f_o is dominated by the voltage drop across the impedance of the circuit due to the injected I_a. The power spectral peak of the signal V_{out} at f_o is hereby termed the 'impedance excitation response', and is directly proportional to the impedance faced by the current I_a.

B. Common Mode Rejection

Another indirect measure of the scalp electrode contact is the common mode rejection ratio (CMRR) of the circuit described in Figure 2. CMRR is a measure of how well the differential amplifier can reject signals that are common to both V_{sig} and V_{ref} as these are typically comprised of uniformly received sources of noise or other undesirable signals. In our experiments, the reference electrode is always an adhesive electrode with 'ideal' skin contact. This means that the relatively higher impedance coupling of the dry signal electrode results in an impedance mismatch with the reference electrode which in turn causes an increase of the common mode at the final output [9]. In other words, as the skin contact of the dry electrode gets better, and matches that of the ideal wet contact reference, the CMRR of the circuit improves. Thus the scalp electrode contact being investigated is strongly correlated with the concept of CMRR. It must be noted that in order to measure the true CMRR of the system dry electrodes must be used for both signal and reference. However, in this work the focus is on the contact quality of the electrodes and CMRR is merely being used as a secondary measure to support the results of impedance excitation response. Using a wet electrode for reference is essential if we are to make CMRR analogous to impedance excitation response. In our setup an AC square wave of known frequency is added to the common mode (shown as V_{cm} in Figure 2). A mismatch in contact between electrodes would result in an increased presence of the known common mode signal in the output voltage V_{out}. The power at the known square wave frequency in V_{out} is inversely proportional to the CMRR, which means that it is directly proportional to the impedance faced by the signal electrode.

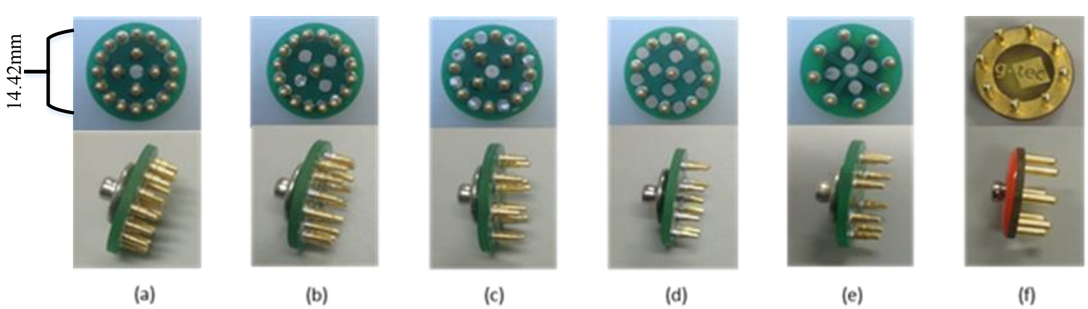

Figure 3 – (a) Circle 20: Outer ring of 16 fingers and an inner ring of 4 fingers (b) Circle 17: Outer ring of 16 fingers and one finger at center (c) Spread12: Outer ring of 8 fingers and an inner ring of 4 fingers (d) Center 9: Outer ring of 8 fingers and one finger at center (e) Default 8: Outer ring of 8 fingers (f) g.Sahara: Dry electrode by g.tec[9]

IV. EXPERIMENTAL SETUP

A. Impedance Measurement on Scalp

We defined three different use cases:

- No Adjust: The EEG cap is put on and no efforts are made to adjust the contact of the electrode
- Adjust: After putting on the cap, the electrode is twisted and pushed downwards in an effort to penetrate through the hair and make good contact
- Headband: As well as making efforts to penetrate the electrode through hair, we add a tight headband on top of the electrode to provide an additional downward push throughout for optimal contact.

The reasoning behind defining these three use cases was to determine whether certain electrode designs would exhibit more robustness in the face of varying capping scenarios.

A custom platform incorporating the TI ADS1299 EEG front end was used for data collection. One trial of the experiment consisted of about 10 seconds of current injection, with the impedance excitation response collected for each of the 10-second trials under each of the three use cases defined above. Three such trials were conducted for each electrode type in both the temporal FT8 position and the frontal AFZ position according to the 10-20 electrode placement system [10]. The cap was taken off and the hair was readjusted between trials to randomize the contact each time. The impedance excitation response was collected on six subjects, with varying amounts of hair across subjects. A 24nA sinusoidal AC current at a frequency of 30.5Hz was injected and the peak of the power spectral density (PSD) estimate at that frequency was taken as the impedance excitation response in units of mV^2/Hz. This in-band frequency is a constraint of the hardware being used; the current injection circuits are internal to the ADS1299 chip and the frequency of current injection is fixed to be in-band.

B. SSVEP SNR and CMRR Measurement

SSVEPs are the brain's response to the subject being presented with a visual stimulus flashing at a regular frequency. When the subject focuses on the flashing object, such as an LED, the EEG signals originating in the occipital region assume a marked frequency response that matches the frequency of the flashing [11]. One session of SSVEP data consisted of 4 separate trials of 10 seconds each with the subject fixating on the target flashing LED (18Hz frequency). Three such sessions were collected for each electrode type, with the cap taken off and put back on between sessions to randomize the contact. The data was collected for seven subjects in the 'Headband' case for the best possible contact scenario. There was only one signal electrode placed at the occipital location OZ, referenced to an adhesive patch at the right mastoid. Successfully captured SSVEPs would result in a peak in the PSD of the EEG data at the target frequency for each subject. For SSVEP, SNR was defined as the ratio of the target frequency PSD peak to the peaks in the nearby non-target frequencies. A higher SNR will make it easier to classify the subject's EEG response in an online BCI. In addition, the common mode square wave of 6mV amplitude was added at 61Hz throughout the tests and the PSD peak at this frequency was noted as the common mode signal power measured in units of $\mu V^2/Hz$. Since the common mode injection is done by our own custom circuit, the frequency of the signal is controllable and is set to be out-of-band for measurements simultaneously with EEG.

C. Impedance Measurement on Forearm

The impedance excitation response was measured with the signal electrode on the forearm in an area with little to no hair for seven subjects in order to confirm the effect of hair on the electrodes.

V. DISCUSSION OF RESULTS

A. Scalp Impedance Measurement

Tables I and II show the average impedance excitation responses (defined in Section III A) for the different electrode types at the FT8 and AFZ scalp locations respectively. The data is ordered according to the three use cases: 'No Adjust', 'Adjust' and 'Headband' defined in Section IV A.

TABLE I: Average impedance excitation response at FT8

	Average Impedance Excitation Response (mV^2/Hz)		
Electrode Type	***No Adjust***	***Adjust***	***Headband***
Default 8	5,637.5	1,591.3	311.3
Center 9	8,663.5	1,923.1	287.0
Spread 12	8,895.9	2,102.5	174.8
Circle 17	21,617.3	7,679.6	470.3
Circle 20	46,327.5	19,853.5	1,029.4
g.SAHARA	9,037.6	3,059.8	511.4

TABLE II: Average impedance excitation response at AFZ

	Average Impedance Excitation Response (mV^2/Hz)		
Electrode Type	***No Adjust***	***Adjust***	***Headband***
Default 8	6,812.7	1,753.61	268.2
Center 9	9,822.0	831.4	67.9
Spread 12	6,046.5	738.9	38.3
Circle 17	28,294.4	4,420.1	692.5
Circle 20	68,994.5	8,026.9	376.0
g.SAHARA	10,157.2	3,864.9	150.8

For both the scalp locations, the electrodes with lower density of fingers show better impedances in all use cases. When comparing the high density electrodes' (Circle 17 and Circle 20) impedance samples with those of the remaining low density electrodes, the one-sided t-test showed a p-value < 0.004, thus rejecting the null hypothesis that the lower density fingers show equal or higher impedance. This can be considered a statistically significant result since there are more than 100 samples of impedance for each electrode type when the data from all subjects and capping conditions is aggregated. It can also be observed that the higher density configurations show exceptionally high impedance excitation responses for the 'No Adjust' case and these responses are drastically reduced for the 'Headband' case. Repeating the above one-sided t-test for only the 'No Adjust' impedance data yields p-values < 0.005 when comparing low vs high density fingers, whereas using only 'Headband' impedance data shows p-values as high as 0.22, thus not rejecting the null hypothesis at the 5% significance level. This shows that these high density electrode types depend on effective preparation of scalp electrode contact by adjusting the electrode to penetrate through the hair, but once effective preparation is made these electrodes could perform similarly to the low-density ones. Conversely, the low density configurations do not show as much variance between the 'No Adjust' and 'Headband' cases which indicates that they are more robust in the face of varying capping conditions.

B. SSVEP SNR and CMRR

Table III shows the average SSVEP SNR values (defined in Section IV B) across all trials for all seven subjects using the different electrode types as well as their respective common mode signal powers (defined in Section III B).

TABLE III: Average SSVEP SNR and Common Mode Power

Electrode Type	Average SNR	Average Common Mode Signal Power ($\mu V^2/Hz$)
Default 8	4.097	1107.42
Center 9	4.035	818.28
Spread 12	3.927	2906.09
Circle 17	4.240	1637.51
Circle 20	3.781	4298.59
g.SAHARA	4.635	601.05

In most cases the SSVEP SNR of each electrode type is strongly correlated with its corresponding common mode signal power. As noted before, the SSVEP experiments were all conducted under ideal 'Headband' conditions, so the disparity in contact impedance between high and low density electrodes is not too large. The Circle 17 common mode power is suitably low due to this preparation, but the Circle 20 continues to have relatively poorer contact and this in turn adversely affects its SSVEP SNR. The data in general validates the assumption that EEG task performance is inextricably linked to the contact quality, and an electrode is unlikely to show a high performance with poor impedance contact. We also noted a significant performance difference between the Default 8 and g.SAHARA configurations despite their similar structure. The probable reason for this is the increased height and thickness of the fingers on the commercial g.SAHARA compared to our designs.

C. Forearm Impedance Measurement

TABLE IV: Average Impedance Excitation Response on forearm

Electrode Type	Average Impedance Excitation Response (mV^2/Hz)
Default 8	8285.49
Center 9	8502.23
Spread 12	7485.19
Circle 17	2777.65
Circle 20	3181.43

Table IV shows the average impedance excitation response collected for each of the different electrode types as they were placed on the forearm of the seven subjects in an area with no hair. The results support the hypothesis that high density configurations suffer in performance primarily due to the effect of hair. Circle 17 and Circle 20 show a marked improvement in impedance compared to the other types.

VI. CONCLUSIONS

Our findings indicate that there are some trade-offs to be considered when designing a dry finger based electrode. The higher density finger configurations showed poor scalp electrode impedances for the different use cases as shown by a one-sided paired t-test with lower density electrode impedances resulting in a p-value < 0.004. This was due to ineffective penetration through hair, as confirmed by the fact that the denser configurations showed better impedance when measured in a region of skin with no hair. However with effective preparation of the contact, SNR performance for the SSVEP EEG task improved. We can conclude that the sparser configurations are more robust to varying conditions, but the advantage is not as significant with proper electrode capping or if the subject has a lower amount of hair. In future we will also investigate the effect of varying the height and thickness of the fingers as this seems to provide an advantage to the commercial g.SAHARA electrode.

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