Signal Quality in Dry Electrode EEG and the Relation to Skin-electrode Contact Impedance Magnitude

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Abstract: Current EEG research approaches are focusing on developing new dry electrode EEG (electroencephalogram) systems providing a high enough signal quality for a wide range of applications. This study proposes several parameters for evaluating signal quality of dry electrodes and relates the results to skin-electrode contact impedance magnitude values. The EEG recordings of a Ag/AgCl pinned electrode and a flexible polymer pinned electrode are evaluated through a comparison to conductive gel electrode recordings. The experimental setup was made up of two EEG acquisition systems connected in parallel. The protocol included open eyes, closed eyes and steady-state visually evoked potentials (SSVEP) sessions in both seated and walking in place conditions. The parameters used for evaluation were signal correlations, signal coherence and signal-to-noise ratios (SNRs). Results showed that the three proposed parameters provided equivalent outcome for signal quality estimation for the same recordings. There was no relation reported between the defined signal quality and the skin-electrode contact impedance in either dry or gel electrodes, although high impedance variations were present among subjects. However, larger impedance magnitude and impedance magnitude variations, and lower signal quality is observed for dry electrodes compared to gel ones.

1 INTRODUCTION

Research in the field of medical technology is currently focusing on developing devices that can facilitate disease prevention and remote monitoring for an easier, more successful patient treatment, while also reducing costs. Due to its non-invasive nature and high temporal resolution, scalp electroencephalographic signals (EEGs) have good prospects for many applications. Electrical activity of the brain can be useful in the diagnosis and monitoring of several neurological conditions, including the detection of epileptic seizures, diagnosis of sleep disorders or stroke rehabilitation (Teplan, 2002). Other applications include brain computer interfaces (BCIs) used either in rehabilitation of impaired individuals or for communication with locked in patients (Minguez et al., 2009), cognitive enhancement trainings for ADHD (Moriyama et al., 2012), and remote monitoring of drivers (Lin et al., 2005).

Currently available EEG technologies are mostly used in controlled research and clinical environments. One of the biggest limitation comes from the electrodes used to obtain good signal quality. A widely accepted method of acquiring high quality EEG signals is through conductive gel electrodes and involves skin preparation. Skin preparation includes skin abrasion and cleansing to remove dead cells from the top layer of the epidermis, causing a decrease in the skin-electrode contact impedance. After this step, the electrodes with conductive gel are applied. This procedure is cumbersome, uncomfortable and can be painful for the patient. If proper care is not taken, adjacent electrodes might get short circuited in high electrode density montages. Moreover, the conductive paste might dry over time, increasing the skin-electrode contact impedance. Therefore, to ensure proper EEG recordings, a trained technician is required for both gel electrode application and for maintenance throughout the entire measurement.

Several research groups are focusing on

improving dry sensor technologies to develop a more practical, easy-to-use, EEG measurement system. Although several types of dry electrodes are already commercially available, EEG users and clinicians are reluctant to adopt them in their regular practices as the lack of conductive gel implies a higher skin-electrode contact impedance and makes the recordings more susceptible to noise and interference. To this day, the quality of these electrode types has not been proven to reach that of the standard gel electrodes. Hence, there is a need to develop a method of testing the quality of the signal obtained from the newly designed biomedical sensors.

A literature survey on the available state-of-the-art EEG signal quality evaluation methods revealed that several protocols are available but there is no clear consensus on the best evaluation procedure (Tăuţan et al., 2013). The lack of agreement on what high EEG signal quality means comes from the fact that different applications require high quality in different aspects of the EEG signal, i.e., a signal that provides good performance in BCIs might prove insufficient for clinical diagnosis. A general purpose evaluation framework should take into account as many signal features as possible, from the time, frequency and spatial domain, to ensure a good quality for a wider variety of applications.

Due to the temporal, spatial and subject variability of the EEG signal, all procedures of evaluating dry electrode recordings make use of a comparison to gel electrode recordings through either a parallel method (the two sensor types record concurrently) or a serial method (the two sensor types record in turn from the same scalp positions) (Ruffini et al., 2008). Most encountered parameters for evaluating EEG signal quality are dry-gel signal correlations and comparisons of the power spectral densities (Ruffini et al., 2008; Gargiulo et al., 2010; Estepp et al., 2005). Other evaluation parameters include signal-to-noise ratios of steady-state visually evoked potentials (Chi et al., 2012) and deflection amplitudes of P300 components of event related potentials (ERPs) (Zander et al., 2011). Performance of the electrodes in several applications, such as BCIs and cognitive monitoring, are also used as an indication of signal quality (Estepp et al., 2005; Gargiulo et al., 2010; Sellers et al., 2009).

When using gel electrodes, it is generally considered that the signal quality is good when the impedance magnitude value measured is low (typically below 5kΩ). Only one study found in literature reported that EEG signal quality did not modify when skin abrasion was not used and the skin-electrode contact impedance reached values of 40kΩ (Ferree et al., 2001). However, it is not known whether even higher impedance, or difference in impedance among electrodes would result in a lower signal quality. Also, to the best of our knowledge, the relation between signal quality and skin-electrode contact impedance is not addressed in the case of dry electrodes. We believe that since the impedance is much higher and since it can be quite different among electrodes it can also be used to estimate the signal quality obtained.

This paper builds on a previous study of EEG signal quality evaluation (Tăuţan et al., 2013), improving the experimental setup and incorporating skin-electrode contact impedance measurements. In our previous study, several parameters were defined to characterize signal quality. The limitations we tried to overcome in this evaluation are the accuracy of the visual stimulation rendering, the synchronization of the stimulation with the recordings and improper contact of the electrodes with the scalp for some subjects due to inadequate headset design. The procedures used for signal quality evaluation include open eyes, closed eyes and steady-state-visually evoked potentials (SSVEPs) recordings performed both while the participant is seated and walking. The parameters defined to quantify signal quality include: signal correlations, signal coherence and signal-to-noise ratios.

The paper is organized as follows. The second section describes the experimental setup, the protocol and the data analysis methods. In the third section, results are presented and discussed. Lastly, conclusions are drawn and suggestions for further research are presented.

2 METHODS

2.1 Experimental Setup

The materials used to build the experimental setup are the following:

- 6 standard gel-filled Ag/AgCl cup electrodes (10mm diameter) used as a reference system
- 4 BioPac EL120 Ag/AgCl dry pin electrodes (10.2mm diameter with 12 pins of 1mm width) used as test electrodes
- 4 conductive polymer electrodes with flexible pins (13mm diameter with 15 pins of 1.2mm width) used as test electrodes (Chen et al., 2013)
- wireless EEG headset used as rigid support for the dry electrodes (Patki et al., 2012)

- EEG boxed system used for connecting the gel electrodes (Mihajlović et al., 2013)
- PC for recording and displaying visual stimulation

This setup is consistent with that found in the previous work: a parallel measurement system is used to make a comparison between signals recorded with dry and gel electrodes. Here, different tools are used to incorporate impedance measurements and to improve the shortcomings of the previous setup by providing more accurate stimulation and better synchronization. To prove the validity of the developed evaluation framework, two types of dry electrodes are tested: Ag/AgCl and polymer electrodes.

2.1.1 Data Acquisition and Stimulation Display

EEG signals and impedance data are acquired concurrently from dry and gel electrodes using the EEG v2.0 wireless headset and boxed system, respectively. These two systems include the same electronics, ensuring that all signals are collected in the same way. Each measurement system contains the following: proprietary active electrodes, EEG analog front-end application-specific integrated circuits (ASIC) with an input impedance of 1.2 GΩ at 10Hz and a built-in current generator, microcontroller, radio and power management circuitry (Patki et al., 2012).

These systems enable continuous skin-electrode contact impedance monitoring per channel. This is done by injecting a square wave current at a frequency of 1024 Hz through the active electrode front ends (Patki et al., 2012). The amplitude of the wave is set manually by the operator for each subject according to his/her skin characteristics and the electrodes used. Typical values are 10nA, 20nA or 50nA. Impedance magnitude is obtained by combining the demodulated first harmonic of the in-phase and quadrature components of the impedance signals.

The acquisition system is limited by the 5mV dynamic range of the input amplifier. In order to keep the acquired signal unattenuated, the maximum input resistance is 500kΩ when the injected current is at the minimum of 10nA.

Functions from the MatLab Psychtoolbox were used to obtain accurate rendering of the SSVEP, while taking into account the different processes running in the background (Brainard, 1997). Since both EEG systems communicate wirelessly to their corresponding receivers, time synchronization is performed in the software through a Matlab script. The script handles the data packets received from both systems and synchronizes them with respect to each other and to the rendered stimuli.

2.1.2 Electrode Positions

As the framework aims at comparing dry and gel sensors, the electrode arrangement is extremely important. First, consistency in positioning must be maintained between participants. Second, relevant locations should be chosen according to the type of information that needs to be obtained (from the type of sensory information provided as stimuli). Third, the EEG signal represents the summed activity of millions of neurons and thus the spatial variability introduced should be accounted for.

To this end, the electrode configuration is similar to that presented by (Estepp et al., 2005). Two groups of one dry electrode and two gel electrodes were placed on the scalp of the participant on the vertices of an equilateral triangle with a 2cm edge (Fig. 1). The dry electrodes were positioned using the EEG headset. They were centred at the Cz and Pz positions of the standard 10-20 International Positioning system. The gel electrodes were positioned with the help of metal markers.

Separate reference and ground electrodes are used for the dry and gel systems to avoid any additional differences introduced by the different skin-electrode contact impedance. The two references are placed close together behind the right ear, while the two ground electrodes are located behind the left ear.

Comparisons are made between the dry test electrode and one of the gel electrodes placed in its vicinity. To give an estimate of the effects of spatial variability, the same computations are made between the two adjacent gel electrodes.

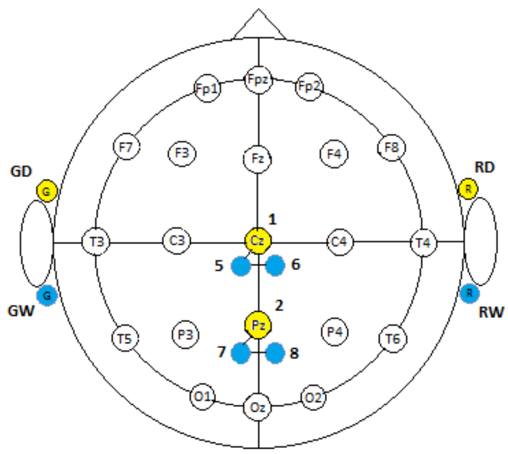


Figure 1: Electrode positioning system. The yellow circles represent dry electrodes, whereas the blue circles are gel electrodes.

2.2 Protocol

Six subjects (all male) aged between 25 and 37 years volunteered for participation in the experiment. After being informed of the content of the experiment and making sure that they do not present any skin allergies or disorders triggered by oscillatory light stimuli, participants signed a consent form. All of them presented normal or corrected to normal sight.

The protocol was approved by the internal review board of Holst Centre. The total experimentation time per participant amounted to 60 minutes. The participants were comfortably seated at a distance of 50 cm from the LCD display where the visual stimulation was presented. They were advised to avoid movement as much as possible to reduce the appearance of motion artifacts. First, the wireless headset was equipped with the four Ag/AgCl dry electrodes and was mounted on the head of the participant. After that, the skin was abraded to eliminate dead cells and the gel electrodes were applied. The quality of the recordings was checked by the operator through visual inspection of the obtained signal. Also, setting the appropriate value of the injected current such that the system can perform reliable impedance measurement was done by the operator. Then, the first recording stage was performed. In the second stage, the Ag/AgCl dry electrodes were replaced with the polymer electrodes and the same protocol was repeated. The protocol consisted of the following paradigms: continuous recordings and SSVEP stimulation.

2.2.1 Continuous Recordings

Continuous EEG recordings are obtained from four paradigms. In the first one, the participant had to keep his/her eyes open for approximately 1 minute (open eyes) while seated. In the second one, the participant had to close his/her eyes for about 1 minute (closed eyes) while seated. For the last two paradigms the subject performed the same two actions while walking in place for about 1 minute.

2.2.2 SSVEP Stimulation

The stimulation in this case was a white flickering square centered on a black screen. The stimulation frequency was 4Hz. This value was chosen to have the first harmonic of the response outside the alpha range (8-13Hz) and so avoid interference. Subjects had to observe the stimulation for approximately 1 minute in two conditions: once while seated (SSVEP seated) and once while walking in place (SSVEP walking).

2.3 Data Analysis

Data processing was performed in Matlab. In the pre-processing stage, both the EEG and impedance signals were band pass filtered between 2-30Hz. A Chebyshev type II filter was applied forward and backwards on the data to eliminate distortion (Acunzo et al., 2012). Severe artifacts are removed from all recordings. This is done manually by the operator with the help of a thresholding algorithm based on the standard deviation of the signal (Tăuţan et al., 2013).

The comparison of the dry and gel recordings should provide information from both the time and frequency domain. Thus, several parameters were computed in both domains: signal correlations (time), signal coherence (frequency) and SNR (frequency). For validation of the framework, mean skin-electrode contact impedance values are also reported and they are compared to the signal quality obtained at each electrode. This section presents the methods used for defining and computing these parameters.

2.3.1 Signal Correlations

Pearson's product moment correlations are used to quantify the similarity between the dry and wet recordings as they provide information on the time coupling and wave morphology (Guevara and Corsi-Cabrera, 1996). Correlation coefficients are computed between dry-gel electrode pairs (1-5, 2-7) and also between gel-gel pairs (5-6, 7-8) on the open eyes and closed eyes recordings.

2.3.2 Signal Coherence

While signal correlations give an idea of the time domain similarity of two signals, signal coherence shows the stability of the similarity by looking at the frequency content. They provide information on the changes in power and phase of the signals with respect to each other, disregarding signal polarity (Guevara and Corsi-Cabrera, 1996). The coherence function between the two signals is computed using Welch's method on a window of 1 second and with a window overlap of 75%. The mean coherence is computed over the band between 2-30Hz. Values are obtained for the dry-gel pairs (1-5, 2-7) and for the gel-gel pairs (5-6, 7-8) on the open eyes and closed eyes recordings.

2.3.3 Signal-to-Noise Ratio

A signal-to-noise ratio (SNR) is defined to quantify the level of contamination of the recordings observed in the frequency spectrum. The alpha wave peaks and

Table 1: Mean correlation values across participants.

		Cz			Pz		
	Paradigm	**1-5**	**1-6**	**5-6**	**2-7**	**2-8**	**7-8**
Electrode A Ag/AgCl	Open eyes seated	0.77	0.75	0.93	0.71	0.70	0.91
	Closed eyes seated	0.82	0.81	0.97	0.80	0.78	0.94
	Open eyes walking	0.27	0.29	0.78	0.30	0.32	0.83
	Closed eyes walking	0.43	0.42	0.79	0.46	0.43	0.79
Electrode B Polymer	Open eyes seated	0.48	0.37	0.85	0.46	0.46	0.93
	Closed eyes seated	0.60	0.59	0.96	0.53	0.50	0.95
	Open eyes walking	0.35	0.29	0.83	0.32	0.26	0.71
	Closed eyes walking	0.50	0.41	0.80	0.44	0.24	0.65

the SSVEP responses are stronger when less noise is present on the recordings and thus can be used to estimate the difference in noise content of the dry and gel electrode signals. Assuming that the noise signal is proportional to the standard deviation of the signal (Mihajlović et al., 2012), the following definition is proposed:

$$SNR = \frac{mean(PSD_{band\ of\ interest})}{mean(PSD_{signal\ band-band\ of\ interest})} \quad (1)$$

The signal power is defined as the mean power spectral density (PSD) in the band of interest, while the noise power is the mean PSD outside this band. For the closed eyes recordings, the band of interest corresponds to the alpha band (8-13Hz) while for the SSVEP recordings, it is between the following intervals: 3-5, 7-9, 11-13, 15-17 corresponding to the stimulation frequency and its harmonics. The PSD of all recordings is computed with Welch's spectrum estimation method on a 1 second window with 75% window overlap.

2.3.4 Impedance Value Analysis

For the analysis of the impedance data, the magnitude is extracted from the in-phase and quadrature components of the impedance signal by taking into account the amplitude of the injected current. Mean values and standard deviations are reported for all dry and gel electrodes.

3 RESULTS

3.1 Signal Correlations

Mean correlation values across participants are presented in Table 1, classified according to the electrode types tested and the paradigms used for recording. In the case of the Ag/AgCl test electrodes, one participant was excluded due to improper contact of the gel electrodes. For the polymer electrodes, recordings on the Cz site of one participant, for the walking paradigms, were excluded also due to improper gel electrode contact. Additionally, measurements with polymer electrodes could not be performed on two of the subjects as the skin-electrode contact impedance was higher than the limit imposed by the amplifier. An example of the signals obtained with different electrodes in seated and walking conditions can be seen in Fig. 2a-d.

Gel-gel signal comparisons resulted in high values for the correlation coefficients in the case of the seated paradigms, indicating a strong similarity between the recordings. These values are useful in estimating the effect of spatial variability on the content of the signal. Stronger correlations are reported when the participants had their eyes closed, due to the increased alpha activity. For the walking paradigms, the coefficients decreased due to the motion artifacts introduced and ranged between 0.65 to 0.83. Also, differences in correlation for eyes closed and eyes open conditions was not consistent.

Coefficients obtained for the dry-gel comparison varied substantially with electrode type and paradigm. Generally, they followed a decreasing trend, from Cz to Pz electrodes and from closed eyes to open eyes paradigms, in both the seated and walking cases. This is consistent with literature (Estepp et al., 2009). The computed values for the Ag/AgCl electrodes were higher than those obtained in our previous seated study. This is caused mainly by the use of active electrodes designed to cope with higher input impedance. The walking paradigms induced a more drastic decrease in the dry-gel correlation values than in the gel-gel ones, revealing a greater sensitivity of dry electrodes to motion artifacts.

The polymer electrodes had a poorer performance than the silver ones: in two cases signal acquisition was not possible while the correlation coefficients obtained were lower for the seated paradigms. For the walking paradigms, the coefficients were in the same

Table 2: Mean coherence values in the frequency band 2-30Hz across participants.

		Cz			**Pz**		
	Paradigm	**1-5**	**1-6**	**5-6**	**2-7**	**2-8**	**7-8**
Electrode A Ag/AgCl	Open eyes seated	0.54	0.51	0.84	0.50	0.49	0.77
	Closed eyes seated	0.59	0.55	0.86	0.54	0.54	0.77
	Open eyes walking	0.12	0.11	0.70	0.16	0.15	0.64
	Closed eyes walking	0.20	0.19	0.77	0.26	0.23	0.62
Electrode B Polymer	Open eyes seated	0.27	0.24	0.83	0.24	0.23	0.79
	Closed eyes seated	0.34	0.34	0.87	0.32	0.29	0.81
	Open eyes walking	0.18	0.14	0.70	0.17	0.10	0.50
	Closed eyes walking	0.24	0.18	0.71	0.23	0.11	0.46

Table 3: Mean SNR values in dB.

		Cz			**Pz**		
	Paradigm	**1**	**5**	**6**	**2**	**7**	**8**
Electrode A Ag/AgCl	Closed eyes seated	7.80	7.98	7.58	8.72	9.87	10.14
	Closed eyes walking	2.12	3.72	3.12	1.80	3.73	4.57
Electrode B Polymer	Closed eyes seated	4.30	6.31	6.17	5.24	8.90	8.42
	Closed eyes walking	1.90	2.15	1.55	1.91	2.24	4.52

range as for the Ag/AgCl electrodes. However, a comparison of the behavior of the two electrodes with respect to gel ones in motion recordings is difficult to make since gel-gel correlations have also changed.

3.2 Signal Coherence

Table 2 reports the mean coherence values obtained from the same recordings as those used for determining signal correlations. Coherence values are slightly lower than the correlation values. This indicates power asymmetries between recordings that could be caused by the different skin-electrode contact impedance and its variation over time. It is expected that the dry electrode impedance slightly decreases over time as the contact with the skin stabilizes and the signal settles, while the gel electrode impedance increases over time due to the drying of the gel.

Nevertheless, the results obtained follow the same trends as those of the correlation computations. Gel-gel comparisons present the highest values across all paradigms. Dry-gel results are substantially lower than the gel-gel ones and values decrease drastically in the case of the walking paradigms. The artifacts introduced through walking are superimposed on the EEG signal, introducing new frequency components in the recordings. Each electrode records the additional interference differently due to the differences in skin contacts as a result of motion. The results confirm that gel electrodes are still better when dealing with motion. Without the gel to ensure adhesion, dry electrode recordings are more susceptible to walking artifacts. Here as well, Electrode A presents higher values than Electrode B.

3.3 SNR

Signal to noise ratios were computed on the closed eyes and SSVEP paradigms while walking and seated. The values obtained are summarized in Table 3 and examples of spectra for the closed eyes paradigms are shown in Fig. 2e-f. For the closed eyes paradigms, the same recordings were excluded from the analysis as in the case of correlation and coherence computations. For the SSVEP, one subject was eliminated due to improper gel contacts for the silver electrode and for the polymer electrode, the same two subjects as in the case of continuous recordings presented no measurable signal.

The SNR values for the SSVEP paradigm were not reported in Table 3 as they did not reflect the observations made through the visual inspection of the signals' spectra. The visual stimulation frequency was too low and thus the first harmonic of the SSVEP response was superimposed on the strong low frequency components present in the dry electrode recordings. As the electrodes are distant from the visual cortex, higher harmonics could be observed on some of the recordings, however their intensity was too low. Thus, no accurate distinction could be made between the peaks of the specific cortical response and the background EEG and noise.

As can be seen in Table 3, the SNRs of the closed

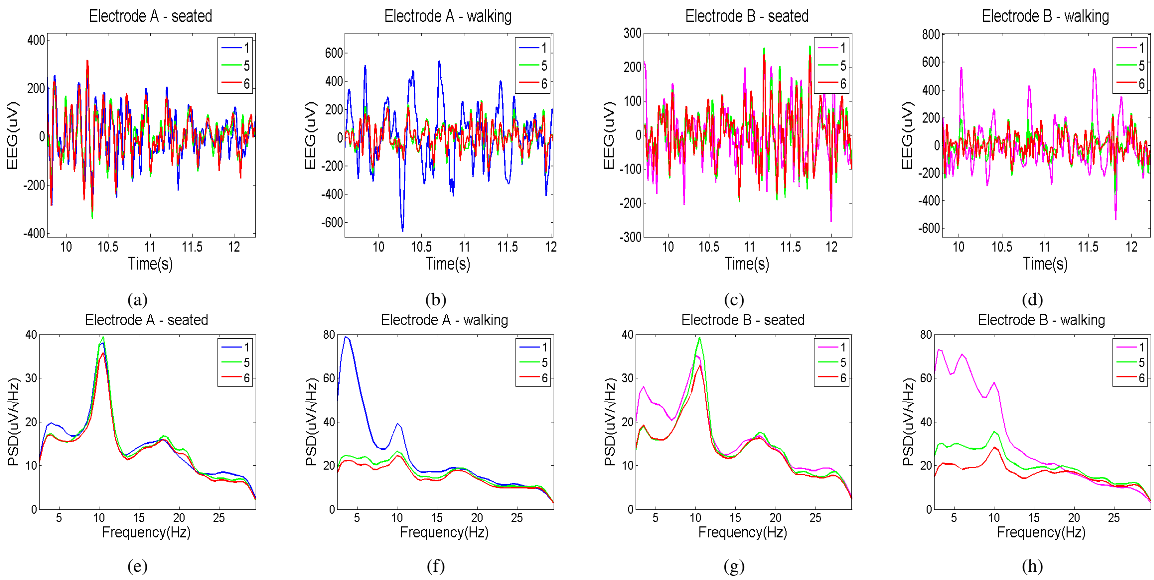


Figure 2: Closed eyes seated and walking recordings of Subject 1 for electrode 1 and 2 at the Cz position. First row presents approximately 2 seconds of the time course of the EEGs while the second row shows the spectrum of the entire recordings. For the legend, see Fig. 1 for the electrode position numbering.

eyes recordings were positive for all electrodes as the alpha wave phenomena could be observed on the spectra of both seated and walking recordings (see Fig. 2e-f). Generally, the values obtained from the parietal site are higher than those from the central position. For the closed eyes seated paradigms, the results for the gel electrodes are comparable to those obtained in our previous study. The Ag/AgCl electrodes exhibited higher SNRs mostly due to the active electrodes used for acquisition. Their performance is now comparable to that of gel electrodes. The polymer electrodes presented higher differences between the SNRs of dry electrodes and of gel electrodes indicating a higher level of noise.

For the walking recordings, as expected, the SNR values decrease. However, the alpha peak can still be seen (Fig. 2f and Fig 2h). For Electrode A, the walking spectra indicates strong low frequency components introduced by the motions. These components are not present on the spectra of the corresponding gel recordings (Fig. 2f) or on the spectra of the seated paradigm (Fig. 2e). These observations are confirmed by the SNR values reported, where a stronger difference exists between the dry and gel SNRs for the walking paradigms than for the seated ones. For Electrode B, the values obtained are lower than those found for Electrode A. A higher sensitivity to motion artifacts causes stronger low frequency components that interfere with the alpha response as their band extends over 10Hz (Fig. 2h).

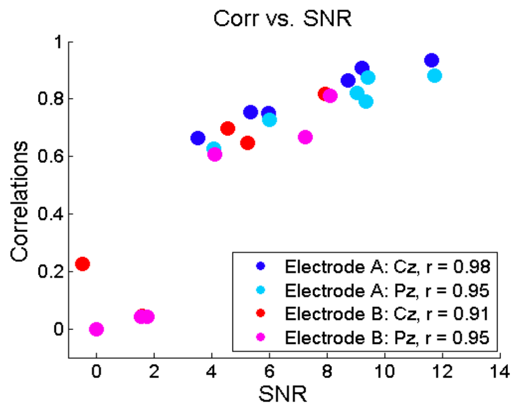


Figure 3: Comparison of the Correlations and SNRs of dry electrodes per subject for the closed eyes seated recording.

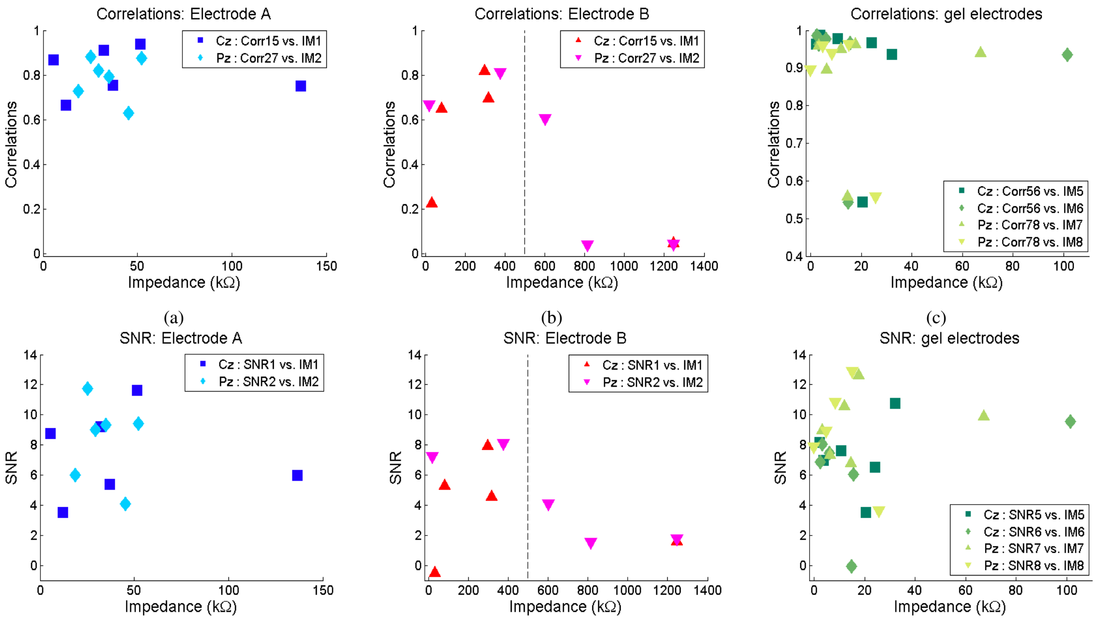
Fig. 3 shows a plot of the correlation values obtained for all dry electrodes per subject versus their corresponding SNR. Correlation coefficients were computed to characterize the relationship between the two variables. The r value for each electrode is mentioned in the legend of Fig. 3. A strong correlation exists between the two defined parameters. Thus, the quality defined in the time domain by the correlation coefficients is comparable to the one defined in the frequency domain by the SNR.

3.4 Impedance and Signal Quality

Table 4 summarizes the results obtained for the

Table 4: Mean (M) and standard deviation (S) skin-electrode contact impedance values in kΩ.

	Paradigm	Cz				Pz			
		1		5		2		7	
		M	S	M	S	M	S	M	S
Electrode 1 Ag/AgCl	Open eyes seated	54.2	61.6	13.6	12.7	34.4	18.7	15.2	16.7
	Closed eyes seated	47.8	52.9	14.6	13.0	37.6	11.2	21.3	26.2
	SSVEP seated	56.4	48.1	26.7	31.4	40.3	16.1	26.9	35.5
Electrode 2 Polymer	Closed eyes seated	212.4	147.1	29.5	21.9	530.0	43.8	23.9	19.5
	Closed eyes seated	183.5	146.8	33.3	25.1	453.9	339.8	27.0	21.3
	SSVEP seated	184.7	127.7	36.6	25.7	396.8	268.7	34.5	30.5



(a) (b) (c) (d) (e) (f)

Figure 4: Comparison of parameters defining signal quality to skin-electrode contact impedance.

continuous impedance analysis. For each measurement electrode, the average of the impedance signal mean values across all included participants and their corresponding standard deviation are presented. The analysis was performed only on the seated paradigms as the walking paradigms introduce artifacts on the impedance signals which influence the mean impedance value in an unpredictable way. Results found for both dry electrode types were greater than those of the gel electrodes. Gel electrodes presented a minimum mean impedance value of 13.6 kΩ and a maximum of 36.6 kΩ both found for electrode 5 in the Cz position. The Ag/AgCl electrodes had values ranging from 34.4 kΩ in the Pz position to 56.4 kΩ in the Cz position, while the polymer electrode impedance was larger, having a minimum of 183.5kΩ at Cz and a maximum of 530.0 kΩ at Pz. The large discrepancies between the values of the skin-electrode impedance of the same electrode type at different locations might be connected to the different contacts between the sensor and the scalp of the participant. High standard deviations are reported for all electrodes. This can be motivated by the different skin properties of each participant and by the differences in measurement conditions. However, larger variations can be observed in the dry electrode impedance values than in the gel ones.

A comparison between the parameters defined for quantifying signal quality and the corresponding skin-electrode contact impedance obtained is made in Fig. 4. Correlations and SNR values of the "test" electrodes are plotted with respect to their impedance values for each subject. No relation can be observed between the impedance and the corresponding dry-gel correlations for Electrode A (Fig. 4a). Electrode B presents a decrease in quality and an increase in impedance compared to Electrode B. Although its impedance values vary a lot, the signal quality obtained seems unaffected. This also suggest no connection between the two variables. Note that the acquired EEG signals for which the skin-electrode contact impedance was higher than 500kΩ were attenuated due to the amplifier saturation and so they are not included in this analysis (see Figure 4b and Fig. 4e).

The same trend is observed for the SNR investigation. Measurements for Subject 4 were not taken with Electrode B due to a very large impedance value. Also, outliers in the SNR and correlation plots of this electrode type represent measurements that were excluded from previous analysis as no EEG signal was observed during visual inspection of the recordings. The results obtained indicate that a large increase in the skin-electrode contact impedance has no effect on the signal quality defined through our parameters. However, a larger data set should be used to confirm these results as the number of available data points was limited to 12 per electrode type.

Gel electrode performance was also evaluated with respect to the corresponding skin-electrode contact impedance. Impedance values presented a large variation and included values above 40 kΩ. Both in the case of correlations and SNR, gel electrodes presented high scores regardless of their impedance value (Figure 4(c) and 4(f)). A lower performance was seen in several gel electrodes, but their impedance value was inside the mean range. These results confirm the findings of Ferree et. al (Ferree et al., 2001) and point out that signal quality is not depended on the impedance magnitude, even when these values are much higher than 40kΩ. However, they also show that electrode material has an impact on both impedance magnitude and obtained EEG signal quality.

4 DISCUSSION

The dry-gel comparison made by using a gel-gel comparison as a benchmark, proved to be a reliable framework for dry electrode EEG signal quality evaluation. Differences between subjects and electrode types could easily be observed and compared, providing useful information on the introduced variability.

All three parameters proposed for evaluation, namely signal correlations, signal coherence and SNR, presented lower values for dry electrodes when compared to gel electrodes. The results suggest that better EEG signal quality is obtained through gel electrodes, followed by the Ag/AgCl electrodes and then by the polymer electrodes. Compared to our previous study, the performance of the Ag/AgCl electrodes was better as a consequence of the use of active electrodes designed to cope with dry electrode recording conditions. The mean coherence values reported were lower than the correlation coefficients obtained, but followed the same decreasing trends. For the electrodes that presented strong dry-gel correlations, high SNR values were also reported. Since the mean coherence, correlation coefficients and SNRs present the same trends and thus equivalent conclusions regarding signal quality, one of them can be considered sufficient for a fast characterization of EEG recordings.

The mean values for the skin-electrode contact impedance magnitude of the gel electrodes was the lowest reported, while the highest was that of the polymer electrodes. When comparing the defined signal quality to the values of the impedance, no trend was observed for different recordings with the same electrode although impedance variations were present. Skin-electrode contact impedance magnitude values that exceeded the recommended 5kΩ threshold were reported for the gel electrodes. However, there was no indication of a decrease in performance, either from the visual inspection of the recordings or from the values obtained for our parameters. The lack of a relationship between EEG signal quality and skin-electrode contact impedance magnitude can be a consequence of the system features: active electrodes combined with high input impedance ASICs that were developed specifically to eliminate the influence of high impedance values. Thus, the recordings are expected to be less dependent on the contact impedance. Nevertheless, differences in recording quality can still be observed between different electrode types.

The evaluation performed on the dry electrodes revealed several disadvantages of this technology. During the walking paradigms, a higher susceptibility to motion artifacts was observed in the dry electrode types. Less adhesion to the skin is one of the factors that contributes the most to this effect. Also, the spectra of dry electrode recordings presented strong low frequency components that extended over the 4Hz SSVEP stimulation frequency even in the case of the seated paradigm. These low frequency components are attributed to signal drifts due to the changes in the skin-electrode contact over time. To cope with these disadvantages, a solution for handling motion needs to be found, either through system or algorithm design or by creating electrodes with different contact properties. Furthermore, to give a better characterization of the sensors, studies on contact impedance variations over time should be carried out. Electrodes that have shorter settling time and that do not suffer from signal drifts are needed to permit a faster, more reliable signal acquisition while eliminating the undesired frequency components.

Due to the limitations of the proposed framework and study design, conclusions regarding the relation between signal quality and skin-electrode contact properties are difficult to formulate. Future studies should include a larger number of participants, both male and female, for a more thorough, unbiased analysis. During some of the experimental sessions, gel electrodes were disconnected or made a poor contact with the scalp of the participant. Also, due to the constraints imposed by the rigid headset used for dry electrode positioning, proper contact for dry electrodes was not always obtained. Variability in the size and head shape of the participants combined with the different hair types, made the contact between the skin and the electrode pins difficult. Results also indicate that the reported outliers in signal quality might be related to a bad contact of the electrodes to the skin and not to the high skin-electrode contact impedance magnitude.

Improvements to the proposed evaluation framework should include a change in the stimulation frequency of the SSVEP protocol. To avoid interference from the strong low frequency components, the alpha band frequency range and the frequencies introduced by the walking paradigms, a flickering frequency higher than 13Hz is recommended. Also, placing the recording electrodes above the occipital lobe would permit the capture of a stronger SSVEP response as this site is closer to the visual cortex. A protocol for eliciting event-related potentials (ERPs) should also be included in the evaluation as ERPs are often used in BCIs and psychological studies. Moreover, the ERPs have a very low amplitudes and thus are very sensitive to different sources of interference.

Overall, better EEG systems and electrode designs are needed to permit the usage of dry electrode technology in a wide range of applications requiring high quality signals. The skin-electrode contact impedance broadly characterizes the interface between the electrode and the skin. To allow the design of better performing electrodes, more work needs to be carried out in identifying the exact contact characteristics that have the largest impact on signal quality. Some of the approaches that could assist in this process involve studying the skin-electrode interface by analyzing the real and imaginary parts of the impedance signal or by creating models of this interface.

REFERENCES

Acunzo, D., Mackenzie, G., and Rossum, M. v. (2012). Systematic biases in early ERP and ERF components as a result of high-pass filtering. *Journal of Neuroscience Methods*, 209:212–218.

Brainard, D. H. (1997). The Psychophysics Toolbox. *Spatial Vision*, 10:433–436.

Chen, Y.-H., de Beeck, M. O., Vanderheyden, Luc Mihajlović, V., Grundlehner, B., and van Hoof, C. (2013). Comb-Shaped Polymer-Based Dry Electrodes for EEG/ECG Measurements with High User Comfort. In *35th Annual International IEEE EMBS Conference*.

Chi, Y., Wang, Y.-T., Wang, Y., Maier, C., Jung, T.-P., and Cauwenberghs, G. (2012). Dry and Noncontact EEG Sensors for Mobile Brain-Computer Interfaces. *IEEE Transactions of Neural Systems and Rehabilitation Engineering*, 20(2):228–235.

Estepp, J., Monnin, J., Christensen, J., and Wilson, G. (2005). Evaluation of a Dry Electrode System for Electroencephalography: Applications for Psychophysiological Cognitive Workload Assessment. In *Proceedings of the 11th International Conference on Human Computer Interaction*, Las Vegas, Nevada, USA.

Estepp, J. R., Christensen, J. C., Monnin, J. W., Davis, I. M., and Wilson, G. F. (2009). Validation of a Dry Electrode System for EEG. In *Proceedings of the Human Factors and Ergonomics Society 53rd Annual Meeting*, pages 1171–1175, San Antonio, Texas, USA.

Ferree, T. C., Luu, P., Russell, G. S., and Tucker, D. M. (2001). Scalp electrode impedance, infection risk, and EEG data quality. *Clinical neurophysiology : official journal of the International Federation of Clinical Neurophysiology*, 112(3):536–44.

Gargiulo, G., Calvo, R., Bifulco, P., Cesarelli, M., and Ohamed, A. (2010). A New EEG recording system for passive dry electrodes. *Clinical Neurophysiology*, 121:686–693.

Guevara, M. A. and Corsi-Cabrera, M. (1996). EEG coherence or EEG correlation? *International Journal of Psychophysiology*, 23:145–153.

Lin, C.-T., Wu, R.-C., Liang, S.-F., Chao, W.-H., Chen, Y.-J., and Hung, T.-P. (2005). EEG-Based Drowsiness Estimation for Safety Driving Using Idependent Component Analysis. *IEEE Transactions on Circuits and Systems*, 52(12):2726–2738.

Mihajlović, V., Garcia-Molina, G., and J., P. (2012). To What Extent Can Dry and Water-based EEG Electrodes Replace Conductive Gel Ones? In *BioDevices conference*, Vilamoura, Algarve, Portugal.

Mihajlović, V., Li, H., Grundlehner, B., Penders, J., and Schouten, A. (2013). Investigating the Impact of Force and Movements on Impedance Magnitude and EEG. In *IEEE Engineering in Medicine and Biology Society*, Okata, Japan.

Minguez, J., Kubler, A., and Antelis, J. (2009). A Noninvasive Brain-Actuated Wheelchair Based on a P300 Neurophysiological Protocol and Automated Navigation. *IEEE Transactions on Robotics*, 25(3):614–627.

Moriyama, T. S., Polanczyk, G., Caye, A., Banaschewski, T., Brandeis, D., and Rohde, L. (2012). Evidence-based information on the clinical use of neurofeedback for ADHD. *Neurotherapeutics : the journal of the American Society for Experimental NeuroTherapeutics*, 9(3):588–98.

Patki, S., Grundlehner, B., Verwegen, A., Mitra, S., Xu, J., Matsumoto, A., Yazicioglu, R., and Penders, J. (2012). Wireless EEG System with Real Time Impedance Monitoring and Active Electrodes. In *IEEE Biomedical Circuits and Systems Conference*, Hsinchu, Taiwan.

Ruffini, G., Dunne, S., Fuentemilla, L., Grau, C., Farres, E., Marco-Pallares, J., Watts, P., and Silva, R. (2008). First Human Trials of a Dry Electrophysiology Sensor Using Carbon Nanotube Array Interface. *Sensors and Actuators A: Physical*, 144(2):275–279.

Sellers, E., Turner, P., Samacki, W., McManus, T., Vaughan, T., and Mathews, R. (2009). A Novel Dry Electrode for Brain-Computer Interface. *Human Computer Interaction Methods and Techniques Lecture Notes in Computer science*, 5611:623–631.

Teplan, M. . (2002). Fundamentals of EEG Measurement. *Measurement Science Review*, 2:Section 2.

Tăuţan, A.-M., Mihajlović, V., Grundlehner, B., Penders, J., and Serdijn, W. (2013). Framework for Evaluating EEG Signal Quality of Dry Electrode Recordings. In *IEEE Biomedical Circuits and Systems Conference*, Rotterdam, The Netherlands.

Zander, T. O., Lehne, M., Ihme, K., Jatzev, S., Correia, J., Kothe, C., Picht, B., and Nijboer, F. (2011). A dry EEG system for scientific research and brain-computer interfaces. *Frontiers in Neuroscience*, 5.