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New Directions in EEG Measurement: an Investigation into the Fidelity of Electrical Potential Sensor Signals

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Abstract: Low frequency noise performance is the key indicator in determining the signal to noise ratio of a capacitively coupled sensor when used to acquire electroencephalogram signals. For this reason, a prototype Electric Potential Sensor device based on an auto-zero operational amplifier has been developed and evaluated. The absence of 1/f noise in these devices makes them ideal for use with signal frequencies ~10 Hz or less. The active electrodes are designed to be physically and electrically robust and chemically and biochemically inert. They are electrically insulated (anodized) and have diameters of 12 mm or 18 mm. In both cases, the sensors are housed in inert stainless steel machined housings with the electronics fabricated in surface mount components on a printed circuit board compatible with epoxy potting compounds. Potted sensors are designed to be immersed in alcohol for sterilization purposes. A comparative study was conducted with a commercial wet gel electrode system. These studies comprised measurements of both free running electroencephalogram and Event Related Potentials. Quality of the recorded electroencephalogram was assessed using three methods of inspection of raw signal, comparing signal to noise ratios, and Event Related Potentials noise analysis. A strictly comparable signal to noise ratio was observed and the overall conclusion from these comparative studies is that the noise performance of the new sensor is appropriate. *Copyright © 2015 IFSA Publishing, S. L.*

Keywords: Sensors, EEG, Biosensors, Assistive technology, Electrical Potential Sensor.

1. Introduction

The traditional methods employed for the acquisition of electroencephalogram (EEG) signals rely on the use of wet silver/silver chloride (Ag/AgCl) transducing electrodes. These convert ionic current on the surface of the body to electronic current for amplification and subsequent signal processing. Such electrodes are cheap and disposable but require the use of a conducting gel between the electrode and the skin, since they rely on maintaining a low electrical resistance contact [1]. Operationally significant care is required in the preparation of the skin, usually involving abrasion, by skilled personnel. In addition, the gel may cause skin irritation and discomfort as well as drying out after a period of time, meaning that wet electrodes are unsuited to long term monitoring applications [2]. The gel may also be responsible for cross coupling or shorting between electrodes in an array if great care is not taken during placement. Dry conducting electrodes provide a more user-friendly approach with electrodes making only resistive contact with the skin [3]. This overcomes the problems caused by the wet electrode gel, but introduces an additional variable, namely the variation in contact resistance due to perspiration, skin creams, or other individual differences in physiology. For these reasons, they tend to be noisier than wet electrodes. Dry electrodes can also suffer more from movement artefacts if they are not securely fastened.

An alternative approach is to dispense with the resistive contact and couple capacitively through an insulating layer [4]. With this method the signal fidelity no longer relies on skin resistance, however they can also suffer from movement artefacts and charge sensitivity. In most embodiments of dry and insulated electrodes an active electrode structure is used with high impedance amplification [4-5]. This minimizes the noise due to cabling and transmission of the signal. Electric Potential Sensor (EPS) is a high performance version of the insulated active sensor.

With specific reference to EEG signal acquisition, evidence exists that smaller, lighter sensors with a higher array density are required in order to reduce movement artefacts and to allow for redundancy [6]. A comprehensive review of wet, dry and insulating electrode technologies concludes that insulated active electrodes offer the most promising solution for future healthcare applications [1]. More recent work on dry electrodes has included a trial of a 6 sensor EEG system [7] and concludes that this could offer a cost effective solution for brain-computer interfacing. A clinical comparison of concurrent measurements with wet and dry EEG electrodes concludes that a high degree of correlation is seen and that dry electrodes offer better long-term performance [8]. New work on motion artefact reduction relies on the simultaneous measurement of the contact impedance of each sensor [9] using a small a.c. current (20 nA @ 1 kHz) and multiple dry spring loaded contacts in each sensor to introduce redundancy. Other workers have designed quasi-dry polymer electrodes which use a small quantity of moisturizing agent to address these problems [10].

In summary, EPS technology has already demonstrated that problems such as offset potentials, signal drift, ease of usability, and invasiveness can be addressed for electrocardiogram (ECG) data acquisition where the inherent DC stability and short settling time of the sensors differentiate them from other insulated electrode implementations [11]. However, the low frequency noise performance required for accurate EEG data acquisition is considerably more stringent and it is this important parameter which will be addressed in this paper. A review of sensor developments for healthcare [11] discusses the low frequency noise performance of a number of active sensors and characterizes them in terms of the noise spectral density at 1 Hz. This is a useful indicator of the performance for EEG use and gives values ranging from 2 μV/√Hz to 10 μV/√Hz, however these values will increase at lower frequencies due to 1/f noise.

The aim of the present paper is to examine whether the signal-to-noise performance of the EPS system is comparable to – or better than – that of a conventional electrode system, specifically in the 0.1-10 Hz bandwidth.

The design and specifications of the EPS sensor used in these experiments are described in Section 2, along with details of the commercial system used for comparing EPS with gel electrodes. In Section 3 the EPS results for free running (spontaneous) EEG are described, followed by data for two event-related potential (ERP) studies in Section 4. The second ERP experiment outlines a comparative study conducted between the two systems. Section 5 discusses three methods for quantifying noise of an EEG recording device, relevant to our EPS system.

2. Prototype Sensor and Systems

The prototype Sussex EPS device for this project is based on an auto-zero operational amplifier, chosen to give the lowest possible low frequency noise [12]. The absence of 1/f noise in these devices makes them ideal for use with signal frequencies ~10 Hz or less, with a quoted noise performance of 22 nV/√Hz and 5 fA/√Hz. The input capacitance is ~8 pF with an associated voltage noise between 0.1-10 Hz of 0.5 μVp-p. After consideration of the expected signal amplitudes and frequency the sensor was configured to have an operational bandwidth of 0.1 Hz to 78 Hz and a voltage gain of x50. The voltage gain was distributed between two stages with ×5 and ×10 respectively for the first and second stages. The operation and circuit details of EPS devices have been published previously by the authors [13]. Here, the sensors are operated from split symmetric power supply rails of ±2.5 V. Two versions were produced with different electrode sizes to enable reliable contact to be made to different parts of the body. The electrodes are electrically insulated through an anodized electrode with diameters of either 12 mm or 18 mm. In both cases the sensors were housed in inert stainless steel machined housings with the electronics fabricated in surface mount on a printed circuit board (PCB) compatible with epoxy potting compounds. Potted sensors are designed to be immersed in alcohol for sterilization purposes.

The gain and operational bandwidth of the sensors was confirmed using a standard spectrum analyzer to be as specified. The most significant parameter for the specification of the sensor in this particular application is the voltage noise referred to the input. This was measured by placing the sensor in a screened environment and recording the spectral noise density over a 1 kHz bandwidth. From this data, shown in Fig. 1, two numbers are produced to characterize the noise performance: the spot noise figure at 1 Hz and the integrated noise from 0.1 Hz to 10 Hz. The results obtained for the voltage noise measurements are: 30 nV/√Hz at 1 Hz and 0.2 µVp-p from 0.1 to 10 Hz; consistent with the data provided by the manufacturer. The absence of 1/f noise in this data confirms that the auto-zero amplifier used in this design is performing as expected.

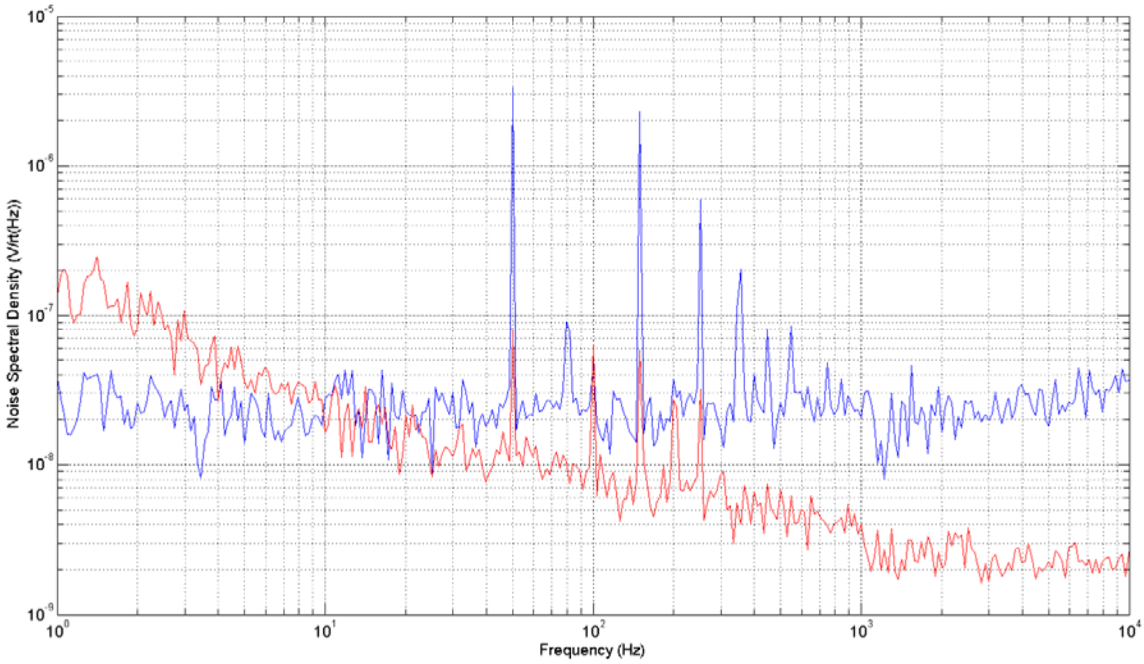

Fig. 1. Noise spectral density plot for prototype auto-zero sensor in comparison with a JFET input stage amplifier. The blue trace represents the prototype sensor. A lack of typical 1/f operational amplifier noise can be observed. Although the red trace has lower noise for frequencies higher than 20 Hz, the prototype sensor has lower noise in 1 to 10 Hz region (which is where the signal of interest lays). The voltage noise referred to input is measured as 30 nV/√Hz at 1 Hz.

In order to confirm, at an early stage in the design process, that the sensor design was both suitable for high quality EEG signal acquisition and that it was compatible with commercial systems and practice we interfaced the sensors to a TMS International (TMSi) system currently in use in the School of Psychology at Sussex. This also enabled us to perform direct comparisons with wet gel electrode measurements. The prototype sensors were interfaced to a Refa8 amplifier produced by TMS International [14] with 64 EEG channels at 24 bit resolution with an input noise of 1 µV_{rms}. All electrode cables have active shielding to reduce 50 Hz mains interference and cable movement artefacts. In the comparative data presented here the TMS International acquisition system and data processing were applied to both sets of data.

In order to provide a comprehensive comparison between the Sussex EPS prototype and the commercial system two different types of EEG data were measured. First we recorded free running EEG focusing on the (well known) 'alpha blocking' signal.

The second type of EEG data recorded were event related potentials (ERPs). Two experiments were run to elicit ERPs: The first was a simple visual “oddball paradigm” [15], and the second a simple visual face perception [16].

In the case of both the EPS and commercial EEG systems, reference electrodes/sensors were placed on left and right mastoid positions. Other sensors/electrodes were placed at scalp positions dictated by the specific experimental paradigm, according to the International Standard 10-20 electrode location system [17]: left and right occipital (O1 and O2) for alpha-blocking; midline parietal (Pz) for the visual oddball paradigm; and left/right parietal (P7 and P8) for the face processing paradigm.

The recorded data from the paradigm-specific sensors/electrodes were offline re-referenced to linked mastoids, by averaging data from the two mastoid positions and subtracting it from each paradigm-specific sensor/electrode.

3. Spontaneous EEG

Initial measurements were carried out on the free running EEG to verify that the prototype sensor had an appropriate noise performance to allow EEG data to be seen. The alpha signal is observed when the eyes are closed and is characterized by an increase in amplitude of the 8-13 Hz EEG signal. Alpha activity can be recorded from 95 % of people [15] and is blocked when the eyes are open. The signal may be seen in real time in the time domain, as shown in Fig. 2, where the alpha blocking caused by opening the eyes may be seen clearly.

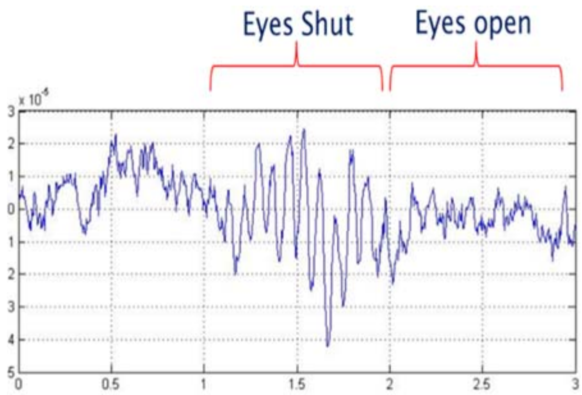

Fig. 2. Time domain data showing the alpha blocking phenomena measured using the prototype sensor.

Alternatively, if the time series data is Fourier transformed we see a broad peak in the frequency domain data. This is illustrated in Fig. 3 where a 40 s section of time series alpha data has been Fourier transformed to show a clear ~10 Hz peak.

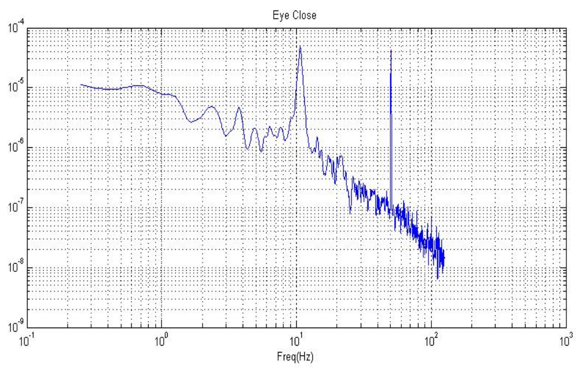

Fig. 3. Fourier transform of time domain data showing a broad alpha signal peak at ~10 Hz. The participant was asked to close their eyes for period of 40 s. The second peak is the 50 Hz mains signal.

A residual 50 Hz mains interference signal may also be seen (the rightmost peak in Fig. 3), however the common mode rejection ratio (CMRR) is sufficient to reduce this amplitude to be comparable to the measured signal.

4. Event Related Potentials

Event-related potentials (ERPs) are time-locked EEG responses to specific events, usually discrete sensory inputs. ERPs are characterized by deflections from baseline (baselines are typically computed over the pre-stimulus period) at specific time-points post-stimulus. The 'oddball effect' is the ERP difference reflecting a contrast between expected and unexpected (frequent and infrequent) stimuli.

Two different stimuli are presented on a screen with one event randomly chosen to occur more often than the other. A volunteer is asked to press the space bar only when they are presented with one of the two events. Typically, signals are averaged and band-pass filtered at 0.1 to 30 Hz, again we have followed this standard practice. A typical oddball paradigm presents letters e.g. X and O on a monitor with 80 % and 20 % relative frequency respectively [15]. The letters are displayed for 100 ms with a blank screen presented for 1.4 s between each letter. In this experiment, the O is the 'oddball' stimulus.

Little information may be gained from real-time data in ERP paradigms, so that averaging over a number of events is usually needed [18]. The data is usually recorded using the Pz position and a reference electrode (s). Fig. 4 shows the results for 67 averages, 53 for the 'X' and 14 for the 'O'. There is a clear time-difference in the major ERP component between the 'X' and 'O' data as expected [19]. This experiment contained a relatively low number of trials for an ERP study, indicating that the EPS prototype sensor is highly capable in this challenging mode of operation.

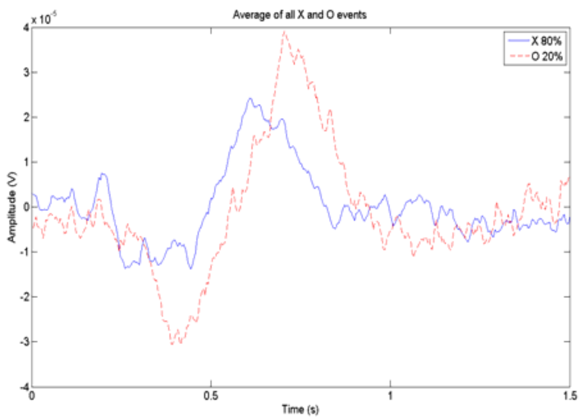

Fig. 4. Averaged ERP data, recorded from Pz, collected by the EPS, showing the oddball effect (delayed response to the rare stimulus).

In a second ERP study we looked at the sensitivity of the EPS for recording category specific effects in a standard face processing study. Three different images were presented to the subjects: faces, inverted faces and scrambled faces. The resulting ERP waveforms are displayed for the wet gel electrodes in Fig. 5 and the EPS in Fig. 6. For both sensor types measurement electrodes were located at the P7 and P8 positions, with the reference sensors on M1 and M2 (left and right mastoids).

In order to improve the quality of the data and to allow a more accurate comparison to be conducted a grand average was produced over 4 subjects.

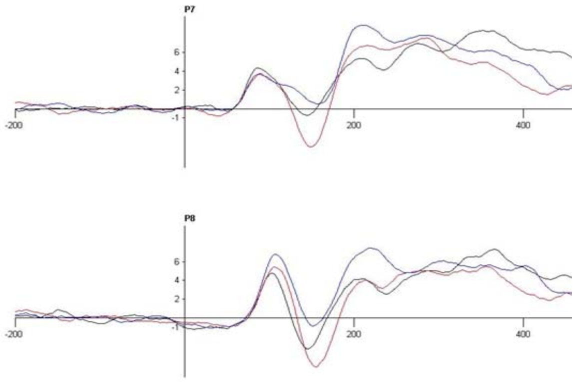

Fig. 5. Grand average of 4 participants ERP face data from wet gel electrodes, positions P7 and P8. Faces (black line); inverted faces (red line) and scrambled faces (blue line). Stimulus onset at 0 ms.

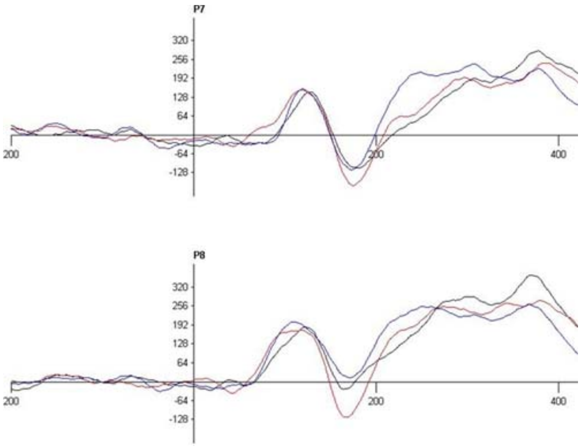

Fig. 6. Grand average of 4 participants ERP face data from EPS, positions P7 (top) and P8 (bottom). Faces (black line); inverted faces (red line) and scrambled faces (blue line). Stimulus onset at 0 ms.

Visual inspection of the results of these ERP measurements show that the grand averaged data for the EPS system is remarkably similar to that produced by a standard commercial EEG system, specifically in terms of demonstrating stimulus-category specific patterns of activity. Moreover, the apparent signal to noise ratio appears to be strictly comparable. From these initial results we therefore conclude that the current prototype EPS device has an adequate level of noise performance for all the EEG signals observed during these tests.

5. EEG Quality

Quantification of noise in an EEG recording is complex due to the signal processing that is applied to the recorded signals. However, if different aspects of an EEG signal are studied individually then useful information can be gained. Here three methods are used to separately characterize the low frequency drift, ERP noise, and broadband noise.

5.1. Raw Signal

EEG data is commonly preprocessed with high and low pass filters to remove any drift caused by wet gel electrodes (and other factors) and to reduce the effects of out-of-band noise. However if the raw signal is inspected for a long period of time then slow drift can be observed. These drifts are due to a combination of different effects such as variations in skin resistance [20] and alteration in half-cell potential of Ag/AgCl electrodes [3].

Fig. 7 displays 10 minutes of EEG recording from both the EPS and wet gel systems. Here the raw signal can be seen with a drift in the wet gel electrode of up to 2 mV. This is often compensated for by using high supply rails to avoid railing the signal, and high precision 24 bit digitizers to be able to record these small signals over larger ranges. This effect is not seen in the EPS as the sensor does not have a DC response.

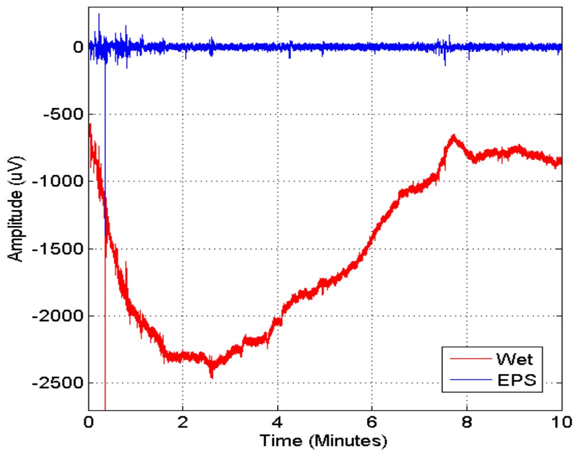

Fig. 7. The raw signal can be seen to drift with time when recorded with a wet gel electrode, this is not the case when compared with the EPS drift over a 10 minute recording.

5.2. Signal to Noise Ratio

As demonstrated in Section 3, when a person closes their eyes a ~10 Hz oscillation appears in their EEG. This power increase in this alpha band of EEG is reversed upon opening the eyes ('alpha blocking'). As Alpha is a signal superimposed on the background EEG and broadband noise, it is possible to define this evoked increase in the Alpha band power by comparing it to the background variations. Signal to noise ratio (SNR) can be calculated using the following equation [20].

$$SNR = 10\log_{10} \frac{EEG(10Hz)^2}{\mathrm{var}(EEG) - EEG(10Hz)^2} \tag{1}$$

A 30 second recording of an Alpha signal from a single participant was gathered simultaneously by the prototype EPS and the TMSi system. This was formed by a differential recording of Oz-Fz. Using FFT, power of the alpha signal at 10 Hz was calculated across this data. The EPS had a SNR of -26.0244 dB compared to a value of -33.2565 dB for the wet gel electrode. This provides values smaller than zero due to the small amplitude of the Alpha signal compared with broadband noise and background EEG. Showing that for this recording the EPS displays a slightly better SNR than a standard EEG system.

5. 3. ERP Noise

Event related potentials are the most commonly studied signal type in EEG. Thus it is important to assess the SNR of an averaged ERP. Fig. 6 and Fig. 7 both present grand average ERPs from the face perception study. The pre-stimulus section (-200 ms to 0 s) of both figures shows no significant variations from the baseline. Assuming no stimulus-specific neuroelectrical activity prior to the trigger signal then any deviations can be associated with noise. The Root-mean-square (RMS) voltage of this time period presents a measure of noise activity independent of frequency [20]. The closer the RMS is to zero the less affected the ERP signal is to non-event related activity. Table 1 contains RMS noise values of 4 subjects who participated in the face perception study.

Table 1. ERP RMS Noise.

Subject	**Pre stimulus ERP RMS Voltage (μV)**	
No.	Wet	EPS
1.	1.0503	0.2574
2.	0.8073	0.6030
3.	0.8992	0.1642
4.	0.2702	0.3711
Mean	0.7568	0.3489
STD	0.3395	0.1893

These RMS values show that the EPS displays a similar noise profile to standard wet electrodes. The variations in mean and SD was expected as these signals were not recorded simultaneously.

6. Conclusions

The Sussex EPS prototype has been verified as suitable for the acquisition of both free running EEG and ERPs. The prototype performance has also been verified by interfacing with a commercial system, and by comparing results with those from wet gel electrodes. All results obtained indicate that the Sussex EPS prototype produces strictly comparable signal-to-noise ratios to conventional wet gel electrode devices for both free running and ERP measurements. The low frequency noise has been identified as the key performance indicator for capacitively coupled active sensors. In particular, the frequency range of typical EEG signals lies within the 1/f noise region of most active devices. The use of an auto-zero operational amplifier within the prototype sensor has been demonstrated to eliminate this problem and yield results which are strictly comparable to wet gel electrodes.

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