# Comparison of Dry and Wet Electrode Systems for Spontaneous and Event Related Electroencephalograms

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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 (EEG) signals. For this reason, a prototype Electric Potential Sensor (EPS) 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 (PCB) 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 EEG and Event Related Potentials (ERP). 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.**

***Keywords-Sensors; EEG; Biosensors; Assistive technology.***

## I. INTRODUCTION

The traditional methods employed for the acquisition of 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 abraision, 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 resistive contact with the skin [3]. This overcomes the problems caused by the wet electrode gel, but introduces an additional variable, the variation in contact resistance due to perspiration or skin creams. For these reasons, they tend to be noisier than wet electrodes and can suffer 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. 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 these problems 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 this work is to produce a high impedance capacitively coupled sensor with noise that is comparable or lower than conventional electrode systems in a 0.1-10 Hz bandwidth.

The design and specifications of the EPS sensor used in these experiments are described in section II, along with details of the commercial system used for comparing EPS with gel electrodes. In section III the results for free running EEG is demonstrated followed by data for two ERP studies in section IV. The second ERP experiment outlines the comparative study conducted between the two systems.

## II. PROTOTYPE SENSOR AND SYSTEM

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 x5 and x10 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 ±2.5V. 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 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 Figure 1, two numbers are produced to characterize the noise performance these are 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.

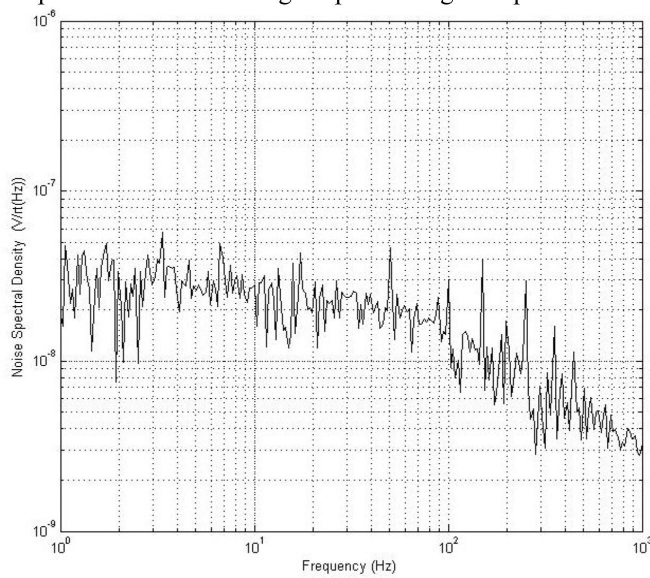


Figure 1. Noise spectral density plot for prototype auto-zero sensor.

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 system currently in use in the School of Psychology at Sussex. This also enabled us to do 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 $\mu V_{rms}$. 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. The spontaneous, or free running, EEG as shown by recording the alpha signal and in particular by observing the, well known, alpha blocking signal. The second type is the ERP, here an oddball paradigm was chosen to record ERP signals [15]. In both cases the International standard 10-20 system was employed to record EEG [16]. In our case all signals were recorded against a reference, this may be the signal from any electrode on the head or an average of all electrodes. Usually a linked mastoid reference is used and we followed this practice with the signals from the O2 and O1 positions and the reference sensor on M1 (left mastoid).

## III. Free Running 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

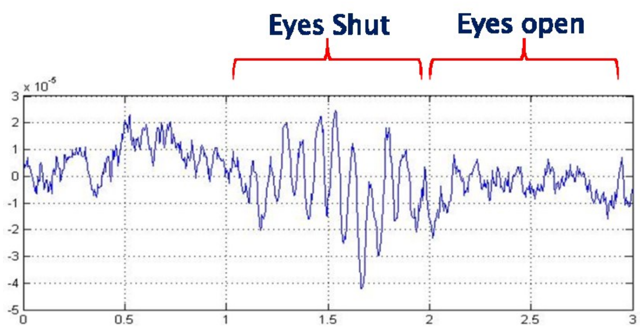


Figure 2. Time domain data showing the alpha blocking phenomena measured using the prototype sensor.

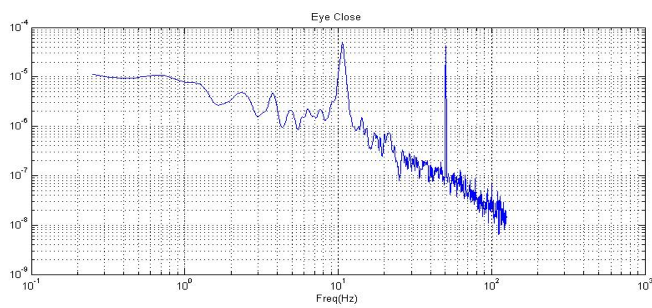


Figure 3. Fourier transform of 40 s of time domain data showing a broad alpha signal peak at ~10 Hz.

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 Figure 2, where the alpha blocking caused by opening the eyes may be seen clearly. Alternatively, if the time series data is Fourier transformed we see a broad peak in the frequency domain data. This is illustrated in Figure 3 where a 40 s section of time series alpha data has been Fourier transformed to show a clear ~10 Hz peak. A residual 50 Hz mains interference signal may also be seen, however the common mode rejection ratio (CMRR) is sufficient to reduce this amplitude to be comparable to the measured signal.

## IV. Event Related Potential

The oddball effect is a measure of the response of the brain to the frequency of occurrence of differing events on a screen. The ERP resulting from this shows a time shift which arises from this difference. Two different events 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 setup is where the letters X and O are displayed on a monitor with 80% and 20% relative frequency respectively. The letters are displayed for 100 ms with a blank screen presented for 1.4 s space between each letter.

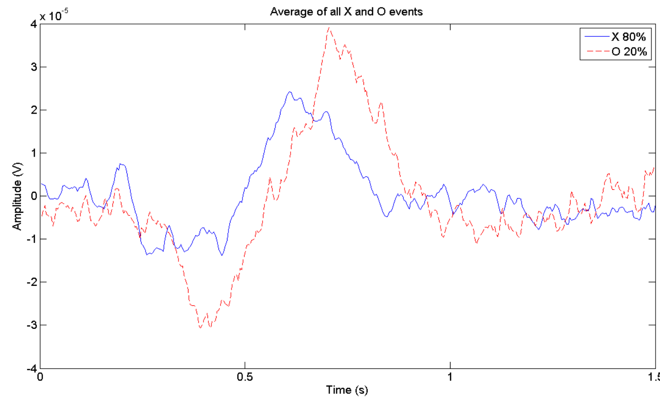


Figure 4. Averaged prototype sensor ERP data showing the expected time delay associated with the oddball effect.

Little or no useful information may be gained from real time data and averaging over a number of events is usual. The data is usually recorded using the Pz position and a reference electrode. Figure 4 shows the results for 67 averages, 53 for the 'X' and 14 for the 'O'. There is a clear time difference between the 'X' and 'O' data as expected. This is a relatively low number of averages for this type of measurement and indicates that the EPS prototype sensor is very capable in this challenging mode of operation. For ERP measurements direct comparisons between the wet gel and EPS systems are shown in Figures 5 and 6 with measurement electrodes located at the P7 and P8 positions. 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.

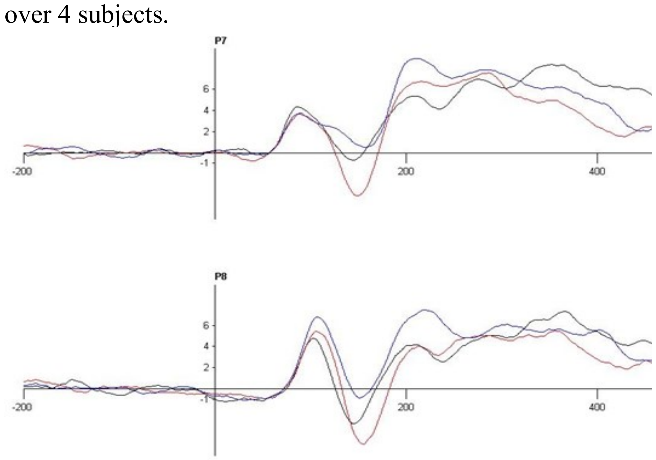


Figure 5. Grand average of ERP data from TMS system.

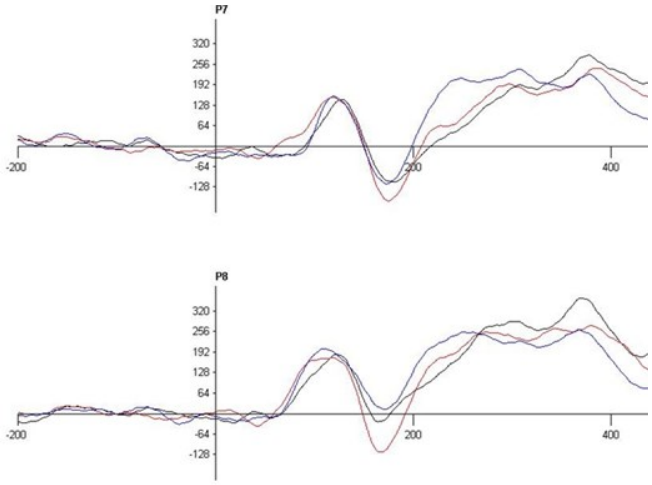


Figure 6. Grand average of ERP data from prototype EPS.

Three different images were presented to the subjects; faces (black line); inverted faces (red line) and scrambled faces (blue line). The overall conclusion from these ERP measurements is that the grand average data sets show a high degree of correlation between the two systems with the apparent signal to noise ratio looking 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.

## V. 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 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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