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# A dry electrode for EEG recording

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**Summary** This paper describes the design, fabrication and testing of a prototype dry surface electrode for EEG signal recording. The new dry electrode has the advantages of no need for skin preparation or conductive paste, potential for reduced sensitivity to motion artifacts and an enhanced signal-to-noise ratio. The electrode's sensing element is a 3 mm stainless steel disk which has a 2000 Å (200 nm) thick nitride coating deposited onto one side. The back side of the disk is attached to an impedance converting amplifier. The prototype electrode was mounted on a copper plate attached to the scalp by a Velcro strap.

The performance of this prototype dry electrode was compared to commercially available wet electrodes in 3 areas of electroencephalogram (EEG) recording: (1) spontaneous EEG, (2) sensory evoked potentials, and (3) cognitive evoked potentials. In addition to the raw EEG, the power spectra of the signals from both types of electrodes were also recorded. The results suggest that the dry electrode performs comparably to conventional electrodes for all types of EEG signal analysis. This new electrode may be useful for the production of high resolution surface maps of brain activity where a large number of electrodes or prolonged recording times are required.

**Key words:** Dry EEG electrode; Insulated EEG electrode

Applications for surface bio-electrodes encompass many fields of medicine and engineering. As diagnostic tools, electrodes are routinely used to monitor and analyze heart conditions, brain activity, and other biological phenomena (Portnoy et al. 1974; Kavanagh and Terrance 1978; Wolpaw and McFarland 1991). Wet scalp electrodes for EEG applications are limited by their size, the requisite skin preparation time, and sensitivity to noise. For example, EEG studies can require attachment of 64–128 electrodes to the scalp of the subject, necessitating small electrode-to-electrode spacing. Commercially available wet electrodes often short out or smear with such small spacing due to electrolyte overflow and size. Scalp preparation and attachment of each wet electrode also take a considerable amount of time.

Signal amplitudes from the brain recorded on the scalp vary from 0.25 to 20 $\mu$V for evoked potentials (EPs) to hundreds of microvolts for spontaneous EEG. Associated bandwidths range from 0.01 Hz to 5 kHz depending upon the application. Noise is the main factor limiting the resolution of wet surface electrodes. There are several important noise sources to consider. Electromagnetic interference is due to fields generated by electrical current passing through wires attached to lights, computers, instruments, and display monitors near the subject. Triboelectrical noise is generated in wires by friction between the signal carrying conductor and the insulating material around the conductor. Skin potentials are generated by ionic charge separation in different layers of the skin. The ionic charge separation acts like charge storage capacitors or batteries. Movements by the patient induce artifacts at the electrode and skin interface due to the slippage of the electrode over the skin.

Application of standard wet electrodes in noisy environments is limited predominantly by the dependency on skin impedance (Huhta and Webster 1974; Tam and Webster 1977). Most surface electrodes utilize a conductive paste and abrasive skin preparation to reduce the skin impedance. The outer most barrier layer of the skin is the major source of electrode to skin impedance and can contribute up to 5 mV of noise due to motion. Reduction in electromagnetic interference is directly linked to the reduction of skin impedance at the electrode-to-skin interface (Tam and Webster 1977). Removing the outer skin layer lowers the interface impedance and decreases motion artifact noise. The skin-to-electrode interface should have an impedance below 5 k$\Omega$ for the wet electrode to operate optimally (Swanson and Webster 1974; Travis 1974). However, the combination of electrolyte drying and skin removal causes skin irritation and degrades signal

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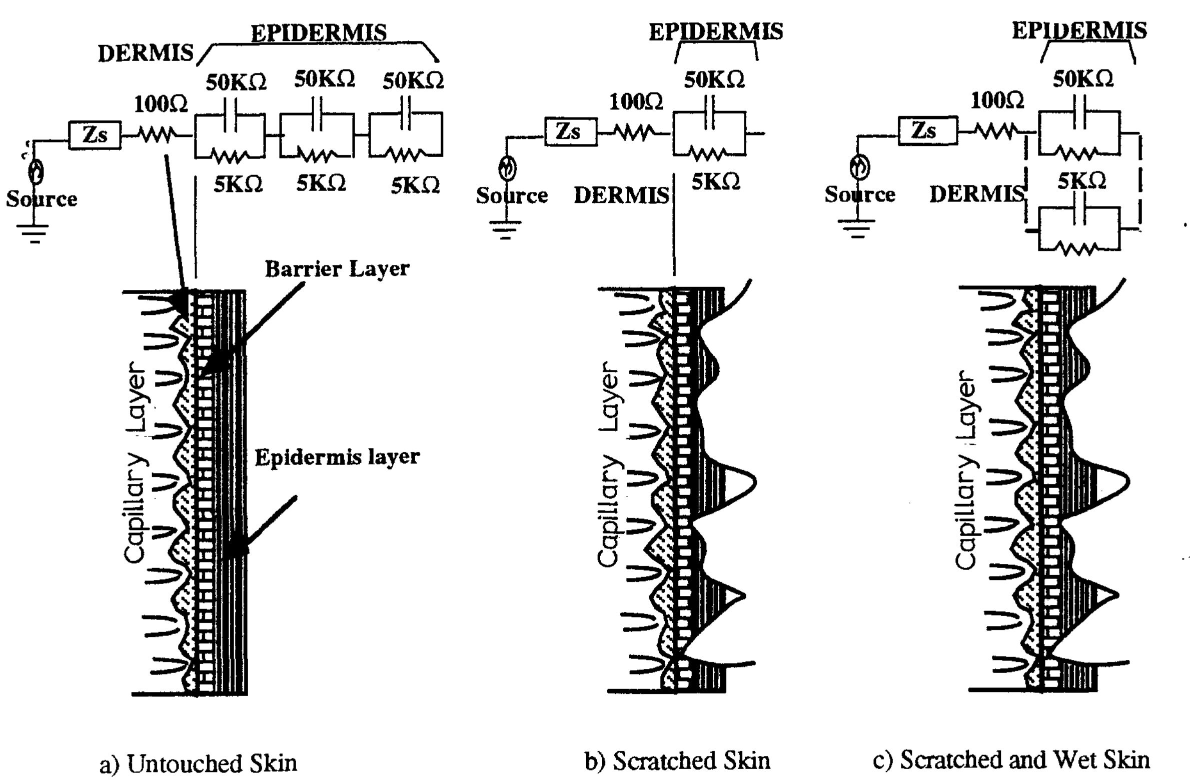
TABLE I

Overview of the historical development of dry electrodes. These electrodes are classified based on their type, the material used for the substrate to hold the electrode, the material used to attach the electrode to the substrate, electronic technology used, number of active sites per electrode, and the application area. MOS op-amp stands for metal oxide semiconductor operational amplifier. CMOS stands for complementary metal oxide semiconductors. BIFET stands for bipolar field effect transistor. $Si_3N_4$ is silicon nitride.

| Year | Investigator | Type | Substrate | Substrate attachment | Electronic technology | No. of sites | Application areas |
|---|---|---|---|---|---|---|---|
| 1910 | Waller | Solution | – | – | – | 1 | ECG |
| 1966 | Geddes and Baker | Ag/AgCl | Ag/AgCl | Bond | – | 1 | ECG |
| 1967 | Richardson | Insulated | – | Adhesive | – | 1 | ECG |
| 1970 | Wise and Angel | Gold | Silicon | – | – | 1 | Cellular EEG |
| 1974 | Portnoy et al. | Insulated | Silicon and Al | Adhesive | MOS op-amp | 1 | ECG |
| 1974 | Ko and Hynecek | Dry | Steel and Si | Adhesive | BIFET process | 1 | ECG |
| 1990 | Padmadinata | Dry | Steel and Si | Adhesive | Bipolar process | 1 | ECG |

levels during long-term recording. Several studies conducted by NASA showed that skin regrowth during long-term recording (i.e., days) degrades the performance of wet electrodes (Richardson 1967, 1968b; Ko and Hynecek 1974). Potential applications requiring long recording times from hours to weeks include eye tracking for surgery (Hillyard and Picton 1987), cursor control using mu rhythm (Wolpaw and McFarland 1991), prosthetic control, and epilepsy monitoring.

Several investigators have discussed ways of improving surface electrode technology for electrocardiographic (ECG) applications (Richardson 1967; Ko and Hynecek 1974; Swanson and Webster 1974). Dry surface electrodes have often been considered as replacements for the wet version due to elimination of procedures for skin removal and electrolyte application. Dry electrodes have the disadvantage of substantially higher electrode-to-skin impedance. For dry electrodes to be




a) Untouched Skin b) Scratched Skin c) Scratched and Wet Skin

Fig. 1. Skin model with typical impedance values at 10 Hz for a 1 $cm^2$ Ag-Agcl electrode used with Burdick paste (Padmadinata 1990). Zs is a variable source impedance that includes the impedance of the scalp and other tissues below the dermis to where the source of potentials are located. a: epidermis layer has both capacitance and resistance, and at least 3 orders of magnitudes higher impedance than the dermis layer. b: epidermis layer is partially removed, reducing its impedance. c: moisture lowers the epidermis impedance. The combination of partial removal of the epidermis and contact with moisture results in lowest overall skin impedance.

useful, this high impedance must be converted locally to a low impedance to minimize the effects of the various noise sources. The most suitable way to lower the interface impedance is via a local impedance converting amplifier that also boosts the signal for transmission over the cable. The advantages of previously reported dry electrodes are: (1) no skin preparation, (2) no need for conductive paste, and (3) reduction of noise due to cable triboelectrical effects and electromagnetic interference. The disadvantages are: (1) bulky size due to additional electronics and limitations of power sources, (2) noise due to limitations of the electronics available in the 1970s, (3) motion artifacts due to poor skin-to-electrode contact, and (4) the higher cost.

Electrodes that employ impedance transformation at the sensing site via active electronic devices and/or circuits are referred to as active electrodes. In the engineering literature active electrodes are subdivided into "dry electrodes" and "insulated electrodes" (Richardson 1968a). The dry electrode has metal in direct contact with the skin, while the insulated electrode is capacitively coupled to the skin. The insulated electrodes use a dielectric material (not electrically conductive) that is in contact with the skin instead of the metallic contact for the dry electrodes. The electrode in this manuscript utilizes a dielectric material for contact and is technically an insulated electrode. However, in the biomedical literature electrodes are often referred to as either dry or wet. For consistency with the biomedical literature, we use the term dry electrode for the active electrode described in this manuscript.

Research on active electrodes was initiated in the late 1960s and ended in the early 1970s with the exception of one report in 1990 by Padmadinata. The results by Ko and Richardson demonstrated that both dry and insulated active electrodes can be used to pick-up ECG signals with good signal characteristics in comparison to wet electrodes (Richardson 1967, 1968a,b; Ko and Hynecek 1974; Portnoy et al. 1974). Table I summarizes known published reports on dry electrodes, citing differences in the techniques and technologies.

The literature on active electrodes has focused on ECG recordings. No reports of EEG studies with active electrodes were found. In addition, no previous work on active electrodes appears to have been done for low-level signal recording applications (below 100 $\mu$V), which is critical for EEG recording.

## Skin-electrode interface

Skin is the exterior interface between the electrical fields induced by neurons and the surface of the electrode. The interface between the skin and an electrode has been extensively studied and modeled since the skin characteristics are critical in determining the frequency response and noise susceptibility at the interface (Swanson and Webster 1974; Travis 1974). Fig. 1 shows the skin layers under 3 different conditions along with their modeled impedance values extracted from the literature (Swanson and Webster 1974; Travis 1974). There are several characteristics of the skin that affect the potential sensed by an electrode. First, the barrier layer between the epidermis layer and the dermis layer gives rise to skin potentials of typically 30 mV. This is proposed to be due to ionic charge separation between the epidermal and dermal layers (Travis 1974).

When skin stretches, the barrier layer potential decreases to 25 mV and this potential change is perceived as a motion artifact. Tam and Webster (1977) showed that this pressure-induced skin potential change can be virtually eliminated by 20 strokes of the skin with fine sandpaper. The scratches reduce the impedance by eliminating the series coupled electrical elements in the epidermis layer as shown in the skin model of Fig. 1b. The scratches do not need to go as deep as the capillary loops, so the procedure draws no blood. However, the barrier layer also serves to protect the skin from irritating substances such as electrode gel. The active dry electrode approach we present here eliminates the need for abrasive skin preparations whereas the passive wet electrode requires them.

Temperature, humidity and changes due to sweat glands also affect skin impedance. With increasing humidity skin resistance decreases and skin capacitance increases. An effective model for this phenomenon is to place another resistor-capacitor pair in parallel with each of the elements in the epidermis layer as shown in Fig. 1c. In the design of skin electrodes, the effects of changing skin impedance can be minimized by introducing an electrode dominating capacitance that is in series with the skin impedance. This capacitor is shown in Fig. 2. The electrode capacitance should be made dominant since its value is constant in contrast to the skin capacitance.

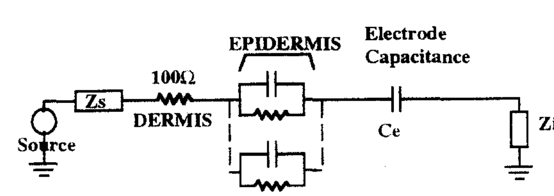



Fig. 2. Skin and electrode model with amplifier input impedance, Zin. The electrode capacitance value is much smaller than the capacitance in the epidermis layer capacitance. This model is used to simulate the characteristics of the skin in the frequency range of 0 Hz to 1 MHz.

The portion of electrode impedance due to capacitance at the highest frequencies recorded (10 kHz) must be smaller than the skin impedance. This is due to the fact that the skin impedance is in series with the electrode capacitance. To calculate the desired electrode capacitance, skin impedances were obtained from the literature (Swanson and Webster 19974). The skin impedance at 10 kHz is approximately 1 k$\Omega$ (Rosell et al. 1988) which results in a capacitance value of 15 nF (nano = $10^{-9}$). The value of skin capacitance sets the upper limit for the series combination of electrode and skin capacitance. The electrode capacitance should be at least two orders of magnitude smaller than 15 nF at 10 kHz to become the dominant frequency setting component in the system. Thus, values of the electrode capacitance between 150 pF and 300 pF (pico = $10^{-12}$) are adequate to dominate the frequency characteristics of the electrode. This represents a significant departure from previous active electrode designs where the skin capacitance was dominant. This approach should significantly improve the performance of the dry electrode.

Introduction of a dominating electrode capacitance in series with the skin would effect the low frequency response of the electrode if there were no high input impedance amplifier following the capacitor. This is not an issue in our electrode design. Since the electrode capacitance and the input impedance of the amplifier form a high pass filter, one must use an amplifier with high input impedance so that the lower frequency limit of 0.01 Hz that is a requirement for measuring low frequency EPs is not compromised. Since capacitance values of the order of 150–300 Pf are recommended, the amplifier following the capacitor should have input impedance of greater than $6.7 \times 10^{14}$ $\Omega$ to $3.3 \times 10^{14}$ $\Omega$. The amplifier employed in our electrode had an input impedance of $10^{15}$ $\Omega$ which is twice the needed impedance value.

Fig. 2 shows a first-order model that does not include the parasitic capacitances and resistances of the electrode. Parasitic or stray components are the unwanted elements in a circuit that form between the substrate of the sensor and the sensing site. These stray values need to be incorporated into the model if the sensor site area is smaller than 100 $\mu m^2$ with dielectric material made of silicon nitride. Since our prototype electrode is greater than 3 mm$^2$, parasitic components do not need to be incorporated into the model.

## Methods and materials

*Sensor construction*

The electrode was built using a 3 mm stainless-steel disk with a layer of 200 nm thick silicon nitride ($Si_3N_4$) deposited onto the front side. The back side of the disk was connected to a wire by a conductive silver adhesive paste to assure low resistance. The disk was mounted on a copper plate with rubber stands to electrically isolate the disk from the plate. The signal wire was passed through a hole in the rubber stand and the copper plate to the input of a buffer amplifier (ICL7621 made by Maxim). The analog output of the amplifier was fed to standard EEG instruments with an additional ground wire connected to an isolated ground. All the components on the plate surface were shielded by a copper box. An additional rubber stand was necessary to balance the copper plate and assure that the sensor surface was in full contact with the skin. The total dimension of this macro sensor was 3″ × 1″ × 2″ (length, width, thickness). A drawing of the dry electrode prototype is shown in Fig. 3.

An example of electrode placement and attachment is shown in Fig. 4. Experiments were conducted to verify the transduction mechanism of the prototype dry

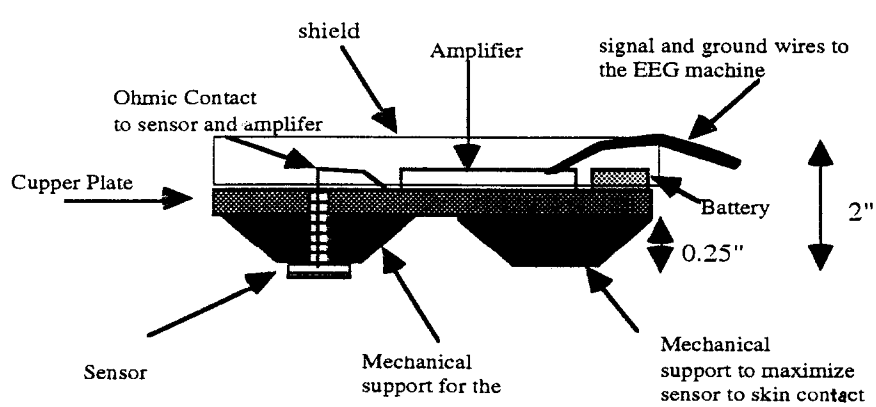



Fig. 3. Experimental dry electrode built with copper plate and packaged amplifiers. The drawing is not shown to scale. All the components including the battery for powering the amplifiers are packaged in a shielded copper box.

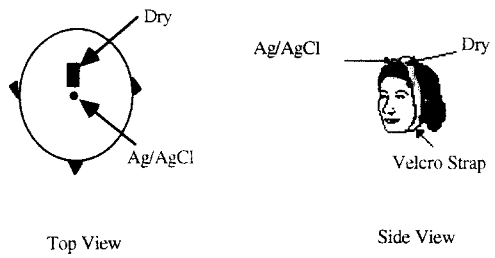



Fig. 4. Electrode placement on scalp and method of attachment. The output of each electrode was directly connected to commercial EEG equipment. The dry electrode was attached using a Velcro strap. The reference and Ag/AgCl were attached by conductive paste.

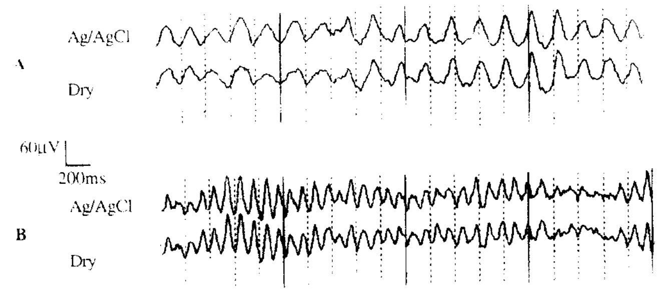



Fig. 5. Simultaneously recorded spontaneous EEG signals from the wet and dry electrodes. The top traces (A) show theta activity from the drowsy state. The bottom set of traces (B) show alpha activity in the awake state.

electrode. The dry electrode was mounted on the scalp without any skin preparation or electrolyte application. This was accomplished using a Velcro strap that held the electrode and associated electronics in place on the scalp. The attachment time for the electrode was less than 15 sec. The reference and Ag/AgCl electrodes were attached with conventional conductive paste.

## Results

Spontaneous and evoked potentials were recorded to compare the performance of the standard wet electrodes (Ag-AgCl) with the dry electrode. In addition to spontaneous EEG potentials we measured two types of evoked potentials including: (a) stimulus-dependent visual evoked potentials to checkerboard stimulation, and (b) endogenous cognitive evoked potentials generated in a memory paradigm. The dry electrode output is fully compatible with standard clinical EEG machines and amplifiers. The data presented in the manuscript were collected using Grass P511 amplifiers (spontaneous EEG and auditory long latencies) and a Cadwell system (VEPs).

### *(a) Spontaneous EEG*

The first study measured spontaneous EEG from both types of electrodes in one subject. The bandwidth of the measurement was from 0.5 Hz to 70 Hz, with amplitude sensitivity of 10 $\mu$V/mm. The output signals are shown in Fig. 5. EEG and spectral analysis were repeated on several occasions over a 15 min epoch to assure reliability. Theta and alpha rhythms compared for both electrodes. The upper traces of each pair are the output of the Ag-AgCl electrode. The bottom traces of each pair are the output from the dry electrode. Note that the raw signal amplitudes of the conventional electrode and dry electrode are comparable.

In addition, the spectral components of the Ag-AgCl and dry electrode signals over the frequency band of 0.05–20 Hz were analyzed by Fast Fourier Transform (FFT). The power spectral density (PSD) for each electrode is shown in Fig. 6. No marked differences were noted in the frequency spectrum of the wet Ag/AgCl and the dry electrode.

### *(b) Evoked potentials*

Evoked potentials (EPs) are time-locked averages of brain electrical activity. EPs can be generated by either sensory inputs and motor outputs or cognitive processes (Hillyard and Picton 1987). Spontaneous EEG signal amplitudes may reach 60–200 $\mu$V, whereas EP amplitudes typically range from 0.25–20 $\mu$V. EPs are buried in the "noise" of spontaneous, random background EEG and require signal averaging techniques for extraction. We recorded both sensory and cognitive EPs with the standard wet electrodes and the prototype dry electrode. The first study recorded stimulus-dependent visual evoked potentials (VEPs). The subjects' eyes were stimulated with a checkerboard pattern delivered on a video monitor. This stimulation gener-

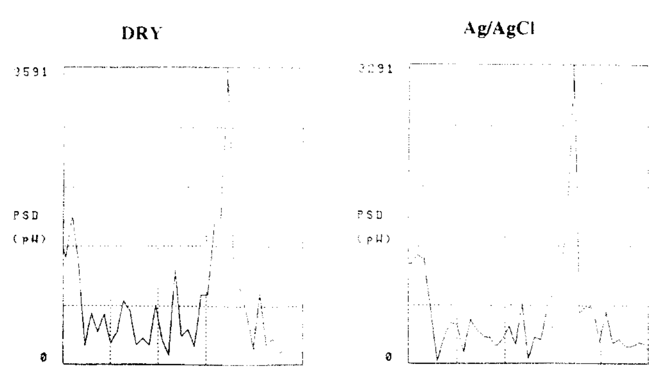



Fig. 6. FFT components of the spontaneous EEG signals shown in Fig. 5. On the left is the dry electrode output and on the right is the Ag/AgCl output. Predominant frequencies are in the alpha range. This are the original data extracted, digitized and analyzed by the EEG instrument. PSD is the power spectral density measured in pico ($10^{-12}$) watts (pW) units. The Cadwell EEG instrument is capable of digitizing and generating the PSD of measured EEG and ERP data after it is recorded.

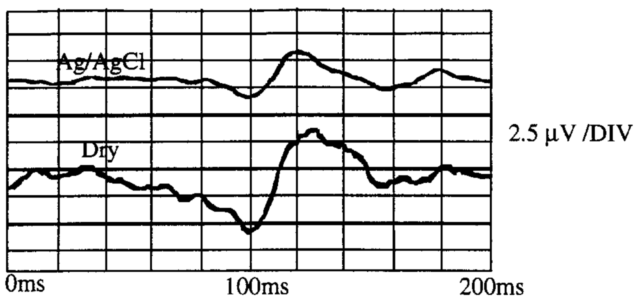



Fig. 7. VEP signals from two types of electrodes. The dry electrode was located 1.5–2 cm caudal to the $O_2$ site. The vertical scale is 2.5 $\mu$V/div for both channels.

ates a prominent P100 component maximal over occipital regions.

Fig. 7 shows the VEP signals from the wet Ag/AgCl and dry electrodes from 1 subject. There were 5 different test runs recorded from the subject to assure data repeatability. The upper trace is the output of the Ag/AgCl electrode and the lower trace is the output of a dry electrode over a 200 msec epoch. At 100 msec, the expected VEP components (P100) are recorded by both electrodes. Higher amplitudes were obtained from the dry electrode in the VEP studies. The impedance match of the dry electrode amplifier to the preamplifier of the Cadwell instrument resulted in these higher signal amplitudes. These amplitude differences were not as apparent in the cognitive ERP studies. This is due to the fact that the EEG instruments used in these experiments were different and had two different input impedance preamplifiers. The amplifier impedance for the spontaneous EEG was an order of magnitude smaller than the impedance of the Grass amplifier (100 G$\Omega$ at DC) used for evoked potential responses. In addition since the noise sensitivity of the dry electrode is lower than the Ag/AgCl there is a higher signal-to-noise ratio for the dry electrode contributing to higher signal amplitudes.

In an auditory recognition memory experiment, cognitive brain potentials including the N400 and P300 responses were recorded (Nielsonn-Bohman et al. 1993). The dry electrode was included in this study along with 23 other Ag/AgCl electrodes. The dry electrode was positioned 2–3 cm lateral to the Cz location. A total of 3 subjects were recorded. Inclusion of the dry electrode in this study permitted further testing of the electrode at low frequency responses. Fig. 8 compares the cognitive evoked potentials of the recorded data from both the dry electrode and Ag/AgCl electrodes located near the center of the scalp (Cz location). Fig. 8a is the grand average of the data from 3 subjects using the Ag/AgCl electrode and Fig. 8b is the grand average of the data from 3 subjects using the dry electrode. There were no significant (n.s.) differences in the $t$ test for amplitudes or latency for the P200 ($t = 1.2$, $P =$ n.s.), N400 ($t = 1.4$, $P =$ n.s.), and P300 ($t = 2.9$, $P =$ n.s.) in this small sample.

The results of these experiments indicate that the dry electrode is effective for spontaneous and evoked EEG recording, with performance comparable to wet electrodes. Key factors in obtaining acceptable signal level are the characteristics of the amplifier. The amplifier has an input impedance of greater than 100 G$\Omega$, output impedance of less than 500 $\Omega$, input bias current of less than 10 pA, consumes low power and has flat amplitude characteristics between 0.01 Hz and 5 kHz. To power the amplifier by batteries, the static and dynamic power consumption should be low. A complimentary metal oxide semiconductor (CMOS) amplifier is desirable since it can continuously operate over 6 months on two 1.5 V cells. Since the output of the dry electrode is fully compatible with standard EEG instruments and amplifiers, no additional hardware is needed to use the electrode with the existing EEG systems.

**Future plans**

The dry electrode and associated electronics will be integrated using the standard microelectronics circuit fabrication techniques. Integrated circuit manufacturing is done by "batch" processing which results in low cost and high reproducibility. The estimated size of the integrated electrode is approximately 4 mm in diameter with a thickness of less than 10 mm, including local battery power sources. Multiple redundant sensor sites will be used on the electrode to increase skin contact. A flexible, head mounted, geodesic strap system is being developed for the attachment of up to 128 dry

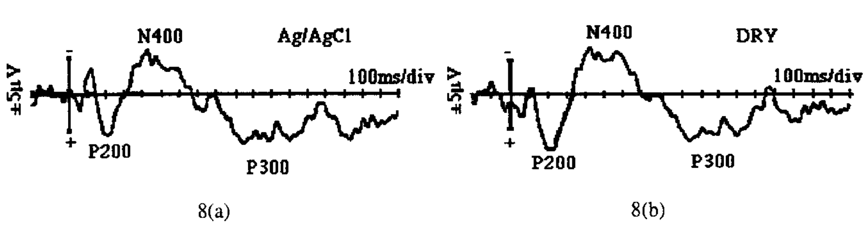



Fig. 8. Cognitive EP signals from 3 subjects using both Ag/AgCl electrodes and dry electrodes located near Cz. a: grand average of the data from the Ag/AgCl. b: grand average of the data from the dry electrode. The staticstics show no significant difference between the two electrodes for the P200, N400 and P300 components.

electrodes in less than 5 min. The geodesic strap will house electrodes with spring contacts that press the electrodes to the scalp through the hair. Medical adhesive tapes will be used to attach the electrode on non-cerebral sites.

## Discussion

Conventional wet electrodes have several features in common: polarization effects are minimum, the electrode-skin interface impedance is low, and the paste is almost physiologically inert. However, the effectiveness of such electrodes is limited under conditions where long recording times are involved. The conductive paste will dry out unless changed, resulting in an interface impedance high enough to cause signal degradation. Over extended periods, the region of the skin in contact with the conductive paste may become irritated. Additionally, skin preparation and attachment times for wet electrodes are high. This is becoming increasingly important in cognitive physiological research where multiple electrodes up to 128 channels have been employed. The dry electrode eliminates the need for paste and skin preparation.

Noise in wet electrodes is enhanced by the space charge layer at the interface between the skin and the electrode paste. The ionic current at this interface is readily altered by motion and sweating. This redistribution of the space charge layer gives rise to interface noise. Since the dry electrode does not use paste, this electrode noise is partially reduced. In addition, the dry electrode senses the electric field of the body rather than the conversion of ionic exchange between the electrode and skin. Electronic noise added to the dry electrode is smaller by two orders of magnitude than the charge layer noise in wet electrodes (Ko and Hynecek 1974). Triboelectrical noise is caused by perturbation (due to stress) of charge at the conductor-insulator interface of the cable that carries the electrode signal to an instrument. Since the dry electrode has a preamplifier for boosting the signal amplitude, triboelectric noise does not appreciably effect signal transmission. The amplifier performs an impedance match, boosts the signal level at the recording site and is able to drive a long cable.

The silicon nitride coated dry electrode with local impedance matching characteristics compared favorably with standard electrodes for EEG and ERP recordings. However, several issues remain to be addressed. Application of active electrodes for EEG requires the electrodes to make good contact to the scalp over the hair to minimize noise caused by motion (Ko and Hynecek 1974; Tam and Webster 1977). Previously reported active electrodes could not be used on the scalp due to the large surface area of the electrodes (typically 6–15 mm in diameter). The electrode's surface contact to the skin varies due to hair and motion causing mechanical decoupling of the electrode surface from the skin. The current dry electrode reduces this problem by utilizing a smaller surface area. Skin-to-electrode impedance changes due to motion are also reduced since a smaller electrode surface area is used. However, the smaller surface area does not guarantee contact, and redundant active sites will be needed to assure adequate contact. During experiments performed in this study low frequency noise was often observed in the dry electrode due to motion artifacts. This noise was caused by the bulkiness and mechanical instability of the "macro" (prototype) dry electrode. Motion artifacts will be minimized by fabricating an integrated electrode using IC technology which will result in a smaller footprint, and reduced sensor surface area.

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