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A Novel Dry Active Electrode for EEG Recording

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***Abstract*—The design and testing of a "dry" active electrode for electroencephalographic recording is described. A comparative study between the EEG signals recorded in human volunteers simultaneously with the classical Ag-AgCl and "dry" active electrodes was carried out and the reported preliminary results are consistent with a better performance of these devices over the conventional Ag-AgCl electrodes.**

***Index Terms*—Bioelectric signal recording, biomedical electrodes, dry electrodes, electroencephalography.**

I. Introduction

"Dry" active electrodes have been studied as an alternative to the silver/silver chloride (Ag–AgCl) electrodes as they can be applied without any skin preparation or gel application [1]–[4]. In this case, *in-situ* active preamplification is essential due to the very high electrode/skin interfacial impedances [2]. The sensor material of the electrode (it makes the contact with skin) can be either a conductor or an insulator and it must be inert in contact with skin sweat, as corrosion often leads to the generation of electrochemical noise and degradation of the biosignal [1], [5]. Most metals are not adequate either because they undergo corrosion [aluminium, stainless-steel (SS)] or because they induce allergic reactions (copper). Several metal oxides also failed due to ageing problems [1]. Matsuo [6] developed a barium titanate sensor but the material proved to generate electrical noise due to its piezoelectric properties. Searle [4] tested a SS-based "dry" active electrodes that, according to the author, compared favourably with the "wet" electrodes. However, Fonseca [5] showed that SS undergoes corrosion after some time in contact with synthetic sweat, with the generation of electrochemical noise. The problem was solved by applying a titanium coating to SS. Thaeri [2] developed a silicon nitride coated SS sensor with suitable chemical and electrical properties, but

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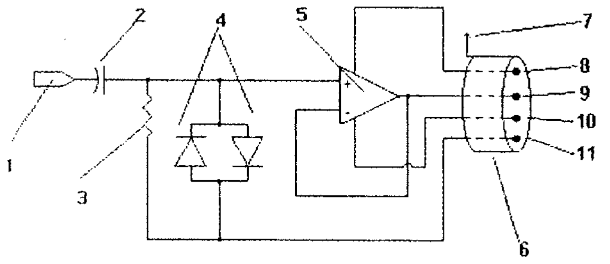

Fig. 1. Layout of the signal processing circuit.

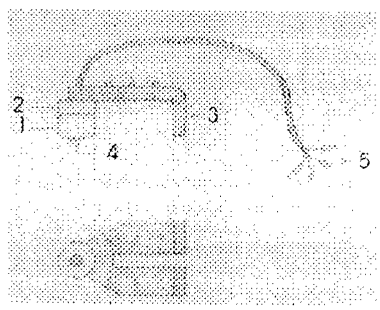

Fig. 2. Schematics of the "dry" active electrode with its support.

owing to the very low thickness of the coating (0.2 μm), it may become unable to withstand an intensive utilization. In this paper we report the development and testing of a novel "dry" active electrode prototype where the sensor is a TiO_2 coated SS substrate. The sol-gel technique, which has been often used to fabricate thin ceramic coatings for electronic applications [7], was used to fabricate films about 0.6 μm thick. The TiO_2-based sensors were tested for their mechanical robustness, electrical and chemical properties in synthetic sweat and they showed to be suitable for electrode application. On the other hand, our preliminary results from a comparative study between the EEG signals recorded in human volunteers simultaneously with Ag-AgCl and TiO_2-based "dry" active electrodes indicate a better performance of these devices over the conventional Ag-AgCl electrodes.

II. Materials and Methods

A. Sensor Fabrication

The sensor was fabricated from a 316L-grade SS disc ($\Phi = 6$ mm, thickness = 2 mm), that was polished to a mirror finish and annealed at 350 °C for 1 hr. The sol-gel method was chosen for the coating procedure. Tetraisopropylorthotitanate (iPrTi) was dissolved in acetilacetone (Acac) (Acac/iPrTi = 1.2, molar) and ethanol (EtOH) (EtOH/iPrTi = 35, molar). Acetone was added to the solution (acetone/iPrTi = 9 in order to obtain a stable stock sol. The coating was applied by using the spin coating method (4000 rpm, 30 s) and the coated samples were annealed at 500 °C for one hour to obtain a TiO_2 layer approximately 150 nm thick (heating and cooling rate: 2 °C/m, 15 m at 60 °C). The coating-annealing cycle was repeated up to four times in order to obtain multilayered systems with suitable mechanical and corrosion resistance properties.

B. Electronic Circuit Assemblage

The schematic of the electronic circuit incorporated in the electrode is reported in Fig. 1. It is composed of a preamplifier (5), two diodes for circuit protection against parasitic high voltages (4), a high-pass filter formed by a 150 pF capacitor (2) and a resistance of 5 GΩ (3). The interface cable (6) has two wires for the power supply, (8,10), one for the electrode output (9), one for reference (should be about the half voltage between 8 and 10) (11) and one for shielding (7). The preamplifier is based upon a very high input impedance operational amplifier in a unity gain configuration. The input impedance must be much larger than the worst case electrode/skin contact impedance. The circuit was mounted in a square PCB with a 6-mm side using SMD electronic components. The differential input resistance of the operational amplifier is 1000 GΩ, the input bias current is 1 pA and the peak-to-peak input noise voltage is 0.76 μV (bandwidth: 0.1–10 Hz), for a power consumption of 2.5 mW.

C. Electrode Assemblage

The electrode's body was machined from a glass ceramic rod (Macor@), Fig. 2 1), height = 15 mm, Φ = 10 mm. Its cover 2) was designed to fit the support accessory 3) currently used with the Ag-AgCl pad electrodes. The sensor 4) was fixed to the electrode with an epoxy resin and the cable 5) contains the wires for electrode interfacing with the acquisition system.

D. Experimental Setup for the EEG Tests Carried Out in Humans

All exams were carried out at the EEG laboratory of HGSA hospital in several healthy volunteers, under the supervision of a neurophysiologist. The new electrode and an Ag-AgCl electrode disc were placed at symmetric positions, P_3 or P_4, or near placed, P_4-O_2 or P_3-O_1, to test alpha rhythms. The Ag-AgCl and the "dry" active electrodes were fixed to the scalp through a conventional rubber head-cap and the usual skin preparation and gel application were performed before application of the Ag-AgCl electrode. The reference electrodes were either an Ag-AgCl electrode (placed at the ear of the volunteer) or a "dry" active electrode placed at the similar or symmetric positions.

III. Results and Discussion

The TiO_2-based sensors were characterized for their chemical stability, mechanical robustness and electrical properties. The chemical stability of the samples in contact with perspiration was mimicked by immersing them in synthetic sweat. After an immersion period of three weeks a very stable and noise-free TiO_2/synthtetic sweat interfacial electrical potential was recorded, using a technique already reported [5]. The potential stability and low electric noise ($<$ 10 μV peak-to-peak, bandwidth: 0.2–4 Hz) observed proved the suitability of these coatings for electrode application from the chemical point of view. The noise bandwidth was tuned to the frequencies typically found for electrochemical noise [5]. It is to remark that if noise was generated by sample corrosion it would be added to the biosignal in recording conditions. The coated samples also showed to have the necessary mechanical properties for daily manipulation, as their performance was not affected by intensive hospital and laboratorial testing. The electrical properties of the films were measured in synthetic sweat solutions, as described [5]. The impedance magnitudes and phase angles (θ) at 2.5, 6.3, and 63 Hz are 2363 Ωcm^2 ($\theta = -70°$), 1142 Ωcm^2 ($\theta = -70°$), and 192 Ωcm^2 ($\theta = 63°$). It is concluded that the oxide displays important dielectric losses but it shouldn't be a significant barrier to signal transfer, at least when a perspiration layer exists. The higher contribution to interfacial impedance should come from skin [8], especially when it is dry. Searle[4] reported a strong decrease of the "dry" active electrodes/skin interfacial impedances in the first 10 min after electrode fixation (3 MΩ – 0.5 MΩ, at 57 Hz), which was ascribed to the formation of a perspiration layer underneath the sensor surface. It is to stress

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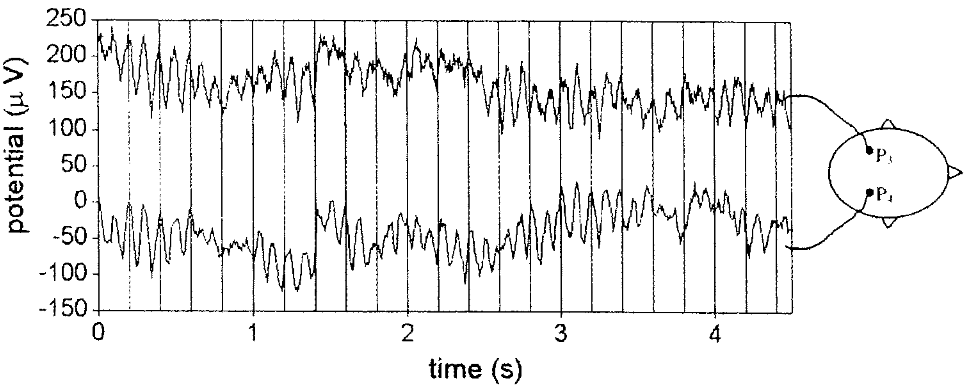

Fig. 3. Simultaneous records of spontaneous EEG signals with the Ag-AgCl (P_3) and "dry" active (P_4) electrodes placed at symmetric positions.

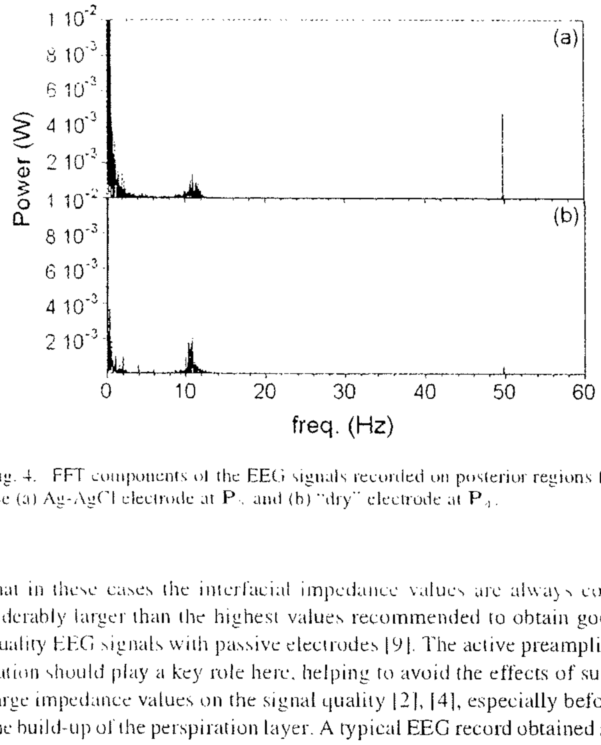

Fig. 4. FFT components of the EEG signals recorded on posterior regions for the (a) Ag-AgCl electrode at P_3, and (b) "dry" electrode at P_4.

that in these cases the interfacial impedance values are always considerably larger than the highest values recommended to obtain good quality EEG signals with passive electrodes [9]. The active preamplification should play a key role here, helping to avoid the effects of such large impedance values on the signal quality [2], [4], especially before the build-up of the perspiration layer. A typical EEG record obtained simultaneously with Ag-AgCl and "dry" active electrodes placed at symmetric positions is reported in Fig. 3, showing that the signals are very similar. However, the power spectra, Fig. 4, prove that the signal monitored by the "dry" active electrode is less affected by the 50-Hz power line noise. Searle [4] and Nishimura [2] reported a similar behavior when they compared Ag–AgCl and "dry" active electrodes, and Searle ascribed such difference to the lower interfacial impedance mismatch between electrodes, due to the active preamplification. From the power spectra it can also be concluded that the "dry" active electrode shows a 8–13 Hz band (α rhythm) similar to that of Ag–AgCl and a lower background noise bellow 2Hz, due to the high-pass filter (cut-off frequency = 0.2 Hz) effect. Even though these conclusions are in good agreement with the results reported by other authors with different active electrodes and experimental setups, further testing is necessary in order to draw definitive conclusions about the effectiveness of this technology for EEG recording. Future work will include the influence of the sensor properties (material, area, and texture), contact pressure, sensitivity to movement, and acquisition time on the performance of the devices.

IV. Conclusions

The prototype of a "dry" active electrode for EEG recording with a TiO_2-based sensor surface was developed. This sensor has proved to have the suitable mechanical properties for daily manipulation and cleaning and it displays a noise-free scalp/sensor interface, even after prolonged contact with the synthetic sweat. Preliminary comparative tests carried out simultaneously in human volunteers with Ag-AgCl and "dry" active electrodes are consistent with the conclusion that these devices perform better than the Ag-AgCl electrodes, particularly in what concerns the low-frequency noise and 50-Hz rejection. Finally, these electrodes can be fixed to the scalp with the conventional rubber head-cap and plugged to the EEG interface, as any conventional Ag–AgCl pad electrode.

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On Computing Dominant Frequency From Bipolar Intracardiac Electrograms

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***Abstract*—Dominant frequency (DF) computed from action potentials is a key parameter for investigating atrial fibrillation in animal studies and computer models. A recent clinical trial reported consistent results computing DF from 30 Hz to 400 Hz bandpass filtered bipolar electrograms in humans. The DF ($<$ 15 Hz and, thus, filtered out) was recovered by rectifying the signal, while the theoretical background of this approach was left uncommented. It is the focus of this paper to provide this background by a Fourier analysis. We demonstrate that it is mainly the timing of the narrow deflections (local activation at the catheter tip) which contribute to the DF peak in the frequency spectrum. Due to the typical signal morphology pronounced harmonic peaks occur in the spectrum. This is a disadvantage when computing the regularity index (RI) as a parameter for local organization and signal quality. It is demonstrated for synthetical and patient data that at low DF the RI is far below the optimal value one even for high underlying organization and good signal quality. The insight obtained promotes the development of better measures for organization. The finding that mainly timing of activation contributes to DF might promote the development of powerful realtime signal processing tools for computing DF.**

***Index Terms*—Atrial fibrillation, Fourier transform.**

I. Background

Atrial fibrillation (AF) is the most common supraventricular arrhythmia [1], [2]. Minimal invasive treatment by catheter ablation has gained increasing clinical importance [3], [4]. However, the obtained long term success rates are still unsatisfactory ($\sim$ 65%) at an unacceptably high risk for severe complications ($\sim$ 7%) [5].

During the 1990s AF was thought to be a completely stochastic atrial activation by a high number of irregularly propagating wavelets. Within the last years new insight from basic research has completely changed the understanding of atrial fibrillation. In the first years of the new millennium it has been shown in animal and computer models that one (or a few) mother wavelets of high-frequency and high spatio-temporal organization (periodicity) are the maintaining mechanism of AF [6], [7]. Recently, these findings have been confirmed in humans by catheter based electroanatomical mapping [8].

The key parameter for developing the new understanding of atrial fibrillation is the dominant frequency (DF, f_D). It reflects the mean local heart rate. In the Langendorff-perfused heart it can be obtained from local action potential recordings [6]. In humans bipolar intracardiac electrograms are the standard approach for recording local activity. These two types of signals have a very different spectral composition. It is not obvious that DF can be computed from both.

Bipolar electrograms are routinely investigated during an electrophysiological (EP) study [3]. Here, the narrow spacing of a pair of catheter electrodes (less than 5 mm) ensures a high sensitivity to local activity. A high-pass filter with a corner frequency of typically about 30 Hz is used for filtering out slow signal changes arising from repolarization. Additionally an anti-aliasing low-pass (about 500 Hz corner frequency) is used. Thus, the signal passing the resulting bandpass contains mainly components of local depolarization. This bandpass is included in the EP recording system clinically used.

Due to the high-pass corner frequency of 30 Hz, one might suspect that the DF (typically 5–15 Hz in humans) is filtered out. In [8] DF is recovered by nonlinear signal processing and consistent results are obtained. However, no theoretical background of the method has been provided. It is the primary scope of the presented study to explain how the DF is retrieved from the spectral components passing the filter and which signal components contribute to the result.

An auxiliary parameter used together with DF is the regularity index (RI) [8]. While naming suggests that it measures regularity and, thus,—organization—it was applied for estimating signal quality. Our study shows that this parameter depends on the DF and, thus, does not provide a valid measure neither for regularity, nor for signal quality.

II. Theory

We first summarize the signal processing approach applied in [8]. The intracardiac signal was bandpass filtered at 30 Hz to 400 Hz and sampled at 1000 Hz. For spectral analysis the signal was rectified (recovering near direct current [dc] spectral components), band bass filtered at 3 Hz to 15 Hz and the amplitude spectrum was computed by a fast Fourier transform (FFT) from a signal segment of $\Delta = 4.096$ s duration ($\sim$ 0.24 Hz resolution). The frequency of the peak with the maximal amplitude is the DF f_D.

In the current study this bandpass is obtained by analyzing only the spectral interval between 3 and 15 Hz after the Fourier transform (FT). This is identical to setting all spectral components outside the bandpass zero (ideal digital filter).

A. *Bipolar Electrogram*

For analytical treatment we choose the time continuous FT and make the following assumptions for modeling the properties of a bipolar electrogram $B(t)$ (see Fig. 1). Within the segment Δ the signal contains N pulses of finite duration. Each pulse $b_n(t - T_n)$ reflects an activation wave front passing the catheter electrodes at time T_n. Due to the small electrode spacing the pulses are narrow, i.e., the characteristic time constant τ_n is much smaller than cycle length (i.e., the interval between two pulses). As the signal is bandpass filtered the FT of each pulse $F\{b_n(t)\}$ is zero at $\omega = 2\pi f = 0$ (i.e., the area under each pulse is zero) and $F\{b_n(t)\} \to 0$ with $\omega \to \infty$ (i.e., the pulse is a

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