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Novel Flexible Dry PU/TiN-Multipin Electrodes: First Application in EEG Measurements

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***Abstract*—Dry biosignal electrodes for electro-encephalography (EEG) are an essential step for realization of ubiquitous EEG monitoring and brain computer interface technologies. We propose a novel electrode design with a specific shape for hair layer interfusion and reliable skin contact. An electrically conductive Titanium-Nitride (TiN) thin layer is deposited on a polyurethane substrate using a multi-phase DC magnetron sputtering technique. In the current paper we describe the development and manufacturing of the electrode. Furthermore, we perform comparative EEG measurements with conventional Ag/AgCl electrodes in a 6-channel setup. Our results are promising, as the primary shape of the EEG is preserved in the signals of both electrodes sets, according to recordings of spontaneous EEG and visual evoked potentials. The variance of both signals is in the same order of magnitude. The Wilcoxon–Mann–Whitney two-sample rank-sum test revealed no significant differences for 25 of the 28 compared signal episodes. Hence, our novel electrodes show equivalent signal quality compared to conventional Ag/AgCl electrodes.**

I. Introduction

BIOSIGNAL electrode technologies for electroencephalography (EEG) have been investigated for several years with the aim of enhancing usability of conventional EEG while also transferring already established knowledge and functional investigation techniques to new fields of application.

Silver/Silver-Chloride (Ag/AgCl) - electrodes are considered the gold standard in biosignal electrodes thanks to their outstanding electrochemical characteristics and stability. However, these electrodes need to be applied in combination with electrolyte gels or pastes. Hence, they suffer from technological inherent drawbacks causing complex, time-consuming and error-prone preparation procedures as well as limited long-time stability of the electrolytes [1], [2]. Moreover, with increasing electrode number and density a permanent risk of falsified measurements, caused by unnoticed conductive bridges between adjacent electrodes, arises.

These limitations constrict usability of conventional EEG in medical research and clinical routine. Furthermore this renders Ag/AgCl electrodes inappropriate for future EEG applications like ubiquitous monitoring in neurology and psychology as well as brain computer interfaces.

The aim for development of so-called dry electrodes is to eliminate the need for additional skin preparation and electrolytes and thus simplify measurements and increase patient comfort. Available concepts include electrically conductive and capacitive [1], [3] as well as opto-electric sensor materials [4]. None of these electrodes are available in a commercially relevant number, as they are either too expensive, require special electronics or once more involve too complex application.

Our research is focused on dry electrode technologies for simplified, mobile EEG acquisition in multichannel setups. In the current study we present a novel concept which we call multipin (MP) electrode. This dry electrode is based on a specifically designed, flexible polyurethane (PU) substrate and an electrically conductive Titanium-Nitride (TiN) coating. We describe our developed electrode design and perform test measurements for direct comparison with conventional Ag/AgCl reference electrodes in terms of electrodes-skin impedance and signal quality.

II. Materials and methods

A. Electrode substrate design

In order to ensure sufficient and stable signal quality, dry, electrically conductive electrodes require a stable and reliable contact between electrode surface and human scalp. Hence, the shape of the electrode substrate must comprise

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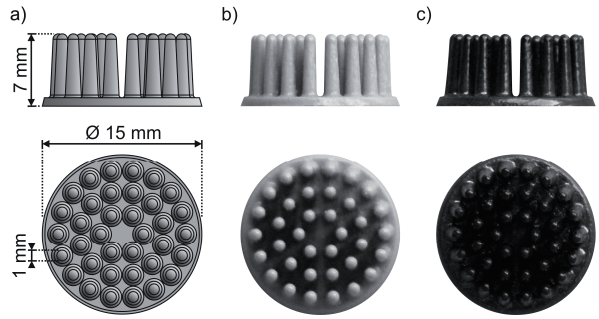

Fig. 1. Side and top view of the multipin electrode: a) basic design, b) uncoated polyurethane substrate, and c) TiN-coated electrode.

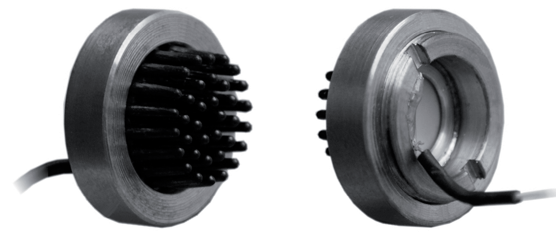

Fig. 2. Different views of the multipin electrode in the brass mounting used for the EEG measurement tests.

several contradictory requirements. Low impedance and stable contact on one hand require a large contact surface, while a small footprint is necessary for hair-layer interfusion and electrode fixation.

Our previous investigations showed that several interconnected titanium pins with small diameters are capable of interfusing the hair-layer and ensure reliable, low contact impedance [5]. Hence, we developed a special electrode shape incorporating 36 electrode pins on a single baseplate. The basic design of the MP electrode is shown in Figure 1a. The number of pins allows for a good balance between patient comfort and large overall contact surface, while the top-diameter of 1 mm enables hair-layer interfusion. Polyurethane was chosen as substrate material, since it is a common and easy to process polymer, whose ability to adapt to the scalp enables improved contact reliability and patient comfort.

B. Titanium-Nitride coating

Titanium-Nitride is known for its outstanding biocompatibility as well as mechanical and electrochemical characteristics thanks to a wide range of well-established medical applications [6]. Different studies also proved sufficient signal quality in multichannel EEG [5] as well as electrochemical stability in contact with human sweat [7].

After cleaning of the electrode substrates in an ultrasound bath with isopropanol and water for 10+10 minutes, homogenous thin films of TiN were deposited during a multi-phase sputtering process, using a reactive DC magnetron sputtering technique in a laboratory-sized deposition system. The specifically developed multi-phase sputtering process produces dense coatings with good adhesion to polymer substrates. The initial polymer substrate and the final TiN-coated electrode are shown in Figures 1b and 1c, respectively.

C. Experimental setup

For electrode contacting and signal acquisition we used unshielded copper cables soldered to special brass electrode mountings. Due to the interconnection of all pins, which is ensured by the conductive coating, the signal was acquired from the outer rim of the electrodes. A picture of an electrode assembly is shown in Figure 2.

According to the international 10-20 system for electrode placement, six electrodes were placed at Fp1, Fp2, TP7, TP8, O1, and O2 positions on the head of a 26 year old volunteer. Next to each MP electrode a conventional sintered Ag/AgCl electrode (EASYCAP GmbH, Germany) was contacted using electrolyte paste (Ten20 conductive, D.O. Weaver and Co., U.S.). The MP electrodes were used in completely dry condition. The actual distance between corresponding MP and Ag/AgCl electrodes was approx. 15 mm (center-to-center).

In combination with two commercial biosignal amplifiers (RefaExt, ANT B.V., The Netherlands) the described measurement setup allowed for simultaneous separate recording and thus direct comparison of the signals acquired with both sets of electrodes.

An additional patient ground electrode (self-adhesive Kendall ARBO electrode, Tyco Healthcare, U.S.) was placed on the left forearm of the volunteer and connected to both amplifiers.

D. EEG tests and evaluation

After application of all electrodes, electrode-skin impedances were measured and several standard EEG tests were performed, including acquisition of resting state EEG, eye blinking artifacts and a visual evoked potential (VEP). During the VEP test 250 checkerboard pattern reversal stimuli were presented according to ISCEV standards [8].

For impedance measurement and signal recording we used the commercial ASA software (ANT B.V., The Netherlands) in combination with a common average reference montage and a sampling rate of 1024 samples/s. For the sake of results comparability, the electrode placement and recording setup was retained during all tests.

Subsequent signal evaluation was executed using MATLAB software (The Mathworks Inc., U.S.) and included filtering (1-40 Hz software 30^{th} order Butterworth bandpass), manual extraction of artifacts visible in the raw data, and VEP averaging. For objective comparison of the recorded signals, Root Mean Square Deviation (RMSD) and RMSD of the normalized signal sequences (RMSDN) were calculated. Therefore corresponding signal episodes of 30 s were analyzed according to equations 1 and 2, respectively.

$$RMSD = \sqrt{\frac{\sum_{i=1}^{n} \left(m_i - a_i\right)^2}{n}} \quad (1)$$

$$RMSDN = \sqrt{\frac{\sum_{i=1}^{n} \left(\frac{m_i}{|m|_{\max}} - \frac{a_i}{|a|_{\max}}\right)^2}{n}} \quad (2)$$

The RMSD represents differences in absolute amplitude, whereas the RMSDN is a parameter for differences in shape. Variables m_i and a_i correspond to the i-th data samples, acquired with the MP and Ag/AgCl electrodes, respectively. Variable n is the overall number of samples within the analyzed data sequence. Furthermore, we separately calculated the variances of the EEG sequences of both electrode sets.

A Wilcoxon–Mann–Whitney (WMW) two-sample rank-sum test at $\alpha = 5\%$ was applied to determine the significance of observed signal differences. Therefore, the corresponding signal episodes (30 s for resting state EEG and eye blinking, 500 ms for VEP) of both electrode sets were directly tested according to the null hypothesis that data vectors m and a are independent samples from identical continuous distributions.

III. Results

A. Electrode-skin impedance

For the MP electrodes we observed impedances between 72 and 125 kΩ before the EEG tests and 65 to 118 kΩ afterwards. The impedance at Fp1 and Fp2 electrodes were considerably lower (72 / 65 kΩ and 79 / 76 kΩ before and after EEG tests) than at the TP7, TP8, O1, and O2 positions (98 – 125 kΩ). The applied Ag/AgCl electrodes showed nearly constant impedances of 18 to 26 kΩ.

B. EEG tests

All electrodes reliably acquired EEG biosignals during the tests which lasted about 40 minutes. However, the MP electrodes were more susceptible to movement artifacts than

TABLE I
EEG signal differences Ag/AgCl – PU/TiN

Parameter	Resting state EEG	Eye-blinking artifacts	Pattern reversal VEP
RMSD	8.4 µV	10.8 µV	1.0 µV
RMSDN	13.6%	15.6%	27.4%
Variance EEG (Ag/AgCl)	63.0 µV	186.5 µV	2.4 µV
Variance EEG (MP)	53.6 µV	185.2 µV	2.9 µV

the Ag/AgCl electrodes. After extracting artifact-afflicted signal episodes, the RMSD and RMSDN parameters were calculated. The results listed in Table I represent the mean values over all six channels. The RMSD and RMSDN values of EEG containing eye-blinking artifacts are higher than in resting state EEG. The variances of MP and conventional Ag/AgCl electrodes are in the same order of magnitude for resting state EEG as well as for EEG containing eye-blinks. Furthermore, the variances of the eye-blinking EEGs are nearly tripled, compared to the variances of the resting state EEGs. It is interesting to notice that averaging of 250 VEP trials expectedly decreases signal differences in terms of mean RMSD. In fact the overall lowest RMSD is achieved in the VEP results. However, the VEP also shows the highest RMSDN value.

An overlay plot of the mean spectral power density of 30 seconds of resting state EEG is shown in Figure 3a. The power is strongly increasing with lower frequencies for both electrode sets. For frequencies below 3 Hz, the spectral power of the MP electrodes is considerably higher than for Ag/AgCl electrodes, what corresponds to an increased drift as well as higher offset potentials. This observation complies with the characteristics of type I electrodes as well as the results of our previous investigations [5]. Figure 3b shows an overlay plot of 3 seconds of EEG containing an eye blinking artifact. An overlay plot of the Global Field Power (GFP) of the VEP is visible in Figure 3c, including an episode of 100 ms prestimulus and 400 ms poststimulus signal. The GFP was calculated according to Lehmann et al.

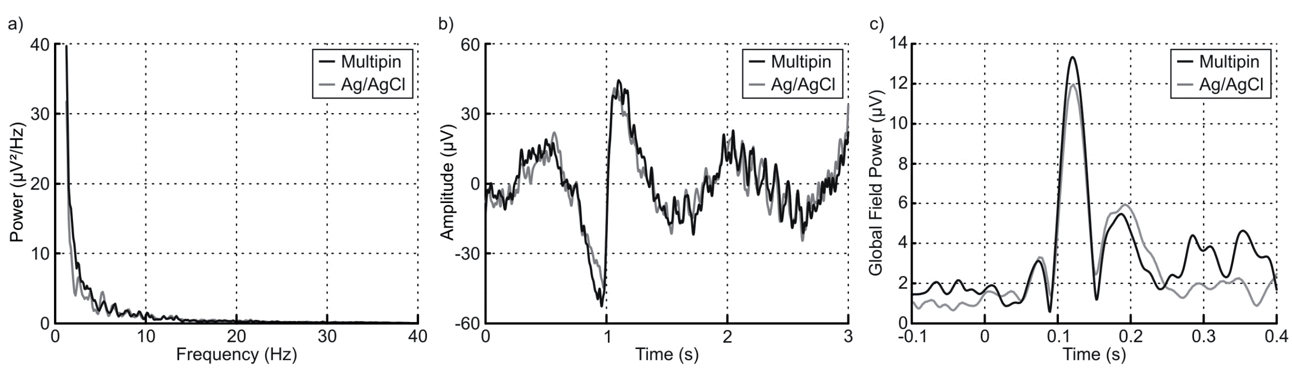

Fig. 3. Overlay plots of characteristic EEG signals acquired with novel multipin electrodes (black) and conventional Ag/AgCl electrodes (gray): a) Welch power spectrum of 30 seconds of resting EEG, b) 3 seconds of EEG containing eye blinking artifact, and c) Global Field Power of the VEP.

[9]. The main components N75, P100, and N135 are clearly visible in both recordings and exhibit amplitudes in the same order of magnitude as well as similar latency. The compared signals show good congruence, especially at higher amplitudes, whereas differences mainly appear at low-amplitude sections.Furthermore, statistical testing revealed no significant differences for 25 of the 28 evaluated corresponding signal episodes. A conventional T-test was not applicable as a preliminary executed Kolmogorov-Smirnov test refused the constraint of normal distributed data.

IV. Discussion

Our developed electrode substrate design proved sufficient for good hair layer interfusion and reduced electrode-skin impedance. Compared to our previous single pin electrode concepts the impedance is considerably decreased. Moreover our volunteers reported increased comfort during application and measurement. This effect is caused by decreased contact pressure and larger contact area thanks to the combined surface of the multiple electrode pins. The systematic impedance decrease during the examination time should be related to the formation of a perspiration layer underneath the electrodes. In addition, the dependence of the electrode impedance on its position may be related to non-existing hair at the frontal electrode positions.

A TiN coating, applicable for EEG biosignal acquisition, was successfully deposited on the PU substrates and allowed for acquisition of EEG signals during several tests.

The electrodes were contacted using a custom brass mounting and soldered copper wires. This contacting technique showed reliable results during validation of the electrode concept. However, it is inappropriate for multichannel caps as it substantially increases the overall sensor weight and complexity and even induces wearing of the TiN coating. Furthermore, this experimental mounting can damage the coating due to shear stress during electrode mounting and dismounting process. This leads to the conclusion that the future mechanical contact must not induce sheer stress to the film.

We successfully acquired EEG biosignals with the MP electrodes in a completely dry condition. For comparison and reference, additional signals were recorded using standard Ag/AgCl electrodes. The results of spontaneous EEG recordings (resting state and eye blinks) and VEPs are promising. Variances and signal shapes are similar. The RMSD values are low with respect to the overall amplitude of the signals. Furthermore, the RMSDN values pinpoint that most of the signal shape and thus most of the signal information is sustained. Nevertheless, the magnitudes of RMSD and RMSDN are slightly increased compared to our previous investigations [5]. This suggests that the errors are not limited to spatial distance between the electrodes. Owing to the fact that the RMSD is strongly decreased in the VEP test, it is reasonable to assume that a considerable amount of environmental noise is included in the recordings. Such noise can be reduced by replacing the present simple copper wires with shielded coaxial cables.

The higher spectral power below 3 Hz is related to increased electrode drift and a considerably higher offset potential. This is a characteristic of electrodes of type I and cannot be eliminated completely.

The MP electrodes showed strong sensitivity to movement artifacts, thus requiring careful contacting and fixation to the skin. This was not sufficiently accomplished within the present experimental setup and will be in focus for development of a cap system for dry multichannel EEG.

V. Conclusion

We developed a novel electrode technology for dry EEG acquisition which can contribute to increased utilization of EEG in clinical routine. Furthermore, they enable new fields of EEG applications like continuous mobile monitoring and brain computer interfaces. Our future work will include optimization of substrate stiffness and coating parameters and the development of a complete multichannel cap system.

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