Effect of Sweating on Electrode-Skin Contact Impedances and Artifacts in EEG Recordings With Various Screen-Printed Ag/AgCl Electrodes

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ABSTRACT In response to the growing clinico-economic need for comprehensive home-based sleep testing, we recently developed a self-applicable facial electrode set with screen-printed Ag/AgCl electrodes. Our previous studies revealed that nocturnal sweating is a common problem, causing low-frequency artifacts in the measured electroencephalography (EEG) signals. As the electrode set is designed to be used without skin abrasion, not surprisingly this leads to relatively high electrode-skin impedances, significant impedance changes due to sweating and an increased risk of sweat artifacts. However, our recent electrochemical \textit{in vitro} investigations revealed that the sweat artifact tolerance of an EEG electrode can be improved by utilizing an appropriate Ag/AgCl ink. Here we have investigated \textit{in vivo} electrode-skin impedances and the quality of EEG signals and interference due to sweating in the population of 11 healthy volunteers. Commercial Ag and Ag/AgCl inks (Engineered Conductive Materials ECM LLC and PPG Industries Inc.) were used to test electrode sets with differently constructed ink layers. Electrode-skin impedances and EEG signals were recorded before and after exercise-induced sweating. There was extensive variation in the electrode-skin impedances between the volunteers and the electrode positions: 14.6–200 kΩ (PPG electrodes) and 7.7–200 kΩ (ECM electrodes). Sweating significantly decreased ($p < 0.05$) the impedances in most cases. The EEG signal quality was assessed by comparing average band powers from 0.5 to 2 Hz before and after sweating. Only slight differences existed between the ECM and PPG electrodes; however, the lowest band power ratio (\textit{i.e.} the smallest increase in the band power due to sweating) was achieved with ECM electrodes.

INDEX TERMS Biomedical electrodes, bioimpedance, electroencephalography, screen printing, sleep monitoring, sweat artifact.

I. INTRODUCTION

The standard medical test for diagnosing sleep disorders such as obstructive sleep apnea (OSA) is in-lab polysomnography (PSG) \cite{1}. PSG is an overnight measurement and provides detailed physiological data including electroencephalogram (EEG), electrooculogram (EOG), electromyogram (EMG), and cardiorespiratory signals. Though PSG has many advantages, it is considered too expensive and time-consuming for routine OSA diagnostics \cite{2}. As OSA is a global public health burden affecting nearly 1 billion adults \cite{3} being associated with severe health consequences, new cost-efficient alternatives to the standard PSG are urgently needed.
Currently, home sleep apnea testing (HSAT) can be used as an alternative diagnostic method in patients with a high pre-test probability of OSA [4]. In some geographical areas, such as in the Nordic countries, HSAT is currently the preferred method [5]. Typically, HSAT includes at least the airflow, respiratory effort, and blood oxygenation measurements [4] but no EEG recordings which are needed in order to make a reliable assessment of sleep quality and quantity. The lack of self-applicable, medical-grade EEG electrodes has limited the wider generalization of HSAT. Therefore, a reliable and self-applicable EEG electrode set suitable for home sleep testing would help meet the evident need for less expensive and more accessible diagnosis of sleep disorders.

To respond to the increasing demand for comprehensive home sleep testing, we have recently developed an easy-to-use facial electrode set [6], [8], [9]. This disposable electrode set provides multichannel EEG, EOG, EMG, and ECG recordings with the screen-printed silver/silver chloride (Ag/AgCl) electrodes embedded in a flexible polyester film. The electrodes are covered by an adhesive hydrogel membrane, which acts as an electrolyte and forms a tight contact with the skin. Therefore, no separate electrolyte gel or skin abrasion is needed. We have previously demonstrated that the facial electrode set is well-suited for patients’ self-use and is highly reliable in providing high-quality signals from the forehead, making it a promising tool for home measurements [6]. However, in a few cases, harmful low-frequency sweat artifacts interfered with the recordings [6].

Excessive nocturnal sweating is a common symptom in OSA patients [10], [11]. Consequently, sweat-induced artifacts typically occur in sleep recordings. The sweat artifact is characterized by high-amplitude, low-frequency waves [12] which may well overlap with the frequencies present during slow-wave sleep (SWS). As SWS is characterized by 0.5-2 Hz waves [1], sweat-induced artifacts may reduce the technical quality of recordings and severely hinder the interpretation of EEG signals.

The sweat artifact originates from the properties of the electrode-skin interface, which is presented with its electrical equivalent circuit model in Fig. 1. The effect of the electrode-electrolyte (i.e. Ag/AgCl-hydrogel) interface is represented by the half-cell potential (V_HC) of the electrode, and the capacitance (C_DL) and resistance (R_DL) of the electrical double layer. The skin is divided into three layers: epidermis, dermis, and subcutaneous tissue. The outer layer of the epidermis, the stratum corneum, has a relatively high resistance and the potential difference between the stratum corneum and the hydrogel, V_SC, in the electrical equivalent circuit represents the potential difference between the stratum corneum and the hydrogel. The effect of the epidermis is represented by the parallel C_E - R_C circuit. The effect of sweat glands and ducts (V_S, R_S, and C_S) can be omitted in the equivalent circuit if there is no perspiration. However, when the sweat glands are filled, they provide a parallel low-impedance channel through the skin and the conductivity of the skin increases rapidly [7], [13]. These rapid changes in skin conductivity and the uneven distribution of the sweat glands may result in an impedance mismatch of the measuring electrode pair, making the measurement vulnerable to major artifacts.

Another reason for sweat artifacts, in addition to the impedance mismatch, may be the change in the electrolyte concentration and composition caused by perspiration [12], [14]. Sweat contains substances such as sodium chloride and lactic acid [15] which may interfere with the electrochemical properties of the electrode-electrolyte interface [14] and possibly change the V_HC (Fig.1). Furthermore, our previous in vitro electrochemical investigations revealed that the electrochemical stability of screen-printed electrodes decreases in artificial sweat, but this reduction can be minimized by a careful selection of Ag/AgCl ink [16]. Therefore, we assume that the impedance mismatch might not be the only reason for sweat artifacts.

Based on the results obtained in our previous in vitro study [16], we selected six out of nine different screen-printed electrodes for the present in vivo study with 11 healthy volunteers. The present study compares two commercial Ag and Ag/AgCl ink materials manufactured by either Engineered Conductive Materials ECM LCC or PPG Industries Inc. and three differently constructed ink layers implemented by Screentec Oy (Oulu, Finland). We have investigated the effect of sweating on electrode-skin contact impedances and the quality of the EEG signal to verify whether the findings of the recent electrochemical bench tests would be transferrable to in vivo measurements.

II. MATERIALS AND METHODS

A. DESIGN OF THE ELECTRODES

The screen-printed electrode sets (Fig. 2) for in vivo measurements were manufactured by Screentec Oy (Oulu, Finland) in a similar technological process as previously described in detail [6], [16]. The electrodes were printed in a flat-bed sheet silk screen printing unit on a flexible polyethylene terephtha-
The test electrode set used in measurements comprising three (A, B, C) differently constructed electrodes (see details in text and Table 1).

TABLE 1. Specification of different electrode types used in skin measurements.

<table>
<thead>
<tr>
<th>Electrode</th>
<th>Ink manufacturer</th>
<th>1st layer of the electrode</th>
<th>2nd layer of the electrode</th>
</tr>
</thead>
<tbody>
<tr>
<td>ECM A</td>
<td>Engineered Conductive Materials ECM</td>
<td>Ag filled circle d=8mm</td>
<td>Ag/AgCl filled circle d=8mm</td>
</tr>
<tr>
<td>ECM B</td>
<td>Engineered Conductive Materials ECM</td>
<td>Ag ring d_{inner}=8mm, d_{outer}=6mm</td>
<td>Ag/AgCl filled circle d=8mm</td>
</tr>
<tr>
<td>ECM C</td>
<td>Engineered Conductive Materials ECM</td>
<td>-</td>
<td>Ag/AgCl filled circle d=8mm</td>
</tr>
<tr>
<td>PPG A</td>
<td>PPG Industries Inc.</td>
<td>Ag filled circle d=8mm</td>
<td>Ag/AgCl filled circle d=8mm</td>
</tr>
<tr>
<td>PPG B</td>
<td>PPG Industries Inc.</td>
<td>Ag ring d_{inner}=8mm, d_{outer}=6mm</td>
<td>Ag/AgCl filled circle d=8mm</td>
</tr>
<tr>
<td>PPG C</td>
<td>PPG Industries Inc.</td>
<td>-</td>
<td>Ag/AgCl filled circle d=8mm</td>
</tr>
</tbody>
</table>

B. IMPEDANCE AND EEG MEASUREMENTS ON SKIN

Six screen-printed electrode types based on two different Ag/AgCl inks were tested in healthy adult volunteers. The Research Ethics Committee of the Northern Savo Hospital District issued a favorable statement (849/2018) on the in vivo measurements. Eleven healthy volunteers, 6 males and 5 females aged 20-40 years were enrolled in the experiment and signed an ethical consent form. First, an electrode set based on either PPG or ECM inks (randomly selected) was attached to the right side of the volunteer’s forehead and the other set to the left side (Fig. 3). Corresponding electrode sets were attached to the skin behind the ears (mastoid) and used as reference electrodes. A common Ag/AgCl ground electrode Ambu Neuroline 71005-K (Ambu A/S, Ballerup, Denmark) was placed in the midline on the forehead between the electrode sets. Before the electrodes were attached to the skin, each electrode placement site was gently swiped with alcohol to improve electrode adhesion and decrease the effect of the stratum corneum (Fig. 1). The test subject was then asked to lie down in a supine position and...
electrode-skin impedances at 15 Hz were measured using Siggi II by Easycap (Brain Vision UK Ltd, London, UK). Siggi II measures simultaneously the impedance between the active (i.e. a forehead) and the common ground electrodes and between the reference (mastoid) and the common ground electrodes using sinusoidal test current. The test subject’s EEG signal from all electrodes was recorded for 5 minutes with a Nox A1 PSG device (Nox Medical, Reykjavík, Iceland). The sampling frequency was 200 Hz. After these initial measurements, volunteers undertook a bicycle ergometer test with incremental 20 W/min protocol and continued to cycle until they started to perspire (i.e. visible droplets of perspiration on face) or they were exhausted. The duration of the ergometer test varied from 9 to 16 minutes. When sufficient sweat secretion had been achieved, the test subjects were asked to lie down again and the impedance and EEG measurements were repeated.

C. STATISTICAL ANALYSIS

The electrode-skin impedance magnitudes were normalized with the maximum reading (200 kΩ) of the Siggi II impedance meter. The normalized mean electrode-skin impedance magnitudes were calculated separately for the ECM A, B, and C forehead and mastoid electrodes as well as for the PPG A, B, and C forehead and mastoid electrodes. The Wilcoxon rank-sum test was applied to identify if there were significant differences in the normalized impedance values between different electrode types. Wilcoxon signed-rank test was used to compare the normalized impedance values before and after sweating and the values of the mastoid and forehead electrodes from each volunteer. Wilcoxon signed-rank test is well suited for this purpose because the samples are dependent and the test does not assume data to be normally distributed. A p-value of less than 0.05 was considered to indicate a statistically significant difference. Power spectral densities (PSDs) and average band powers (from 0.5 to 2 Hz) of the EEG signals were calculated using Welch’s periodogram and band power functions of Matlab R2017b (Mathworks, Natick, Massachusetts, USA) for each electrode type in all volunteers before and after sweating to estimate the magnitude of the sweat artifact. The average band power was computed by integrating the PSD from 0.5 to 2 Hz. The duration of the EEG signal sample used for both the PSD and average band power calculation was 60 seconds. The 60-second time period was chosen to achieve an adequate frequency resolution for the PSD and average band power calculations. The selected EEG sample before sweating was the first artifact-free (no eye or body movements) 60-seconds period. After the exercise, the first 60-second period without motion artifacts was selected. The frequency range from 0.5 to 2 Hz is typical in slow-wave sleep [1], and thus the same range was selected for the calculations. The average band power after sweating was then divided by the average band power prior to sweating and this ratio was calculated for each electrode type.

III. RESULTS

A. ELECTRODE-SKIN IMPEDANCES

There was extensive variation in the absolute values of the electrode-skin impedances between the volunteers: 14.6–200 kΩ (PPG electrodes) and 7.7–200 kΩ (ECM electrodes). Sweating changed the absolute impedance values to 4.1–138% (PPG) and to 8.4–139% (ECM) of the original values. In most cases, sweating significantly decreased (p < 0.05) the impedance values with the exception of volunteer #1 (the impedance values of four electrodes increased) and volunteer #5 (the impedance values of three electrodes increased). Fig. 4 shows the normalized average electrode-skin impedances with each electrode type. The mean electrode-skin impedance with the mastoid electrodes was higher than the impedances with the forehead electrodes, but due to the extensive inter-subject variation, these differences were not statistically significant (p > 0.4) at a group level. After the bicycle ergometer test, the normalized average impedance values with all electrode types decreased. Except for ECM C electrodes (both forehead and mastoid, p = 0.066 and p = 0.66, respectively), the decreases in the impedances were statistically significant (p = 0.001 – 0.023). After exercise, the decrease of impedance values was more significant with the forehead electrodes (p = 0.001 – 0.01) than with the mastoid electrodes (p = 0.003 – 0.023) (Fig. 4). There were no systematic differences in the impedance values between the ECM and PPG electrodes (p = 0.15 – 0.95) or between the A and B electrodes (p = 0.28 – 0.9), but the impedance values with ECM A and ECM C mastoid electrodes were different (p = 0.028). The decrease in impedance due to sweating was slightly smaller with C electrodes in comparison to the other electrode types (Fig. 4). The ECM and PPG test electrodes remained attached in all cases and all 11 test subjects were included in the electrode-skin impedance investigation.

B. EEG SIGNAL QUALITY

Four out of 11 test subjects did not generate visible droplets of sweat on their forehead before exhaustion and consequently, no sweat-induced artifacts occurred in the measured EEG signal. These four cases were therefore excluded from the signal quality investigations. The sweat artifact was clearly visible in all measured EEG signals of the remaining seven volunteers and a representative example of this artifact is shown in Fig. 5. The observed frequency range of sweat artifacts was up to 2 Hz (Fig. 6). No major differences existed between the ECM and PPG electrodes but the lowest average band power ratio (i.e. the smallest increase in the band power due to sweating) was invariably achieved with ECM electrodes in all volunteers (Table 2). No correlation between the impedance mismatch (i.e. the impedance difference between the measuring electrode pair) and the sweat artifact (i.e. the band power of the signal) was found (Table 2). In addition, no systematic differences in signal quality between layer configurations (i.e. A, B, and C electrode types) were found (Table 2).
FIGURE 4. Normalized mean impedances of each electrode type before and after sweating. The error bars indicate standard deviations. The impedance ratio (after/before) of each electrode type: 0.33 (ECM A forehead), 0.19 (ECM B forehead), 0.58 (ECM C forehead), 0.54 (ECM A mastoid), 0.78 (ECM C mastoid), 0.28 (PPG A forehead), 0.33 (PPG B forehead), 0.39 (PPG C forehead), 0.51 (PPG A mastoid), 0.54 (PPG B mastoid) and 0.58 (PPG C mastoid). The impedance ratio between the forehead and mastoid electrodes before/after sweating: 0.86/0.52 (ECM A), 0.90/0.30 (ECM B), 0.71/0.53 (ECM C), 0.85/0.46 (PPG A), 0.83/0.51 (PPG B), and 0.95/0.64 (PPG C). Statistically significant \( p < 0.05 \) change in normalized impedance after sweating is denoted with an asterisk (\(^*\)).

TABLE 2. Impedance mismatch \( \Delta Z \) after sweating between measuring electrode pair and the band power ratio (i.e. the average band power after sweating divided by the average band power before sweating) in the frequency range from 0.5 to 2 Hz. N/A: Measurement failed.

<table>
<thead>
<tr>
<th>Test subject</th>
<th>ECM A ( \Delta Z ) (( \Omega ))</th>
<th>ECM B ( \Delta Z ) (( \Omega ))</th>
<th>ECM C ( \Delta Z ) (( \Omega ))</th>
<th>PPG A ( \Delta Z ) (( \Omega ))</th>
<th>PPG B ( \Delta Z ) (( \Omega ))</th>
<th>PPG C ( \Delta Z ) (( \Omega ))</th>
<th>BP</th>
</tr>
</thead>
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<tr>
<td>1</td>
<td>130</td>
<td>150</td>
<td>150</td>
<td>120</td>
<td>160</td>
<td>N/A</td>
<td>1.1</td>
</tr>
<tr>
<td>2</td>
<td>35</td>
<td>120</td>
<td>110</td>
<td>10</td>
<td>N/A</td>
<td>80</td>
<td>1.0</td>
</tr>
<tr>
<td>3</td>
<td>40</td>
<td>30</td>
<td>100</td>
<td>60</td>
<td>N/A</td>
<td>80</td>
<td>1.2</td>
</tr>
<tr>
<td>4</td>
<td>90</td>
<td>100</td>
<td>N/A</td>
<td>60</td>
<td>100</td>
<td>60</td>
<td>2.7</td>
</tr>
<tr>
<td>5</td>
<td>40</td>
<td>70</td>
<td>100</td>
<td>60</td>
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<td>80</td>
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<td>1.7</td>
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<td>7</td>
<td>40</td>
<td>60</td>
<td>40</td>
<td>70</td>
<td>40</td>
<td>30</td>
<td>0.8</td>
</tr>
</tbody>
</table>

IV. DISCUSSION

In the present study, we compared the sweat artifact tolerance of six different types of screen-printed electrodes with 11 healthy volunteers. The aim was to investigate the effect of sweating on the electrode-skin impedances and the quality of the measured EEG signal. We measured the electrode-skin impedances and the EEG signal before and after exercise-induced sweating to reveal possible differences between the electrode types. Our results demonstrate that sweating reduced the absolute skin-electrode impedance values in almost all electrodes. Furthermore, the impedances with the forehead electrodes decreased more than the impedance values with the mastoid electrodes. However, there were no systematic differences between the impedances with the
ECM and PPG electrodes (different Ag/AgCl ink manufacturers) or between the A, B, and C electrodes (different ink shape) before or after sweating, although the impedance decrease due to sweating was slightly smaller with C electrodes. A sweat artifact was clearly visible in all measured EEG signals when droplets of sweat were seen on the volunteer’s forehead. Based on the conducted signal quality analysis (i.e. comparison of the average band power from 0.5 to 2 Hz before and after sweating), the lowest increase of the average band power due to sweating was invariably achieved with the ECM electrodes. No systematic differences in signal quality between A, B, and C electrode types were found as was expected based on the results of the previous in vitro study [16].

It is known that the skin impedance varies between individuals and it depends on various skin properties such as the number and distribution of sweat glands, skin contour, and the thickness of the stratum corneum [13], [15], [17]. In the present study, the measured electrode-skin impedances on unabraded skin varied between 10 and 200 kΩ, which is in line with previous reports [18], [19].

As a study limitation, it should be noted that the measurement frequency of the impedance meter was constant (15 Hz) instead of impedance spectroscopy with multiple frequency points, which could have provided more detailed information on electrode-skin impedance as a function of frequency. However, as the sweat accumulated on the skin decreases quickly after exercise, the measurement time for one electrode should be as short as possible to maintain a similar sweat level during the impedance measurements. The frequency sweep for each electrode requires extra time and the impedance spectroscopy would not have been optimal for this study. Another limitation was the operation range (0-200 kΩ) of the impedance meter Siggi II and the values above the maximum reading of 200 kΩ were regarded as maximum after normalization. Consequently, this saturation of the measurement range slightly disrupts the impedance results. However, the aim of the impedance analysis was to investigate changes due to sweating, not to compare the absolute impedance values. Therefore, we believe that the present impedance measurements are adequate for the purposes of this study. However, to investigate the sweat-induced impedance changes with...
more details, an impedance meter with wider measurement range should be used.

In addition to the large variation in the skin impedances between individuals, the results of this study show that the impedances of the mastoid electrodes were systematically higher than the impedances of the forehead electrodes (Fig. 4). This difference is probably caused by the more curved contours of the skin behind the ear, the larger number of the hair follicles, and the larger amount of sebum. Although the electrodes in this study are flexible, better skin contact is achieved when the skin under the electrode is flat and hairless (e.g. forehead). Since the EEG signals in our recently developed electrode set [6] are measured using bipolar inputs (i.e. voltage measurement between the forehead and the corresponding mastoid electrode, which is the current clinical recommendation [1]), an impedance mismatch between the measuring electrodes might make the measurement more susceptible to sweat artifacts. However, no correlation was detected between the degree of this impedance mismatch and the observed sweat artifacts (Table 2). Four test subjects did not generate visible droplets of sweat on their forehead and although their electrode-skin impedances decreased, no sweat-induced artifacts occurred. Thus, we assume that the reason for the sweat artifacts observed in this study is more likely the change in the electrolyte concentration caused by accumulated sweat on the skin. It should be, however, noted that the number of the test subjects is limited, and further investigations with proper statistical analyses are needed to verify these results.

The average band power for each electrode type before and after perspiration was calculated to estimate the signal quality. As the sweat artifacts are easily misinterpreted as SWS, the frequency band from 0.5 Hz to 2 Hz was selected for these calculations. However, it must be noted that the frequency bands of sweat artifact and SWS are not completely equivalent, which may have a slight effect on the presented results of the signal quality.

According to the previous in vitro measurements conducted in an electrochemical cell, the PPG electrodes were the most unstable after immersion in artificial sweat whereas the ECM electrodes performed in the most stable manner [16]. We hypothesized that the differences between the inks might be due to their chemical compositions. In the present study, the observed differences between the PPG and ECM electrodes were minor, even though the lowest increase in the average band power due to sweating was always achieved with the ECM electrodes (Table 2). The smaller differences were, however, expected due to the more complex nature of the electrode-skin interface compared to the physiological saline solution and artificial sweat environments used in the previous study [16].

Especially in home-based sleep testing, any means to improve the electrodes’ tolerance of the sweat artifact would be beneficial as modifications to the test set-up (such as decreasing the room temperature or reattaching the
electrodes) are not normally possible. The frequency ranges of the sweat artifacts observed in this study varied between volunteers but overlapped with the frequency band of slow-wave sleep (0.5-2 Hz) in all cases. Furthermore, even in healthy individuals, sweating is most intense during slow-wave sleep [21]. Sweat artifacts may, therefore, be easily confused with slow-wave sleep, complicating the interpretation of the signals and possibly leading to erroneous diagnostic decisions. Thus, a reduction of sweat artifacts is critically important, even though the total elimination of these artifacts was not achieved.

In addition to the diagnosis of sleep disorders, flexible EEG electrodes have been recently developed for long-term measurements, brain-computer interfaces, and personalized healthcare applications [22]–[25]. These applications are also vulnerable to sweat-induced artifacts due to the long-term use and the daily activities to which they are exposed. Thus, further studies aiming to minimize sweat artifacts would also be beneficial for the development of these applications. For example, the effect of the hydrogel layer should be further studied, as in this work, the type and the thickness of the hydrogel layer were kept unchanged. However, this present study, together with the previously published study [16] provides a sweat artifact test platform for researchers developing, testing and validating new wearable sensors.

To conclude, we recorded EEG signals and measured the electrode-skin impedances of wake test subjects with the six different screen-printed electrodes before and after exercise-induced sweating. This study aimed to investigate the sweat-induced changes in impedances and in power spectral densities of the measured EEG signals to reveal possible sweat artifacts. Our study revealed substantial inter-subject, inter-location and sweating induced variations in the electrode-skin impedances. Consistently with the previous in vitro test [16], the smallest sweat artifact was achieved with the ECM electrodes, albeit no overall elimination of sweat artifact was achieved. Even though the observed differences between the ECM and PPG electrodes were not extensive, these novel testing methods and results can be used as a solid foundation for further studies. For example, our facial electrode set will be updated with the ECM based ink and further in vivo studies will be conducted.

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REFERENCES


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