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

LAURA KALEVO[1,2], TOMI MIETTINEN[3], AKSELI LEINO[1,2], SAMU KAINULAINEN[1,2], HENRI KORKALAINEN[1,2], KATJA MYLLYMAA[1,2], JUHA TÖYRÄS[1,2,4], TIMO LEPPÄNEN[1,2], TIINA LAITINEN[1], AND SAMI MYLLYMAA[1,2]

[1]Diagnostic Imaging Center, Kuopio University Hospital, 70210 Kuopio, Finland
[2]Department of Applied Physics, University of Eastern Finland, 70210 Kuopio, Finland
[3]Cancer Center, Helsinki University Hospital, Finland
[4]School of Information Technology and Electrical Engineering, The University of Queensland, Brisbane, QLD 4072, Australia

Corresponding author: Laura Kalevo (laura.kalevo@uef.fi)

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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 *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 *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 (*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) [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 [2]. As OSA is a global public health burden affecting nearly 1 billion adults [3] being associated with severe health consequences, new cost-efficient alternatives to the standard PSG are urgently needed.

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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 term 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.

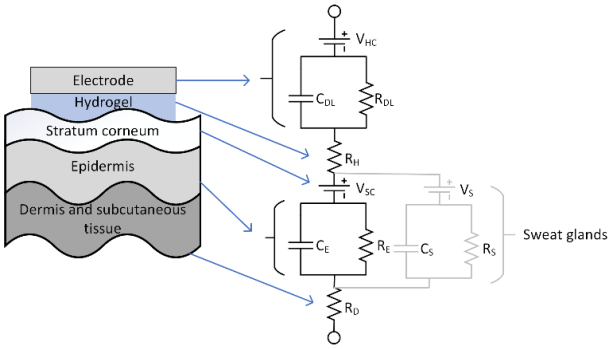

FIGURE 1. The skin layers and the equivalent circuit model of an electrode-skin interface. V_{HC}: half-cell potential of the electrode, C_{DL}: capacitance of the double layer in the electrode-electrolyte (i.e. hydrogel) interface, R_{DL}: resistance of the double layer in the electrode-electrolyte interface, R_H: resistance of the hydrogel, V_{SC}: potential difference between the stratum corneum and the hydrogel, C_E: capacitance of the epidermis, R_E: resistance of the epidermis. The effect of the filled sweat glands is represented by the terms V_S, R_S, and C_S. If there is no perspiration, V_S, R_S, and C_S can be omitted. R_D representing the resistance of the dermis and subcutaneous tissues is negligible [15], [20].

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-

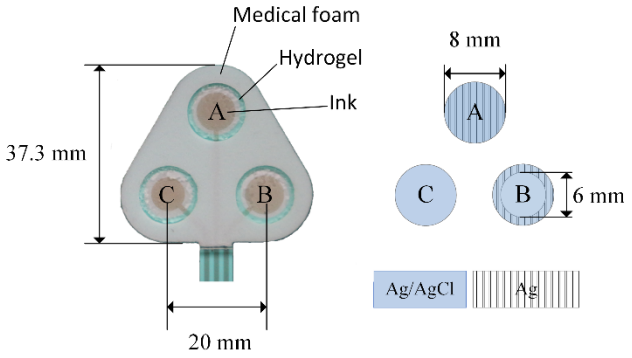

FIGURE 2. The test electrode set used in measurements comprising three (A, B, C) differently constructed electrodes (see details in text and Table 1).

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

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

late (PET) film (MacDermid Autotype Ltd, Wantage UK). The mesh count of the screen was 77 wires/cm^2, the wire diameter was 0.05 mm with a 40% open area fraction. First, conductive silver (Ag) ink was printed to form conduction traces and electrodes (Table 1). The Ag/AgCl based ink was then printed on the electrode area. The total thickness of the printed layers was 15 μm (C electrodes) and 30 μm (A and B electrodes) (Table 1 and Fig. 2). After printing, the electrodes were dried with infrared and air, and the conduction traces were encapsulated using printed insulation paste. Each electrode area was covered with an adhesive hydrogel membrane (AG602, Amgel Technologies, Fallbrock, CA, USA) to improve contact with the skin. Electrode attachment to the skin was ensured with non-conductive medical foam (Ventu-

FIGURE 3. The test electrode sets, worn by one of the authors (L.K.) are positioned on the forehead and behind the ear (mastoid). The electrode sets based on the inks of the other manufacturer are attached to the corresponding location on the opposite side of the head.

re 7432M, Venture Tape, MA, USA) which surrounds the hydrogel circles (Fig. 2). Each 3-electrode set consisted of A, B, and C electrodes (Table 1). Ag and Ag/AgCl inks from one manufacturer; PPG Industries Inc. (Pittsburgh, USA) (currently used in our facial electrode set [6]) or Engineered Conductive Materials ECM LLC (Delaware, USA) were used in each electrode set. Different electrode layer configurations A, B, and C (Table 1) were implemented to investigate the effect of the Ag/AgCl interface area size on skin impedances and the quality of recorded EEG signals and their possible alterations due to sweating.

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 ($p \geq 0.4$) 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).

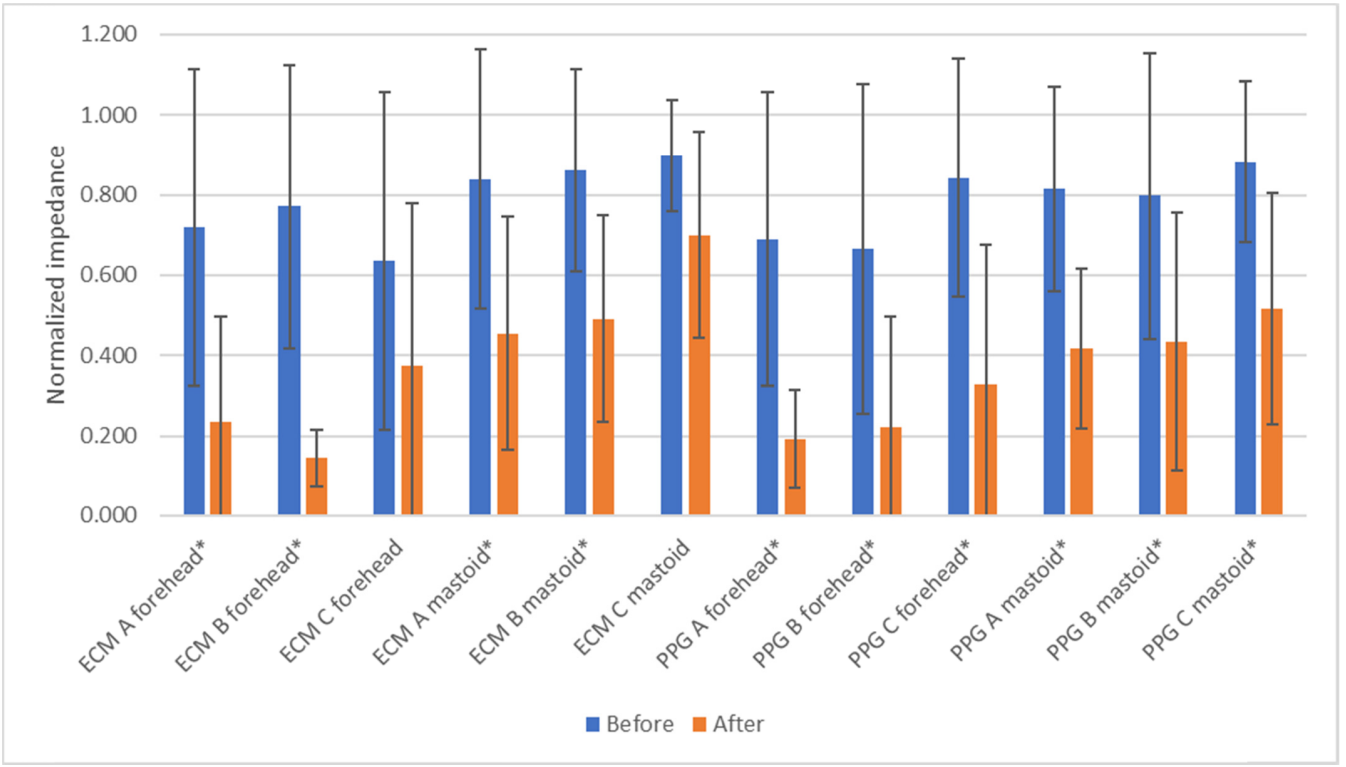

FIGURE 4. Normalized mean impedances of each electrode type before and after sweating. The error bars indicate standard deviations. The impedance ratio (after/before) of each electrode type: 0.33 (ECM A forehead), 0.19 (ECM B forehead), 0.58 (ECM C forehead), 0.54 (ECM A mastoid), 0.57 (ECM B mastoid), 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 ΔZ 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.

Test subject	ECM A		ECM B		ECM C		PPG A		PPG B		PPG C	
	ΔZ (Ω)	BP	ΔZ (Ω)	BP	ΔZ (Ω)	BP	ΔZ (Ω)	BP	ΔZ (Ω)	BP	ΔZ (Ω)	BP
1	130	1.1	150	0.9	150	1.3	120	1.5	160	1.3	N/A	1.0
2	35	1.0	120	1.0	110	0.8	10	1.1	N/A	1.0	N/A	1.0
3	40	1.4	30	2.7	100	1.1	60	N/A	80	1.6	10	1.2
4	90	1.4	100	1.7	N/A	1.6	60	1.5	100	1.8	60	2.7
5	40	1.2	70	1.4	100	0.8	60	1.5	10	2.3	80	1.3
6	50	1.4	100	1.2	60	1.3	60	2.1	80	1.7	70	1.7
7	40	0.9	60	1.5	40	0.4	70	1.3	40	1.1	30	0.8

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

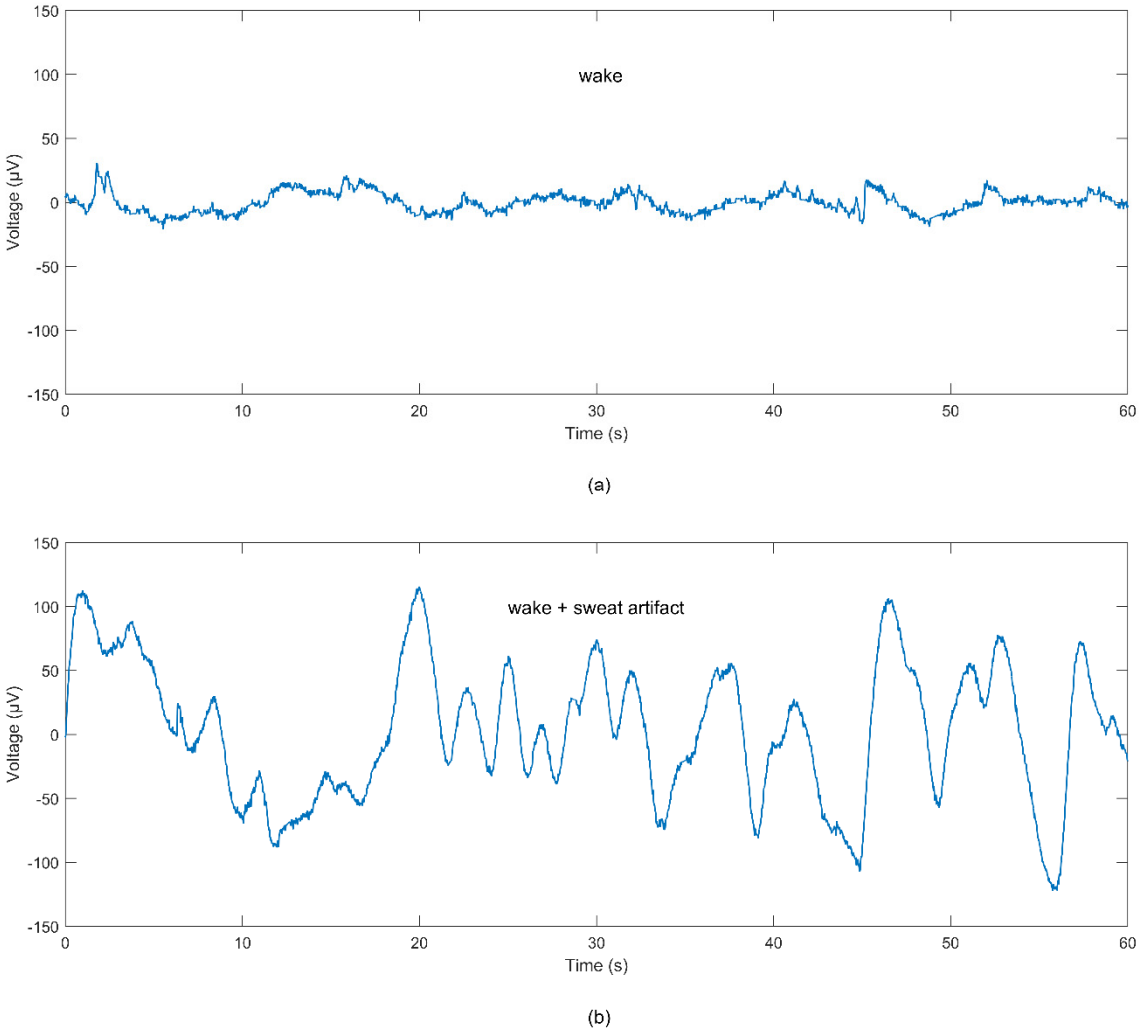

FIGURE 5. Examples of EEG signal measured with a PPG A electrode on the forehead before sweating (a) and after exercise-induced sweating (b) when low frequency, high amplitude signal oscillations are clearly visible.

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

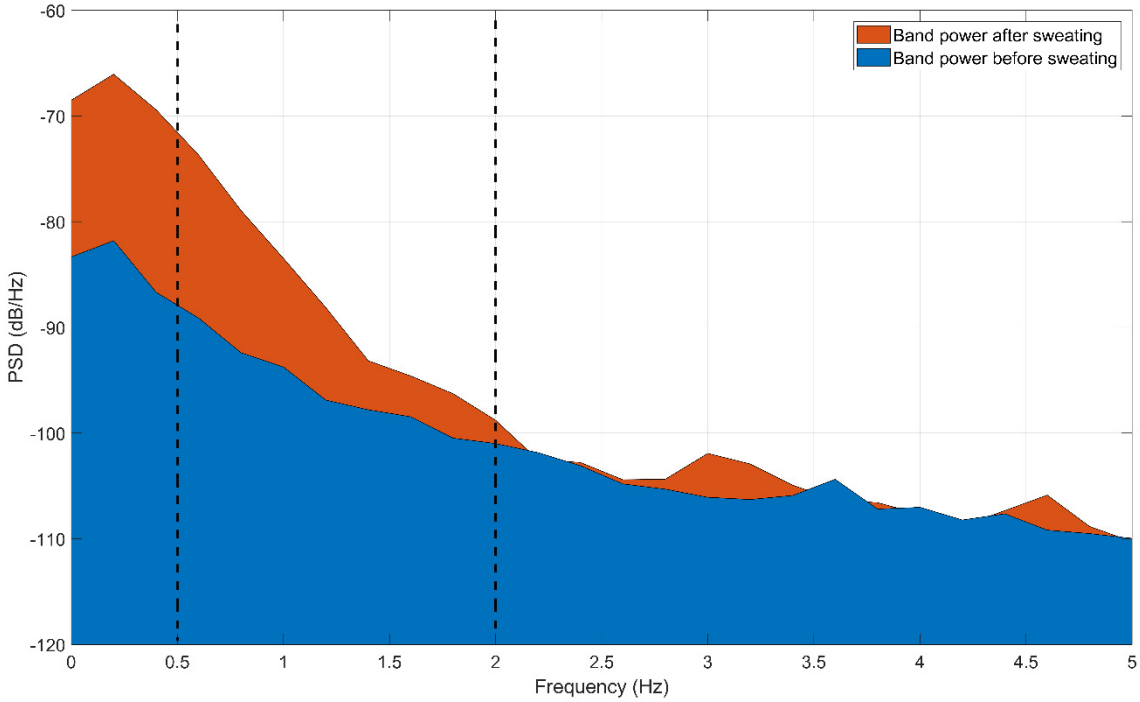

FIGURE 6. An example of the average band power calculation: to calculate the change of the band power due to sweating, the average band power after sweating (orange area) was divided by the average band power prior to sweating (blue area) at the frequency range from 0.5 to 2 Hz. This ratio value was calculated for each electrode type. Power spectral densities (PSDs) and the average band powers were calculated using Welch's periodogram and band power functions of Matlab.

more details, an impedance meter a 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 trial, 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

[1] R. B. Berry, C. L. Albertario, S. M. Harding, R. M. Lloyd, D. T. Plante, S. F. Quan, M. M. Troester, and B. V. Vaughn, *The AASM Manual for the Scoring of Sleep and Associated Events: Rules, Terminology and Technical Specifications, Version 2.5*. Darien, IL, USA, American Academy of Sleep Medicine, 2018.

[2] M. D. Ghegan, P. C. Angelos, A. C. Stonebraker, and M. B. Gillespie, "Laboratory versus portable sleep studies: A meta-analysis," *Laryngoscope*, vol. 116, no. 6, pp. 859–864, Jun. 2006.

[3] A. V. Benjafield, N. T. Ayas, P. R. Eastwood, R. Heinzer, M. S. M. Ip, M. J. Morrell, C. M. Nunez, S. R. Patel, T. Penzel, J.-L. Pépin, P. E. Peppard, S. Sinha, S. Tufik, K. Valentine, and A. Malhotra, "Estimation of the global prevalence and burden of obstructive sleep apnoea: A literature-based analysis," *Lancet Respiratory Med.*, vol. 7, no. 8, pp. 687–698, Aug. 2019.

[4] N. A. Collop, W. M. Anderson, B. Boehlecke, D. Claman, R. Goldberg, D. J. Gottlieb, D. Hudgel, M. Sateia, and R. Schwab, "Clinical guidelines for the use of unattended portable monitors in the diagnosis of obstructive sleep apnea in adult patients. portable monitoring task force of the american academy of sleep medicine," *J. Clin. sleep Med., JCSM, Off. Publication Amer. Acad. Sleep Med.*, vol. 3, no. 7, p. 737, Dec. 2007.

[5] E. S. Arnardottir, J. Verbraecken, M. Gonçalves, M. D. Gjerstad, L. Grote, F. J. Puertas, S. Mihaicuta, W. T. McNicholas, and L. Parrino, "Variability in recording and scoring of respiratory events during sleep in europe: A need for uniform standards," *J. Sleep Res.*, vol. 25, no. 2, pp. 144–157, Apr. 2016.

[6] T. Miettinen, K. Myllymaa, S. Westeren-Punnonen, J. Ahlberg, T. Hukkanen, J. Töyräs, R. Lappalainen, E. Mervaala, K. Sipilä, and S. Myllymaa, "Success rate and technical quality of home polysomnography with self-applicable electrode set in subjects with possible sleep bruxism," *IEEE J. Biomed. Health Inform.*, vol. 22, no. 4, pp. 1124–1132, Jul. 2018.

[7] G. Li, S. Wang, and Y. Y. Duan, "Towards gel-free electrodes: A systematic study of electrode-skin impedance," *Sens. Actuators B, Chem.*, vol. 241, pp. 1244–1255, Mar. 2017.

[8] P. Lepola, S. Myllymaa, J. Töyräs, A. Muraja-Murro, E. Mervaala, R. Lappalainen, and K. Myllymaa, "Screen-printed EEG electrode set for emergency use," *Sens. Actuators A, Phys.*, vol. 213, pp. 19–26, Jul. 2014.

[9] S. Myllymaa, P. Lepola, J. Töyräs, T. Hukkanen, E. Mervaala, R. Lappalainen, and K. Myllymaa, "New disposable forehead electrode set with excellent signal quality and imaging compatibility," *J. Neurosci. Methods*, vol. 215, no. 1, pp. 103–109, Apr. 2013.

[10] E. S. Arnardottir, B. Thorleifsdottir, E. Svanborg, I. Olafsson, and T. Gislason, "Sleep-related sweating in obstructive sleep apnoea: Association with sleep stages and blood pressure," *J. Sleep Res.*, vol. 19, no. 1p2, pp. 122–130, Mar. 2010.

[11] E. S. Arnardottir, C. Janson, E. Bjornsdottir, B. Benediktsdottir, S. Juliusson, S. T. Kuna, A. I. Pack, and T. Gislason, "Nocturnal sweating—A common symptom of obstructive sleep apnoea: The icelandic sleep apnoea cohort," *BMJ Open*, vol. 3, no. 5, May 2013, Art. no. e002795.

[12] S. Motamedi-Fakhr, M. Moshrefi-Torbati, M. Hill, C. M. Hill, and P. R. White, "Signal processing techniques applied to human sleep EEG signals—A review," *Biomed. Signal Process. Control*, vol. 10, pp. 21–33, Mar. 2014.

[13] E. S. Kappenman and S. J. Luck, "The effects of electrode impedance on data quality and statistical significance in ERP recordings," *Psychophysiology*, vol. 47, no. 5, p. 888, Sep. 2010.

[14] O. W. Tatum, *Atlas of Artifacts in Clinical Neurophysiology*. New York, NY, USA: Springer, 2019.

[15] L. A. Geddes, *Electrodes and the Measurement of Bioelectric Events*. New York, NY, USA: Wiley, 1972.

[16] L. Kalevo, T. Miettinen, A. Leino, S. Kainulainen, K. Myllymaa, J. Töyräs, T. Leppanen, and S. Myllymaa, "Improved sweat artifact tolerance of screen-printed EEG electrodes by material selection-comparison of electrochemical properties in artificial sweat," *IEEE Access*, vol. 7, pp. 133237–133247, 2019.

[17] E. Huigen, A. Peper, and C. A. Grimbergen, "Investigation into the origin of the noise of surface electrodes," *Med. Biol. Eng. Comput.*, vol. 40, no. 3, pp. 332–338, May 2002.

[18] J. Rosell, J. Colominas, P. Riu, R. Pallas-Areny, and J. G. Webster, "Skin impedance from 1 Hz to 1 MHz," *IEEE Trans. Biomed. Eng.*, vol. TBME-35, no. 8, pp. 649–651, Aug. 1988.

[19] M. S. Spach, R. C. Barr, J. W. Havstad, and E. C. Long, "Skin-electrode impedance and its effect on recording cardiac potentials," *Circulation*, vol. 34, no. 4, pp. 649–656, Oct. 1966.

[20] J. G. Webster, *Medical Instrumentation*. Boston, MA, USA: Houghton Mifflin, 1978.

[21] R. Kobayashi, Y. Koike, M. Hirayama, H. Ito, and G. Sobue, "Skin sympathetic nerve function during sleep—A study with effector responses," *Autonomic Neurosci., Basic Clin.*, vol. 103, nos. 1–2, pp. 121–126, Jan. 2003.

[22] J. J. S. Norton, D. S. Lee, J. W. Lee, W. Lee, O. Kwon, P. Won, S.-Y. Jung, H. Cheng, J.-W. Jeong, A. Akce, S. Umunna, I. Na, Y. H. Kwon, X.-Q. Wang, Z. Liu, U. Paik, Y. Huang, T. Bretl, W.-H. Yeo, and J. A. Rogers, "Soft, curved electrode systems capable of integration on the auricle as a persistent brain–computer interface," *Proc. Nat. Acad. Sci. USA*, vol. 112, no. 13, pp. 3920–3925, Mar. 2015.
[23] Y. Liu, M. Pharr, and G. A. Salvatore, "Lab-on-skin: A review of flexible and stretchable electronics for wearable health monitoring," *ACS Nano*, vol. 11, no. 10, pp. 9614–9635, Sep. 2017.
[24] W. Gao, H. Ota, D. Kiriya, K. Takei, and A. Javey, "Flexible electronics toward wearable sensing," *Accounts Chem. Res.*, vol. 52, no. 3, pp. 523–533, Feb. 2019.
[25] C.-T. Lin, L.-D. Liao, Y.-H. Liu, I.-J. Wang, B.-S. Lin, and J.-Y. Chang, "Novel dry polymer foam electrodes for long-term EEG measurement," *IEEE Trans. Biomed. Eng.*, vol. 58, no. 5, pp. 1200–1207, May 2011.

LAURA KALEVO received the M.Sc. (Tech.) degree in electrical engineering from the Tampere University of Technology, Tampere, Finland, in 2007. She joined the Sleep Technology and Analytics Group of the Kuopio University Hospital and the University of Eastern Finland (Kuopio, Finland) in 2018. She is currently working on her Ph.D.

From 2007 to 2018, she worked as a hardware designer and project manager in engineering companies. She works currently as a Junior Researcher in the Diagnostic Imaging Center, Kuopio University Hospital and the Department of Applied Physics, University of Eastern Finland. Her current research interests include the development EEG electrodes and the development of an easy-to-use ambulatory electrode set for home polysomnography.

TOMI MIETTINEN was born in Vehmersalmi, Finland, in 1987. He received the M.Sc. degree in medical physics from the University of Eastern Finland (Kuopio, Finland) in 2015 and a Ph.D. degree in 2019 with a thesis related to the suitability of an ambulatory electrode set for the home polysomnography based assessment of sleep bruxism.

From 2017, he has been employed in the Cancer Center of the Hospital District of Helsinki and Uusimaa (Helsinki, Finland) as a Specializing Medical Physicist. His current research interests include the development of an easy-to-use ambulatory electrode set for home polysomnography.

AKSELI LEINO received the M.Sc. (Tech.) degree in electrical engineering from the Tampere University of Technology, Tampere, Finland, in 2017. He is currently working on his Ph.D. degree in the Department of Applied Physics, University of Eastern Finland, Kuopio, Finland.

He has been a Junior Researcher in the Diagnostic Imaging Center, Kuopio University Hospital and the Department of Applied Physics, University of Eastern Finland since 2017. His research interests include the development of sleep apnea diagnostics with deep learning methods and investigating the relationship between stroke and sleep apnea.

SAMU KAINULAINEN (Joensuu, Finland, 7.9.1992) received the B.Sc. degree in applied physics and M.Sc. degree in medical physics from the University of Eastern Finland in 2017 and 2018. He is currently pursuing the Ph.D. degree in medical physics.

Since 2018, he has been a Junior Researcher in the Department of Applied Physics, University of Eastern Finland and in the Diagnostic Imaging Center, Kuopio University Hospital. His research interests include the development of signal analysis and deep learning methods in medical diagnostics, mainly in the diagnostics of obstructive sleep apnea.

HENRI KORKALAINEN received the B.Sc. and M.Sc. degrees in applied physics from the University of Eastern Finland in 2017 and 2018. He is currently pursuing the Ph.D. degree in medical physics.

Since 2018, he has been a Junior Researcher in the Department of Applied Physics, University of Eastern Finland and in the Diagnostic Imaging Center, Kuopio University Hospital. His research interests include the development of deep learning methods in medical diagnostics, mainly in the diagnostics of obstructive sleep apnea, and in improving of the diagnostic accuracy and severity assessment of obstructive sleep apnea.

KATJA MYLLYMAA was born in Haapajärvi, Finland, in 1981. She received a M.Sc. degree in medical physics from University of Kuopio, Finland in 2003, and a Ph.D. degree in medical physics from the University of Eastern Finland, Kuopio, Finland in 2010.

Currently she is working as a Medical Physicist in the Diagnostic Imaging Center in Kuopio University Hospital and holds a title of Docent in Medical Physics in the University of Eastern Finland. Recently she has been very active in investigating the relationship between stroke and sleep apnea as well as the development of novel methods for the diagnostics of sleep disorders.

JUHA TÖYRÄS was born in Rovaniemi, Finland, in 1975. He received the M.Sc. and Ph.D. degrees in medical physics from the University of Kuopio in 1999 and 2001, respectively.

He is currently a Professor of Biomedical Engineering in The University of Queensland, Brisbane, Australia, Professor of Medical Physics and Engineering in the Department of Applied Physics, University of Eastern Finland, Kuopio, Finland. He also serves as a Chief Physicist in the Diagnostic Imaging Center, Kuopio University Hospital. He is co-head of the Biophysics of Bone and Cartilage Research Group and co-head of Sleep Technology and Analytics Group in the Department of Applied Physics, University of Eastern Finland. His current research interests include the development of quantitative arthroscopy and computed tomography methods for use in diagnostics of osteoarthrosis, as well as in the development of artificial intelligence solutions for diagnostics and severity estimation of obstructive sleep apnea.

TIMO LEPPÄNEN received the B.Sc., M.Sc., and Ph.D. degrees in medical physics from the University of Eastern Finland in 2014, 2015, and 2016, respectively.

He started his studies on the University of Eastern Finland, in 2011, after two years of service with the Finnish Defense Forces. From 2012 to 2016, he was a Researcher in the Kuopio University Hospital. Since 2017, he has been a Postdoctoral Researcher and Co-Head of the Sleep Technology and Analytics Group, Department of Applied Physics, University of Eastern Finland. He was appointed as a senior researcher, in 2018. His research interests include the development of artificial intelligence solutions for diagnostics and severity estimation of sleep apnea, phenotyping and progression of obstructive sleep apnea, sleep apnea related psychomotor vigilance and daytime sleepiness, and development of wearable sensors for home sleep apnea testing.

Dr. Leppänen is a member of the European Sleep Research Society, the Finnish Sleep Research Society, the Finnish Society for Medical Physics and Medical Engineering, and the Finnish Society of Clinical Neurophysiology. He received the M.Sc. thesis award in 2016 and was shortlisted for Award of Young Scientist in University of Eastern Finland in 2017.

TIINA LAITINEN received the M.Sc. and Ph.D. degree in medical physics from University of Kuopio, Finland, in 1991 and 1998, respectively. She received in 1998 the degree of medical Physicist from University of Kuopio, Finland. Her Ph.D. research was concerned with arthroscopic measurement of knee joint cartilage stiffness.

She is currently the Chief Physicist in the Imaging Center, Kuopio University Hospital and Docent in University of Eastern Finland. Her current research activities include methodology of cardiovascular physiology, especially signal analysis, ultrasound imaging and nuclear medicine.

SAMI MYLLYMAA was born in Rauma, Finland, in 1975. He received his M.Sc. (Tech.) degree in electrical engineering from Tampere University of Technology, Finland in 2001, and the Ph.D. degree in biomedical engineering from the University of Eastern Finland, Kuopio, Finland in 2010.

He is currently acting as a Senior Researcher, Docent and Co-head of the Sleep Technology and Analytics Group in the Department of Applied Physics, University of Eastern Finland, and Researcher in the Diagnostic Imaging Center, Kuopio University Hospital. His current research interests include the development of novel wearable sensors and measurement techniques for personalized diagnostics and finding ways to improve the prognostics of sleep disorders.