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An In-Laboratory Comparison of FocusBand EEG Device and Textile Electrodes Against a Medical-Grade System and Wet Gel Electrodes

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ABSTRACT In this study, FocusBand (FocusBand Technologies, T 2 Green Pty Ltd.) textile electrodes were tested against medical-grade wet electrodes to investigate the possible differences in skin-electrode interface behavior and EEG signals' quality. *In vivo* electrical impedance spectroscopy and simultaneous forehead EEG measurements were performed with ten healthy subjects. In addition, the FocusBand device was tested in a stand-alone manner against a medical-grade reference system using similar test measurements. Compared to wet electrodes, textile electrodes had higher median absolute skin-electrode impedances five minutes after the attachment, but this difference decreased to 50–55 % of the initial during the first 60 minutes on all measured frequencies (1–1000 Hz). Textile and wet electrodes produced highly consistent EEG signals at forehead Fp1 and Fp2 locations. From those, Fp1 signals were more consistent in terms of normalized cross-correlations and agreement of the relative spectral powers. A stand-alone comparison showed that the FocusBand device can be used to record forehead biopotential signals, but the quality was not as consistent as with the medical-grade system. Based on impedance characteristics, a recording made using FocusBand textile electrodes may be more susceptible to artifacts than recording made using the medical-grade wet electrodes. However, FocusBand textile electrodes can be used, after a short stabilization period, to reliably record forehead EEG signals with a quality almost equal to that reached with medical-grade wet electrodes.

INDEX TERMS Dry electrode, electroencephalography, impedance, sleep, textile electrode, wearable.

I. INTRODUCTION

Electroencephalography (EEG) is the most common measurement technique for recording the electrical activity of the brain. In sleep studies, EEG is a compulsory part of polysomnography (PSG) to detect sleep stages and cortical arousals [1]. Despite recent technological advances, current medically approved EEG/PSG devices with accompanying sensor technologies still possess many disadvantages in terms of portability, costs, and ease of use, negatively affecting the availability of the measurement [2]. As sleep studies including EEG are mainly restricted to the laboratory environment, other disadvantages of PSG include a general deterioration in normal sleep structure and limitation to recording of one-night only [2]. As the demand for sleep studies is increasing due to growing awareness of sleep disorders [3], new

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wearable solutions for measuring EEG also outside the sleep laboratories are urgently needed.

An increasing number of consumer and research-grade EEG devices have been recently developed for cognitive studies, gaming, education, and brain-computer interfaces [4]. These devices show advantages in terms of self-usability and comfortability; however, according to recent reviews [4], [5], most of them are not properly validated for clinical or research use. Especially the data quality and susceptibility to various forms of noise and artifacts arising from the electrodes and hardware of these devices are regarded as major concerns [5]. The previous validation studies of novel EEG electrodes or devices have considered the accuracy and reliability of the systems with resting-state EEG measurements [6]–[8], blink detection ability [6], event-related potential measurements [9], and recording of the EEG during a driving task [6].

Consumer-grade EEG devices frequently apply dry electrodes because of their re-usability and ease of use. These electrodes are often fixed to a firm plastic casing [4], which might not be applicable in sleep studies due to discomfort and movement of the casing. Furthermore, conducting recordings at a hairy area is technically challenging; a simpler option is to use hairless areas, for example, the forehead. Forehead EEG measurement devices have been previously validated against the standard PSG with good accuracy in manual scoring of sleep stages [10]–[13]. Various types of dry EEG electrodes, e.g. flat type metal electrodes and stretchable conductive plastic electrodes, have been previously studied with electrochemical impedance spectroscopy and fitting of the equivalent circuits to compare the electrode characteristics to those of standard wet electrodes [14]–[17]. Furthermore, previous studies have reported an increase in the noise levels due to the higher skin-electrode impedances of dry electrodes [9], [14]. Other concerns regarding dry electrodes are their instability, arising from the susceptibility to movement artifacts, contact pressure, and sweating [16]–[19]. Conversely, devices with modern high input-impedance amplifiers are shown to be able to record high-quality EEG signals regardless of somewhat higher absolute skin-electrode impedances and impedance imbalance between the measuring electrodes [19], [20]. Thus, dry electrode configurations could potentially be utilized in high-quality EEG recordings e.g. in various sleep monitoring applications.

An interesting branch of dry electrodes is textile-based electrodes, which represent the convenient integration of the electrodes into garments [21], [22]. Most research on textile electrodes is conducted in relation to various applications of electrocardiogram or electromyogram [21]. Instead, the suitability of these electrodes as EEG electrodes has not been studied in a systematic, objective, and quantitative manner [21]–[23]. Despite this, the issues related to the manufacturing and material selection of these electrodes have already been reviewed [21], [22]. Silver has been previously used in textile electrodes because of its high conductivity and good biocompatibility [21]. In accordance, wet silver/silver chloride (Ag/AgCl) electrodes are the most commonly used electrodes in biopotential measurements from the skin. They exhibit nearly perfect nonpolarizable characteristics and were chosen as the gold standard reference in this study. We hypothesize that silver oxide-based textile electrodes can record forehead EEG signals with a quality close to that obtained with clinical electrodes when recorded with a modern high-quality biopotential measurement device. To investigate this hypothesis, we quantitatively evaluate the signal quality of a wearable headband (FocusBand Technologies, T 2 Green Pty Ltd., Windaroo, Australia), which includes three silver oxide textile electrodes, originally intended for neurofeedback-oriented sports training. First, we aim to investigate the possible differences in the skin-electrode interface behavior and forehead biopotential signal quality between FocusBand textile electrodes and medical-grade wet gel electrodes when all signals are simultaneously recorded with a medical-grade PSG device. To further evaluate the overall performance of the FocusBand EEG device, the second aim is to test the whole data acquisition chain, including the FocusBand textile electrodes, amplifier, and data logger against the chain consisting of medical-grade PSG device and wet gel electrodes.

II. MATERIALS AND METHODS

A. SUBJECTS

Ten healthy volunteers, six men, and four women, aged between 20 to 40 years participated in the present study. Volunteers were accepted to participate if they had no cardiovascular or respiratory diseases. The study protocol was given a favorable statement by The Research Ethics Committee of the Northern Savo Hospital District (849/2018) and all subjects assigned an informed consent form before the measurement. The study was conducted following the standard ethical guidelines of Good Clinical Practice and the Declaration of Helsinki.

B. ELECTRODES AND DEVICES

The FocusBand is a wearable headset with three silver oxide textile electrodes integrated on a neoprene band and designed to be used in the forehead region. The FocusBand textile electrodes cover the standard frontopolar 10-20 EEG system positions of Fp1, Fpz, and Fp2 (Fig. 1). To test the textile electrodes, we compared their performance *in vivo* against medically approved Ag/AgCl wet electrodes (Neuroline 720, Ambu A/S, Copenhagen, Denmark). Neuroline 720 electrodes are pre-gelled and self-adhesive and thus intended for long biopotential measurements from hairless skin. Details of the electrodes used in this study are given in Table 1.

When comparing only the electrode performance between the textile and wet electrodes, EEG signals from each electrode were recorded with a portable PSG/EEG device (Nox A1, Nox Medical, Reykjavik, Iceland). The sampling frequency of the Nox A1 device is 256 kHz. Signals are then downsampled and stored with a sampling frequency

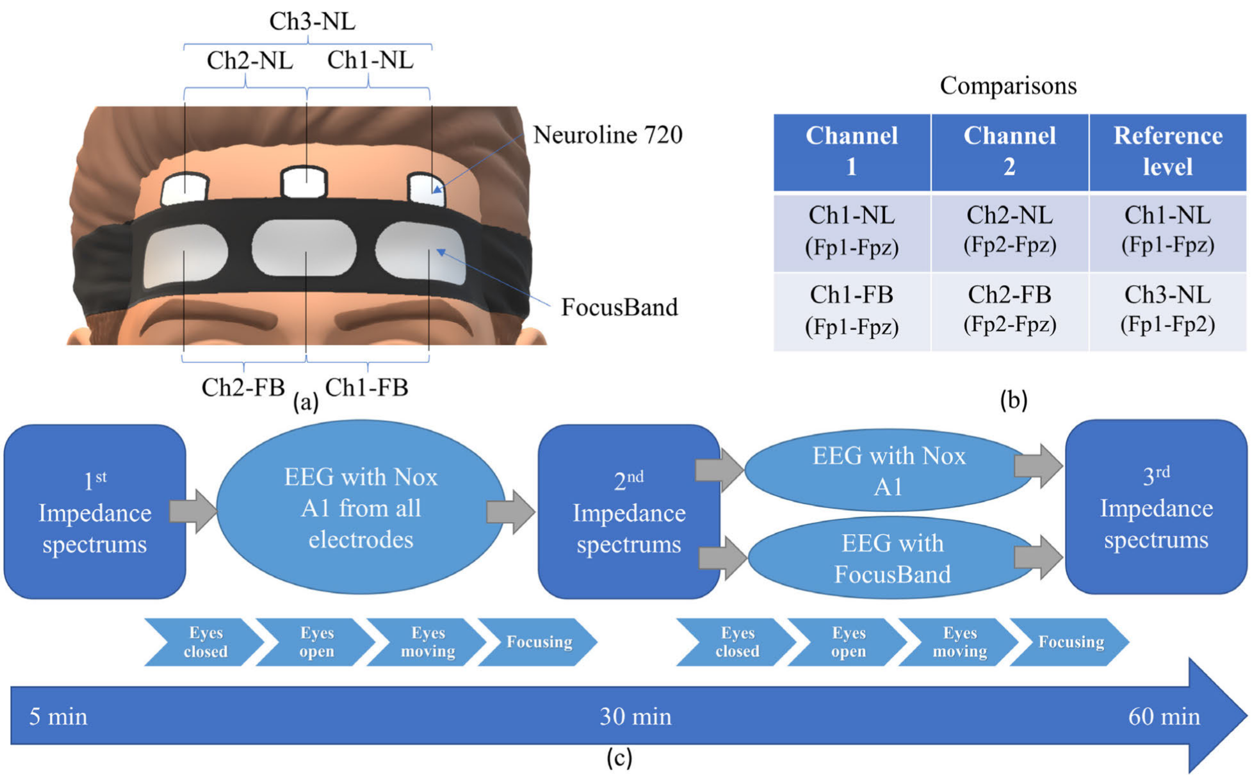

FIGURE 1. The sensor placement for simultaneous electroencephalography (EEG) recordings and impedance measurements and respective differential channels (a) of the electrodes that were used in comparisons (b). Below is an illustration of the measurement protocol (c) applied to each of the volunteers.

TABLE 1. Characteristics of the electrodes used in the study.

Manufacturer	Model	Type	Sensor Material	Electrode area (mm^2)	Other
Ambu A/S[a]	Neuroline 720	Wet/pre-gelled	Ag/AgCl	18 (95)[c]	Self-adhesive
FocusBand[b]	Sleep Band	Dry/textile	Silver-oxide	875	Reusable

[a]Copenhagen, Denmark
[b]T 2 Green Pty Ltd, Windaroo, Australia
[c]Sensor area (Gel area)

of 200 Hz. Besides the electrode comparison, the FocusBand device was tested as a stand-alone recording setup, and the gathered EEG data were compared with the data collected with a setup consisting of Nox A1 and Neuroline 720 electrodes.

FocusBand's recording device has a sampling frequency of 128 Hz and the raw signals are saved using a 100 Hz frequency through a Bluetooth connection on a developer's phone application (FocusBand WIZGO). Although the sampling frequencies of the compared devices are not identical, they are both appropriate to sample the most interesting content of the EEG signals in clinical sleep medicine, which is defined to be between 0.3 and 35 Hz [1].

PalmSens4 with a MUX8-R2 multiplexer (PalmSens, Houten, Netherlands) was used to measure the skin-electrode impedances from both types of electrodes as a function of logarithmically sampled frequency, ranging from 1 Hz to 1 kHz with 31 sample points. Fp1 and Fp2 electrodes were used as a working electrode separately and Fpz electrodes were used as a reference point of the measurement (Fig. 1). This allowed a comparison of the impedances simultaneously between textile and wet electrodes and between similar electrode types, i.e. computing the impedance imbalance. The multiplexer enabled the rapid acquisition of impedance spectrums of each electrode. The duration of a single impedance sweep for one channel was 36 seconds indicating a total measurement time of around two minutes for four electrodes. PSTrace 5.8 -software by PalmSens enabled an automatic injection current selection in a range of 10 nA to 1 mA set before measurement. The sinusoidal voltage applied to electrodes was 0.01 V in amplitude and had a zero DC offset.

C. STUDY PROTOCOL

Each volunteer was examined according to a measurement protocol and a sensor placement visualized in Fig. 1. This included the above-described skin-electrode impedance measurements at 5, 30, and 60 minutes after the attachment of the electrodes. EEG signals were acquired between the impedance measurements. First, the signals were measured simultaneously with both types of electrodes using the Nox A1 recording device. The non-measuring ground electrode of Nox A1 was attached to subjects' nasion during EEG acquisitions. Second, the FocusBand's recording device was used to measure the signals from the textile electrodes and as previously, the medical-grade reference was measured using Nox A1 device with the wet electrodes. All measurements were carried out in an air-conditioned room in a quiet environment.

The wet electrodes were attached to a volunteer's head as close above to the corresponding textile electrode as possible to achieve similar skin properties and biopotential activity areas (Fig. 1). The textile electrodes were attached to a volunteer's head without any skin preparation, whereas a standard preparation was applied for the wet electrodes; first, a gentle abrasion of the outermost skin layer, and secondly, cleansing with alcohol wipes. It was also ensured that no hair or eyebrows were left beneath the electrodes. The neoprene-based headband was tightened on a Velcro-type strap so that the volunteers felt comfortable wearing it.

Subjects followed a certain protocol during the EEG measurements; resting on supine position eyes closed for five minutes, resting eyes open for five minutes, moving their eyes up and down and right to left for 20 seconds, blinking eyes, and chewing/teeth-gritting for 20 seconds and finally, focusing on a given mental task for five minutes. These tasks were set to increase variability in the frequency content of the measured forehead signals, rather than expecting a simple neural response. Moreover, the forehead biopotentials are also influenced by the eye's electrical activity. The aforementioned tasks include different levels of eye movements enabling the investigation of these effects to signals of different electrode types. The task on the last part of the measurement was a Sudoku game on a smartphone and the volunteers were sitting on a bed when conducting it.

D. DATA ANALYSIS

The impedance data of each volunteer were exported from PSTrace 5.8 -software and imported to MATLAB R2017b (MathWorks Inc., Natick, Massachusetts, USA). The median, the maxima, and the minima of the absolute impedance values were computed. Similar computations were applied to the phase difference data to reveal more information on the electrochemical characteristics of the skin-electrode interfaces. In addition, the imbalance of the absolute impedances between similar types of electrodes on each volunteer was examined by calculating the absolute differences of the impedances between the two sides for wet electrodes and textile electrodes separately. Finally, the maxima, the minima, and the medians of the differences in absolute impedances at frequencies of 1, 16, 32, and 1000 Hz were computed.

The EEG data measured with Nox A1 was exported from the Noxturnal software (Nox Medical) and analyzed in MATLAB R2017b. Before any filtering, the raw signals were used to derive the FocusBand textile electrodes channel 1 (Ch1-FB; Fp1-Fpz) and channel 2 (Ch2-FB; Fp2-Fpz) by re-referencing. Similarly, channel 1 (Ch1-NL; Fp1-Fpz) and channel 2 (Ch2-NL; Fp2-Fpz) were derived to represent the signals of the medical-grade Neuroline wet electrodes. In addition, we derived an extra channel, Ch3-NL (Fp1-Fp2) (Fig. 1a), which was later used in the estimation of the reference level of the similarity metrics (Fig. 1b).

All EEG signals were filtered before the analysis with a 5th order type II Chebyshev 0.3–35 Hz bandpass filter with a 40 dB stopband attenuation from peak passband value. This filter was chosen to have a smooth passband and relatively fast transit between the pass- and stopband in the filtered signals frequency content. Signals were visually checked before and after the filtering in both, time and frequency domain, to ensure stable and rational signals for further quantitative analysis.

The similarity of the EEG signals was examined using three different analyses. First, the temporal overall correspondence of the signals was studied by computing the cross-correlation of the comparable signals utilizing zero lag normalization. Second, mean absolute errors (MAEs) of the signals in microvolts were computed to compare absolute differences in signal amplitudes. Third, relative powers of the typical EEG signal frequency bands; Delta 0.5–4 Hz, Theta 4–8 Hz, Alpha 8–14 Hz, and Beta 15–30 Hz frequencies, were computed in 30-second windows with Welch's method in a non-stationary manner. To study the correlations of the relative powers between different measurement setups, Pearson correlation coefficients were calculated. In addition, Bland-Altman plots and analyses were used to quantify the agreement of the relative powers between different signals.

All the analyses were done separately for Ch1 and Ch2 signals and the different stages of the measurement protocol; Eyes open, Eyes closed, Eyes moving (including blinking of the eyes and teeth-gritting), and Focusing (playing Sudoku game). The reference level of these similarity metrics was estimated by comparing signals that were recorded using Nox A1 and wet electrodes (Ch1-NL vs Ch3-NL). As these signals were acquired using the same medical-grade setup and had the same polarity, but slightly different electrode placement, they are thought to represent the effect of electrode displacement on the quantitative measures.

When comparing the FocusBand's device to the medical-grade reference setup in a stand-alone mode, the EEG signals, recorded with Nox A1, were downsampled to 100 Hz to match the FocusBand's signals as well as possible. Signals were temporarily synchronized by manual inspection of differentiable signal features. Due to the limitation of absolute time-synchronization, cross-correlations of the signals were not computed. Besides that, the same analyses of differences in signal amplitudes and relative powers were computed. All subjects were included in the analysis, and no artifacts were removed from the recordings. However, all signals were truncated to similar lengths, and 30 seconds from the beginning of the measurements were excluded to avoid comparing the signals during calibration of the devices.

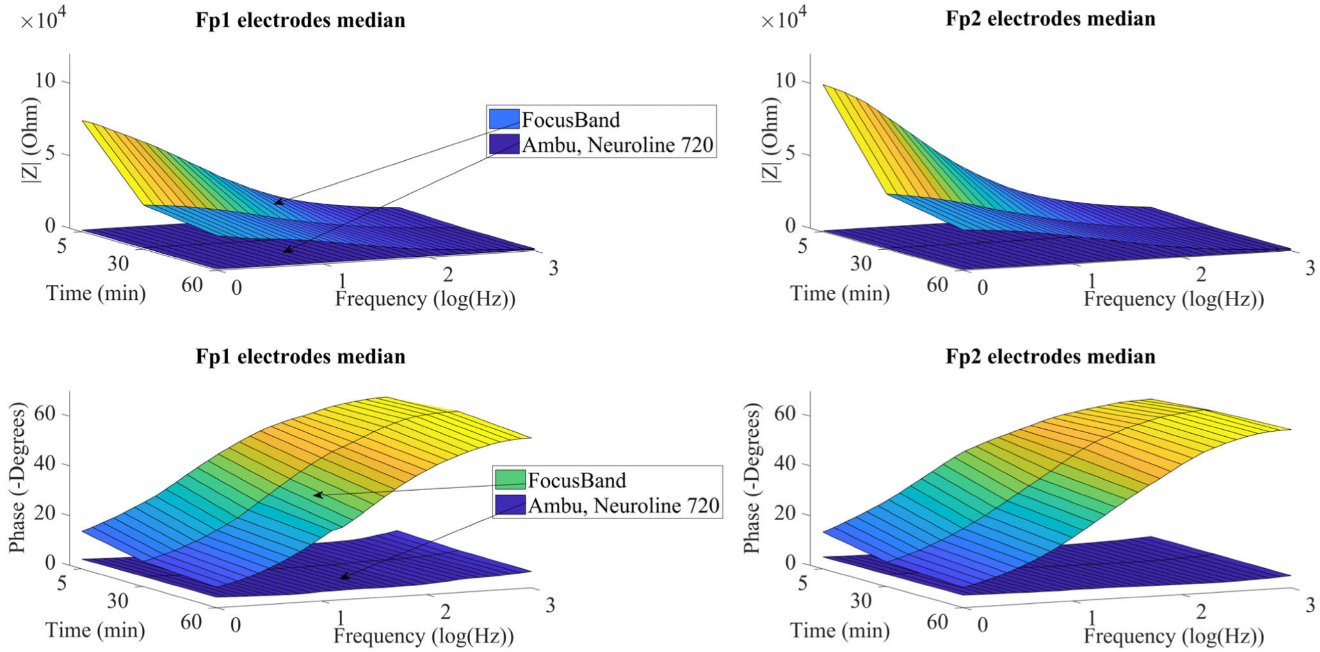

FIGURE 2. The median absolute skin-electrode impedances and phase shifts as a function of time and frequency. A major difference in the median absolute impedances between FocusBand textile electrodes and medical-grade wet electrode (Ambu A/S, Neuroline 720) was observed 5 minutes after the attachment on low frequencies. However, e.g. at the frequency of 16 Hz, the difference was reduced to 61 % after 30 minutes and further to 55 % after 60 minutes from the initial value. The FocusBand electrodes also produced a higher phase shift over the whole frequency range. Abbreviations: Fp1 and Fp2 indicate the frontopolar electrode of the impedance measurement and |Z| was the measured median absolute impedance.

III. RESULTS

A. SKIN-ELECTRODE IMPEDANCES

Clearly different impedance characteristics were observed between the textile and wet electrodes (Fig. 2). Median absolute impedances of the wet electrodes remained at a low (<1.1 kΩ) level at all measured frequencies and over the 60-min test. In contrast, the textile electrodes exhibited impedance spectrums that were highly dependent on the excitation frequency; the lower the frequency was, the higher the absolute impedances were. In addition, absolute impedances of textile electrodes systematically decreased with time in skin contact. At the first (5 min) measurement, the difference in median absolute impedances between compared electrode types was greatest (e.g. 26.4 kΩ vs 0.8 kΩ at 16 Hz), but this difference was reduced to 55–61 % (e.g. 16.8 kΩ vs 1.0 kΩ at 16 Hz) of the initial values after 30 minutes and further to 50–55 % (e.g. 14.9 kΩ vs 0.7 kΩ at 16 Hz) after 60 minutes (Table 2). Furthermore, the textile electrodes caused a greater negative phase shift that increased towards higher frequencies. However, there were no systematic changes in terms of phase shifts as a function of time.

Median imbalances in the absolute skin-electrode impedances between similar electrodes (Fp1 vs Fp2) were also higher for textile electrodes in the first (5 min) measurement (e.g. 11.1 kΩ vs 0.7 kΩ at 16 Hz). However, the median imbalance of textile electrodes reduced after 30 minutes to 30–56 % (e.g. 5.6 kΩ at 16 Hz) and after 60 minutes to 22–33 % (e.g. 3.7 kΩ at 16 Hz) of the initial values (Table 2). For wet electrodes, the median imbalance was initially low (<0.8 kΩ) and reduced after 60 minutes to 73–86 % (<0.6 kΩ) of the initial values.

B. SIGNAL QUALITY WITH FOCUSBAND TEXTILE ELECTRODES

The Ch1-FB and Ch1-NL signals were found to have a similar level of average cross-correlations (0.67–0.80) as the Ch1-NL and Ch3-NL signals (0.65–0.81) between the different stages of the measurement (Table 3). Compared to the reference level cross-correlations of Ch1-NL and Ch3-NL signals, the average cross-correlations between Ch2-FB and Ch2-NL signals were 0.14 units lower on average for all stages. Average cross-correlations were lower for Ch2 signals compared to Ch1 signals in all stages except the Eyes closed stage. The mean absolute error between the Ch1-FB and Ch1-NL signals was 5.07 μV over the whole recording, whereas between the Ch2-FB and Ch2-NL signals it was 4.59 μV. These values were close to the reference level of 3.88 μV and did not exhibit any major differences between the different stages of the protocol, except that the MAEs were greatest for all comparisons in the Eyes moving stage.

As a result of the relative band power analysis, a significant median correlation ($p < 0.05$) was found between the Ch1-FB and Ch1-NL signals on almost all frequencies and stages of the measurement (Table 4). Only Delta frequencies on Eyes closed and Eyes moving stages did not show a significant

TABLE 2. The medians and range (minimum-maximum) of the absolute impedances |Z| and impedance imbalance ΔZ between measuring electrode pairs (Fp1-Fp2) in Ohms among the studied population (n = 10).

Time	Frequency [Hz]	Neuroline 720 |Z| [Ω]	FocusBand |Z| [Ω]	Neuroline 720 ΔZ [Ω]	FocusBand ΔZ [Ω]
5 min	**1000**	674 (312–2235)	2041 (704–11482)	498 (53–1842)	656 (104–8759)
	32	756 (369–6576)	19686 (6088–67511)	734 (61–10666)	6895 (931–41034)
	16	773 (382–6929)	26457 (7130–98337)	738 (65–11840)	11065 (1138–51744)
	1	845 (433–7396)	48895 (9306–254572)	756 (84–12932)	31900 (1052–92548)
30 min	**1000**	821 (306–1798)	1569 (816–2179)	443 (35–1068)	273 (22–2114)
	32	958 (366–4485)	12580 (5987–25950)	635 (45–4153)	3875 (101–14556)
	16	977 (381–4690)	16753 (7242–37024)	639 (48–4449)	5554 (112–19039)
	1	1058 (441–4971)	30491 (10018–77796)	651 (64–4752)	9456 (711–38371)
60 min	**1000**	590 (305–2122)	1337 (664–2502)	428 (20–949)	225 (18–862)
	32	672 (386–5053)	10957 (5454–28144)	545 (25–2944)	2701 (446–20026)
	16	689 (409–5297)	14929 (7002–39153)	547 (28–3137)	3693 (230–29257)
	1	765 (488–5655)	24668 (10169–72459)	554 (42–3332)	6965 (598–47453)

TABLE 3. Mean maximum cross-correlations (±SD) of the filtered signals in different stages measured with Nox A1 from FocusBand (FB) textile and Ambu A/S Neuroline 720 (NL) electrodes.

Stage	Ch1-FB vs Ch1-NL	Ch2-FB vs Ch2-NL	Ch3-NL vs Ch1-NL
Eyes closed	0.67 ± 0.18	0.69 ± 0.16	0.81 ± 0.07
Eyes open	0.70 ± 0.17	0.50 ± 0.26	0.65 ± 0.14
Eyes moving	0.68 ± 0.29	0.54 ± 0.31	0.80 ± 0.06
Focusing	0.80 ± 0.18	0.71 ± 0.23	0.74 ± 0.06

Zero-lag correlations were taken as maximum values of the normalized cross-correlation.

median correlation between Ch1-FB and Ch1-NL signals. Median correlations of the relative powers were not statistically significant ($p > 0.05$) for Theta frequencies between the Ch2-FB and Ch2-NL signals. In addition, relative powers at high frequencies (Beta) on the Eyes moving stage and low frequencies (Delta) on Eyes open stage did not show a statistically significant median correlation ($p > 0.05$) between the Ch2 signals.

The Bland-Altman analysis revealed that signals recorded with the textile electrodes had an average of 8 % more power in Delta frequencies than the signals recorded with the wet electrodes (Fig. 3). Other frequency bands had highly similar powers between the signals recorded with the textile and the wet electrodes, as the mean absolute difference was lower than 3 % in all of them. Furthermore, the estimated 95 % confidence intervals were similar to the reference level comparisons.

C. SIGNAL QUALITY WITH FOCUSBAND DEVICE

When the FocusBand device was used to acquire the signals from textile electrodes, MAE between the Ch1-FB and Ch1-NL signals amplitudes was 74.4 μV over the whole recording, whereas between Ch2-FB and Ch2-NL it was 73.7 μV. The highest MAEs were found in the Eyes moving stage, being 193.9 μV and 183.4 μV for Ch1 and Ch2 comparisons, respectively. The reference level of the MAEs was computed similarly as before between the Ch3-NL and Ch1-NL signals and it was lower than 6 μV between all stages. Furthermore, when comparing Ch1-FB and Ch1-NL signals, relative powers at Alpha frequencies had a statistically significant median correlation ($p < 0.05$) within all stages of the measurement (Table 5). The main differences between FocusBand and Nox A1 measurements were in the low (Delta) and high (Beta) frequencies (Table 4). A similar finding was observed on the Bland-Altman plots (Fig. 4). An example of the signals measured during the focusing task is shown in Fig. 5.

IV. DISCUSSION

The main aim of this study was to investigate the electrical characteristics of FocusBand textile electrodes and to compare their performance to medical-grade EEG electrodes. Furthermore, we tested the FocusBand device as a stand-alone device against a medical-grade reference system to compare signal powers and qualities. Validation was conducted by means of *in vivo* measurements of the skin-electrode impedances and by simultaneous forehead EEG measurements. The results of the skin-electrode impedance measurements demonstrated clear differences in

TABLE 4. Median and ranges (minimum-maximum) of the Pearson correlation coefficients for the relative powers of frequency bands Delta 0.5–4 Hz, Theta 4–8 Hz, Alpha 8–14 Hz, and Beta 15–30 Hz in different stages of the measurement when the electrodes were compared.

Channel	Frequency	Eyes closed	Eyes open	Eyes moving	Focusing
Ch1-FB vs Ch1-NL	Delta	0.72 (0.03–0.96)	**0.76 (0.06–0.93)**	0.66 (-0.23–0.96)	**0.77 (-0.41–0.88)**
	Theta	**0.85 (0.32–0.99)**	**0.81 (0.08–0.98)**	**0.89 (0.07–0.97)**	**0.83 (0.15–0.99)**
	Alpha	**0.89 (0.67–0.93)**	**0.85 (0.61–0.99)**	**0.89 (0.32–0.97)**	**0.85 (0.26–0.98)**
	Beta	**0.77 (0.27–0.97)**	**0.82 (0.17–0.96)**	**0.81 (-0.06–0.93)**	**0.81 (0.18–0.98)**
Ch2-FB vs Ch2-NL	Delta	**0.74 (-0.42–0.97)**	0.67 (-0.01–0.94)	**0.80 (0.36–0.96)**	**0.78 (0.11–0.98)**
	Theta	0.62 (-0.55–0.95)	0.49 (-0.53–0.86)	0.58 (-0.65–0.95)	0.56 (-0.18–0.97)
	Alpha	**0.76 (0.37–0.96)**	**0.82 (0.46–0.96)**	**0.81 (0.33–0.97)**	**0.87 (0.29–0.97)**
	Beta	**0.82 (0.03–0.91)**	**0.78 (0.04–0.96)**	0.68 (0.27–0.90)	**0.72 (0.11–0.92)**
Ch1-NL vs Ch3-NL	Delta	**0.87 (0.55–0.97)**	**0.86 (-0.30–0.98)**	**0.83 (0.25–0.96)**	**0.90 (0.46–0.99)**
	Theta	**0.80 (0.21–0.99)**	**0.82 (-0.64–0.94)**	**0.79 (0.25–0.98)**	**0.84 (0.41–0.98)**
	Alpha	**0.89 (0.59–0.96)**	**0.92 (0.60–0.98)**	**0.86 (0.52–0.97)**	**0.88 (0.69–0.98)**
	Beta	**0.90 (0.60–0.97)**	**0.87 (0.79–0.97)**	**0.84 (0.41–0.97)**	**0.84 (0.30–0.98)**

FB refers to FocusBand electrodes and NL to Ambu A/S Neuroline 720 electrodes. Bold values indicate a statistically significant ($p < 0.05$) median correlation.

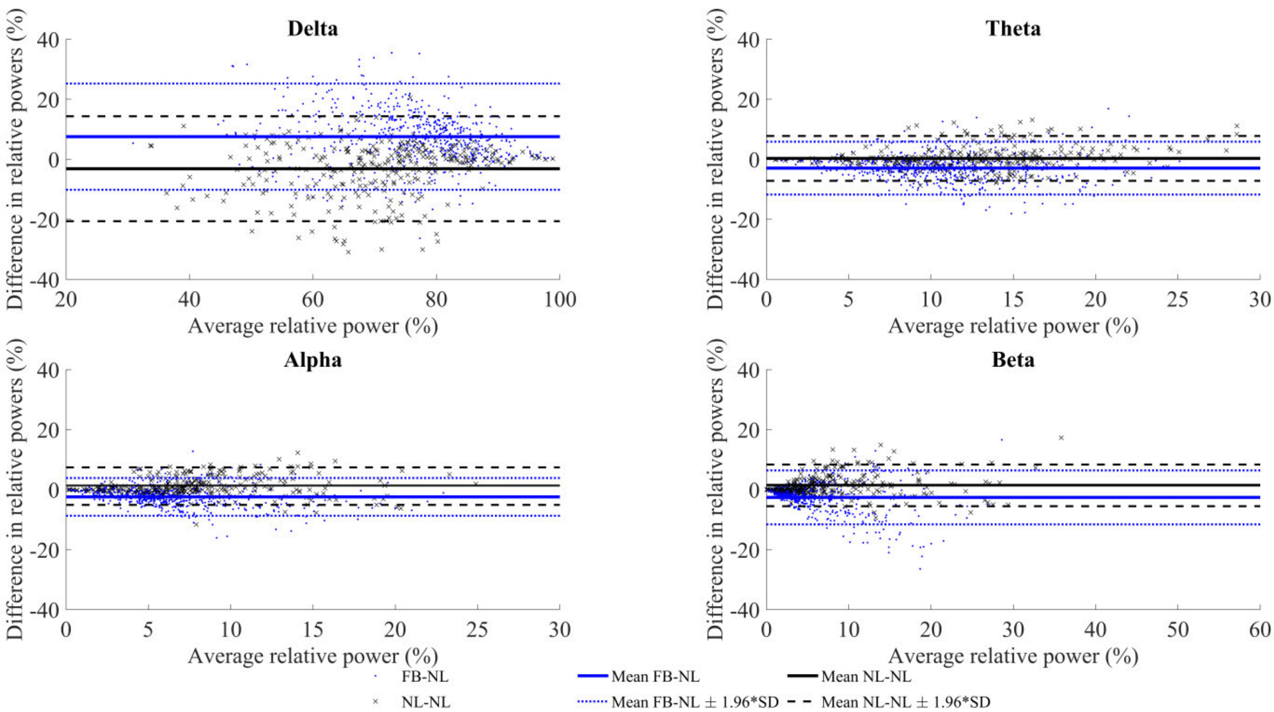

FIGURE 3. Bland-Altman plots of the relative powers between the compared signals in the setup used to compare electrode performance only. All signals were recorded with FocusBand (FB) textile electrodes and Neuroline 720 (NL) wet electrodes using a portable medical-grade PSG (Nox A1) device. Highly similar relative powers were observed between signals recorder with the textile and the wet electrodes in Theta (4–8 Hz), Alpha (8–14 Hz), and Beta (15–30 Hz) frequencies. Signals recorded with the textile electrodes had an average of 8 % more power in Delta (0.5–4 Hz) frequencies than signals recorded with the wet electrodes. As a reference, differences in Ch3-NL and Ch1-NL signals were also computed. The estimated 95 % confidence intervals for FB and NL differences were similar as in the reference level comparisons. Abbreviations: FB-NL refers to differences between Ch1-FB and Ch1-NL signals and between Ch2-FB and Ch2-NL signals. NL-NL refers to differences between Ch3-NL and Ch1-NL signals. SD refers to standard deviation of the respective differences.

TABLE 5. Median and ranges (minimum-maximum) of the Pearson correlation coefficients for the time-series of the relative powers of frequency bands Delta 0.5–4 Hz, Theta 4–8 Hz, Alpha 8–14 Hz, and Beta 15–30 Hz in different stages of the measurement when the FocusBand hardware was tested.

Channel	Frequency	Eyes closed	Eyes open	Eyes moving	Focusing
Ch1-FB vs Ch1-NL	Delta	0.03 (-0.53–0.80)	0.45 (-0.03–0.65)	0.26 (-0.50–0.75)	0.18 (-0.53–0.98)
	Theta	**0.76 (0.38–0.94)**	0.70 (-0.41–0.91)	0.66 (-0.14–0.86)	**0.71 (0.41–0.97)**
	Alpha	**0.87 (0.39–0.96)**	**0.86 (0.51–0.95)**	**0.85 (0.27–0.97)**	**0.81 (0.22–0.97)**
	Beta	0.39 (-0.35–0.89)	**0.75 (-0.09–0.92)**	0.42 (-0.24–0.89)	0.54 (-0.47–0.87)
Ch2-FB vs Ch2-NL	Delta	0.24 (-0.88–0.88)	0.34 (-0.52–0.89)	0.57 (-0.26–0.83)	0.04 (-0.60–0.99)
	Theta	**0.74 (0.19–0.84)**	**0.73 (0.15–0.92)**	0.51 (-0.03–0.67)	**0.74 (0.31–0.93)**
	Alpha	**0.73 (0.14–0.99)**	**0.85 (0.49–0.95)**	**0.79 (0.17–0.99)**	0.69 (-0.11–0.95)
	Beta	0.59 (-0.14–0.95)	**0.81 (0.51–0.95)**	0.58 (-0.03–0.97)	0.44 (-0.15–0.82)
Ch1-NL vs Ch3-NL	Delta	**0.89 (0.74–0.97)**	**0.82 (0.31–0.96)**	**0.90 (0.62–0.98)**	**0.93 (0.40–0.97)**
	Theta	**0.81 (0.54–0.96)**	**0.73 (0.47–0.94)**	**0.74 (0.46–0.98)**	**0.73 (-0.01–0.95)**
	Alpha	**0.89 (0.46–0.97)**	**0.87 (0.58–0.99)**	**0.82 (0.55–0.99)**	**0.82 (0.48–0.99)**
	Beta	**0.78 (0.05–0.97)**	**0.81 (0.60–0.92)**	**0.89 (-0.10–0.95)**	**0.88 (-0.10–0.99)**

FB refers to FocusBand hardware and NL to a setup consisting of Neuroline 720 (Ambu A/S) electrodes and the Nox A1 (Nox Medical) PSG/EEG device. Bold values indicate a statistically significant ($p < 0.05$) median correlation.

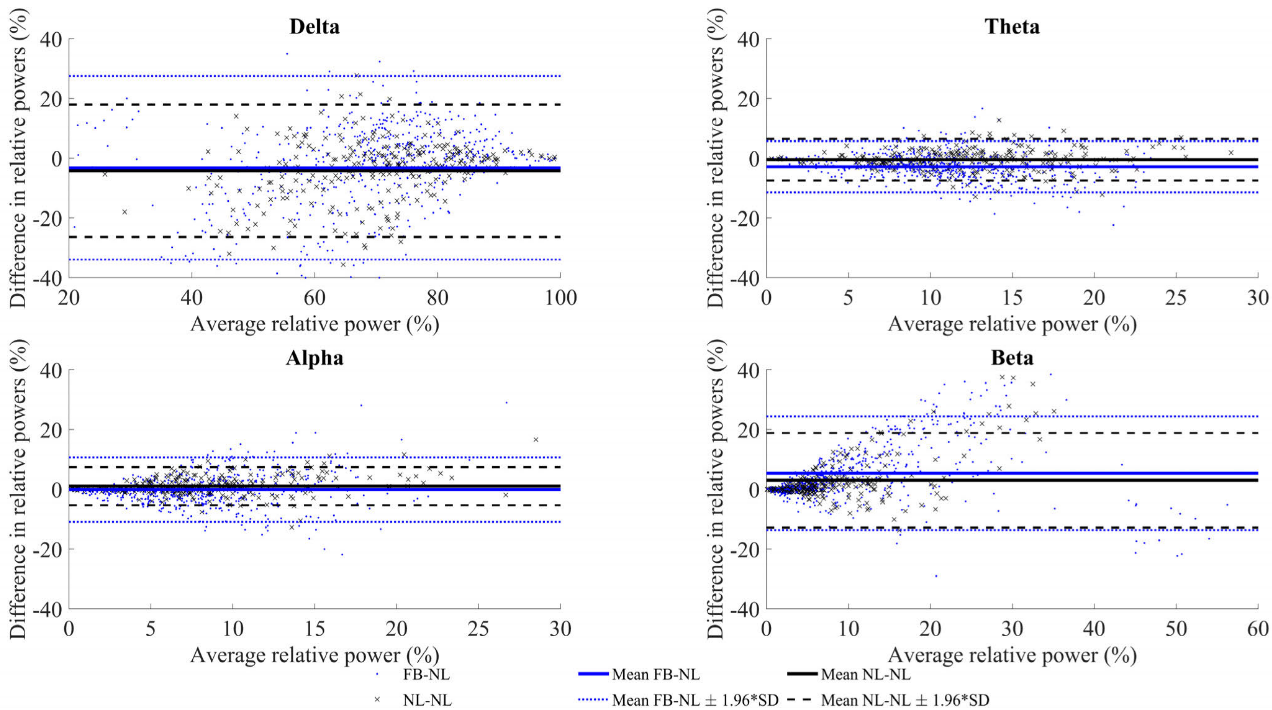

FIGURE 4. Bland-Altman plots of the relative powers between the compared signals in stand-alone comparison. The main differences between FocusBand (FB) device and Nox A1 (NL) measurements were in the low (Delta) and high (Beta) frequencies. As a reference, differences in Ch3-NL and Ch1-NL signals were also computed. The estimated 95 % confidence intervals showed a higher deviation of the differences between the FB and NL signals than between the signals measured with the medical-grade reference system. Abbreviations: FB-NL refers to computed differences between Ch1-FB and Ch1-NL signals and between Ch2-FB and Ch2-NL signals. NL-NL refers to differences between Ch3-NL and Ch1-NL signals. SD refers to standard deviation of the respective differences.

characteristics of the FocusBand textile electrodes compared to the medical-grade wet electrodes. However, the high initial difference (e.g. 26.5 kΩ vs 0.8 kΩ at 16 Hz) in the median absolute skin-electrode impedances was reduced to 50–55 % during the 60 minutes measurement. In addition, power spectral analysis performed on

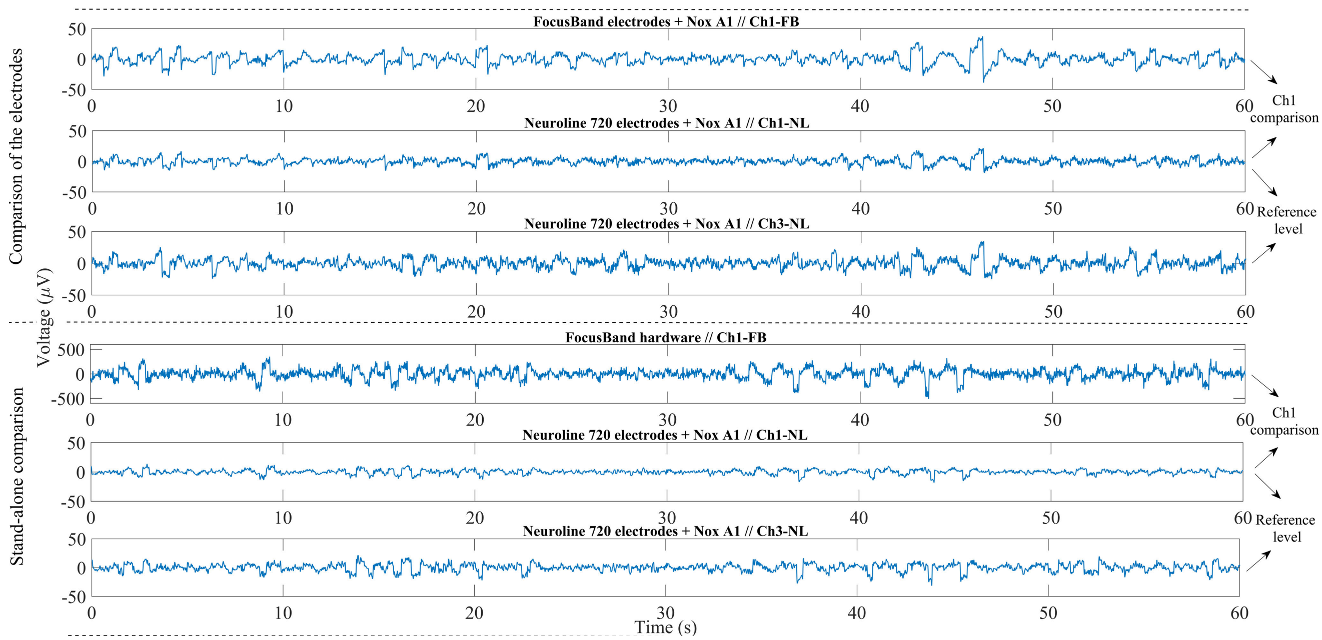

FIGURE 5. Examples of the filtered signals from Fp1-Fpz channel (Ch1) during the focusing task. The upper three traces show comparative signals recorded by a medical-grade PSG amplifier from FocusBand textile electrodes and from medical-grade Neuroline wet electrodes. The bottom three traces show a stand-alone comparison of the devices (FocusBand vs. medical-grade reference system). The signals from Fp1-Fp2 channel (Ch3) were used in the reference level estimation of the comparisons and are recorded using the medical-grade system. The normalized maximum cross-correlations of these exemplary signals were 0.83 and 0.73 for Channel 1 and reference level, respectively, in the electrode comparison.

simultaneously recorded EEG signals from textile electrodes and medical-grade wet electrodes revealed statistically significant correlations on multiple frequencies in different stages of the protocol. Bland-Altman plots of the relative powers verified the level of agreement between the compared signals. However, a slightly lower correlation was observed between the Ch2 signals than between the Ch1 signals. Furthermore, the results of the stand-alone comparison of the FocusBand device demonstrated its ability to record forehead biopotential signals, but the quality was not as consistent as that of the medical-grade reference system. It is well known that dry electrodes have higher absolute impedance values and are less stable than wet electrodes [14], [16]. However, the change in the absolute impedances as a function of time has not been systematically reported in the literature. Li *et al.* reported opposite results, as the absolute impedance of a dry (not textile) electrode was found to be increased over time [16]. Other variables known to affect the skin-electrode impedance measurements are room temperature, humidity, the pressure applied to the electrodes [18], [24], and the contact area of the electrode [14]. In the present study, the effects of room temperature and humidity were minimized by a constant measurement environment. The measurement room was well air-conditioned, which might be one reason for high absolute impedances during the first measurement before any sweat or moisture was formed between the electrodes and the skin. The absolute impedance values reported in the present study were at the same level as reported in the previous studies [14], [24], but are not easily comparable due to possible differences in the measurement devices and configurations. We did not compute the area-normalized impedance values, because these differences in the areas are part of the design of the electrodes we wanted to include in the comparison.

Based on the higher absolute impedances, the textile electrodes might be more susceptible to artifacts arising from the skin-electrode interface than the medical-grade wet electrodes. In addition, a high inter-subject variation in the absolute impedances (e.g. 7–98 kΩ at 16 Hz) of the textile electrodes five minutes after the attachment was revealed. This suggests that these electrodes might be more convenient and suitable for some particular skin types. However, the FocusBand textile electrodes systematically reach a lower level of absolute impedance (maximum $<$ 40 kΩ at 16 Hz) after 30 minutes of stabilization, when comparing to the fairly high initial level (maximum 98 kOhm). Furthermore, both of the studied electrodes maintained their electrical characteristics during the next 30 minutes on all subjects. Based on this, the electrical behavior of the skin-electrode interface of these textile electrodes could be suitable for long biopotential measurements where wet electrodes can lose their stability after gel drying. Thus, a study consisting of longer recording periods needs to be conducted to further assess these abilities and constraints of the electrodes.

EEG signals of Ch1-FB and Ch1-NL were similarly cross-correlated as Ch1-NL and Ch3-NL signals; maximum values at zero lag and the same level of correlation. However, Ch2 signals had lower average cross-correlation compared to Ch1 signals. This asymmetry of the results was observed in the Eyes open and Eyes moving stages, which might indicate that Ch2 signals were more susceptible to eye movement. On the other hand, the comparison of the relative spectral powers revealed that this asymmetry was focused on a single frequency range (Theta) and might be due to distorted measurement configurations where the measurement devices were mainly located on the Ch2 side of the volunteers. In fact, multiple factors may contribute to this asymmetry of the measures of the EEG signals. These include, for example, manufacturing differences regarding the electrodes, leads, and connectors and their susceptibility to electrical noise, light stimuli, or other physiological signals. In this study, the electrode leads, and the electrodes were not shielded. Despite the asymmetry, the present results show that FocusBand textile electrodes can be used to record the electrical activity of the forehead area with similar frequency content as recorded with wet electrodes. Although the forehead biopotential signals are referred to as EEG in this study, it must be noted that these signals are presumably not purely from a neural source as the eye's and facial muscle's electrical activity can easily contribute to the forehead signal. The EEG content was not extracted from these forehead signals as eye movement is valuable information in sleep studies. However, the signals were filtered to a frequency range of 0.5–35 Hz, which is the most interesting in sleep studies [1]. This filtering excludes the power line noise as well as some unwanted high-frequency artifacts and low-frequency drift.

The stand-alone comparison revealed the signals recorded with the FocusBand device had significantly higher amplitudes than the signals recorded using the medical-grade reference system. However, the visual comparison of the signals revealed that the signal waveforms were mainly similar, despite the amplitude differences. Few short segments of high-amplitude artifacts were visually found on all of the compared signals. These were presumably due to excessive eye movement or the volunteer's head movement during a resettling of the position before the focusing task.

There was a statistically significant correlation between the relative spectral powers of the signals measured with the FocusBand device and the reference system at Theta and Alpha frequencies during multiple stages of the measurement protocol. However, there was no statistically significant median correlation between the relative powers of these signals at the low frequencies (Delta) in any of the stages. Bland-Altman plots confirmed the level of agreement that was seen in the correlation analysis of the relative powers. The differences in relative powers between the FocusBand and medical-grade reference channels were more deviated than differences between the channels which were recorded using a similar medical-grade measurement setup, i.e. the reference comparison of Ch3-NL and Ch1-NL. The lack of consistency in an agreement between the FocusBand's and reference system's signals in stand-alone comparison was pronounced at low (Delta) frequencies. A similar finding was observed in a previous validation study of a wireless dry electrode system and was suggested to arise from differences in internal filtering, gain and other settings of the devices compared [25]. The present study also showed that injection signals with low frequencies produce greater absolute impedance differences between dry and wet electrodes, which may explain the detected differences in low-frequency powers. A common problem of portable biopotential measurement devices is that the device input impedances are not high enough, which leads to a relatively small common-mode rejection ratio, thus affecting the ability of the amplifier to resist common-mode signals.

Based on the results of the present stand-alone comparison, the FocusBand device may be more susceptible to artifacts than the medical-grade Nox A1 device. In addition, the high differences in Delta frequencies might be problematic in sleep recordings, especially concerning the detection of the N3 stage. In contrast, the FocusBand device performed relatively well on frequencies we believe it is originally designed for, i.e. Theta and Alpha. The differences at Beta frequencies might be due to a different susceptibility of the FocusBand's device to electromyographic activity, which can affect the high frequencies of EEG signals. Thus, the FocusBand device may not reach similar signal quality as the medical-grade device but may still be able to record EEG signals of mainly similar waveforms and relative powers of Theta and Alpha frequencies as a medically standard reference.

This study has certain limitations that must be discussed. First, the study population consisted of only ten healthy subjects, and thus the present conclusions can be generalized only with caution. However, as a technical comparison, this population was sufficiently large to quantitively reveal the differences in EEG signal quality and skin-electrode interface behavior of the compared electrodes and devices. Secondly, the sources of the artifacts affecting the similarity measures could not be identified based on the present results. As discussed earlier, low-power EEG signals are hard to record without any artifacts even with medical-grade devices. Acknowledging these challenges, the results showed a reasonable similarity, e.g. between Ch3-NL and Ch1-NL signals. Moreover, the measurement setup in electrode comparison may favor the wet electrodes as the patient ground was of this type of electrode also. However, we assume this has a minor effect on the results as modern devices can resist well against small impedance imbalances and the channels are re-referenced after the measurement to represent signals of a single electrode type. In addition, electrode placements could not be varied due to limited space on the forehead and because the headband was attached to a position, where it is originally intended to be used. This might cause a small difference in the susceptibility of the textile and wet electrodes to the eye's electrical activity as well as in the impedance measures. Finally, the signals of the stand-alone comparison could not be precisely temporarily synchronized due to different sampling frequencies, possible delays on the wireless connections, and timings of the compared devices. Following this, the cross-correlation analysis was missing in this part of the study. However, the two other analyses gave reasonable evidence on the similarity of the signals recorded using the FocusBand device and medical-grade reference system. Furthermore, in the present pilot study, only daytime measurements were conducted to eliminate various uncertain factors of unattended home sleep recordings. To evaluate whether the textile electrodes are suitable for sleep studies, overnight recordings need to be conducted in the future.

The present technical comparison supports the findings of a recently published home overnight validation of dry textile electrodes [26]. However, their protocol did not include recording of the standard EEG channels, determined by the American Academy of Sleep Medicine [1], to be used in sleep staging. Our next step is to investigate FocusBand textile electrodes in sleep recordings against a portable medical-grade EEG/PSG system.

V. CONCLUSION

The FocusBand textile electrodes can be used, after a short stabilization period, to reliably record forehead EEG. This conclusion is based on the revealed skin-electrode interface behavior and EEG signal quality of the textile electrodes when compared to a medical-grade reference. The FocusBand device performed relatively well on a frequency range of 4–14 Hz, but on other frequencies, it did not reach the quality of a medical-grade reference system. In addition, FocusBand textile electrodes are easy to use, reusable, and comfortable when integrated into a headband. They need no skin preparation, gel, or adhesive fixing. When all of these favorable characteristics of the textile electrodes are summed up with the results provided in this study, FocusBand textile electrodes make promises to be used in sleep recordings as well.

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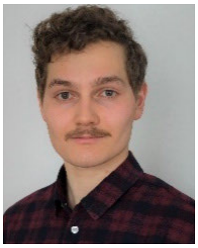

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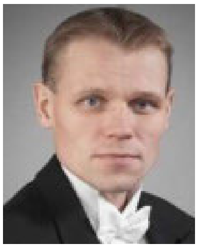

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