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Technical Performance of Textile-Based Dry Forehead Electrodes Compared With Medical-Grade Overnight Home Sleep Recordings

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This work involved human subjects or animals in its research. Approval of all ethical and experimental procedures and protocols was granted by the Research Ethics Committee of the Northern Savo Hospital District under Application No. 849/2018.

ABSTRACT The current clinically used electroencephalography (EEG) sensors are not self-applicable. This complicates the recording of the brain's electrical activity in unattended home polysomnography (PSG). When EEG is not recorded, the sleep architecture cannot be accurately determined, which decreases the accuracy of home-based diagnosis of sleep disorders. The aim of this study was to compare the technical performance of FocusBand, an easily applicable textile electrode headband, to that of clinical EEG and electrooculography (EOG) electrodes. Overnight unattended recordings were conducted at participants' ($n = 10$) homes. Signals were recorded using a portable Nox A1 PSG device. The FocusBand's forehead EEG (Fp1-Fp2) signals contained features that are visible at both, the standard EEG (F4-M1) and EOG (E1-M2) signals. The FocusBand's EEG signal amplitudes were significantly lower compared to standard EEG (F4-M1; average difference 98%) and EOG (E1-M2; average difference 29%) signals during all sleep stages. Despite the amplitude difference, forehead EEG signals displayed typical EEG characteristics related to certain sleep stages. However, the frequency content of the FocusBand-based signals was more similar to that of the standard EOG signals than that of standard EEG signals. The majority of the artifacts seen in the FocusBand signals were related to a loosened headband. High differences in the frequency content of the compared signals were also found during wakefulness, suggesting susceptibility of the textile electrodes to electrode movement artifacts. This study demonstrates that the forehead biopotential signals recorded using an easily attachable textile electrode headband could be useful in home-based sleep recordings.

INDEX TERMS Electroencephalography, electrooculography, textile electrode, wearables, sleep.

I. INTRODUCTION

Sleep is fundamentally a neural process that has an important role in physical and cognitive restoration [1]. Despite the crucial importance of sleep, it can be easily disturbed due to external factors, such as light, noise, medical conditions, or sleep disorders [2]. The most prevalent sleep disorders are insomnia and sleep apnea [3]; solely obstructive sleep apnea (OSA) affects nearly 1 billion adults globally [4]. In addition to the high prevalence, economic burden [5], and major health

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risks of OSA [6], most OSA subjects remain undiagnosed [7]. Thus, there is a need to enhance the current practices in sleep disorder diagnostics.

Traditionally sleep disorders have been diagnosed by conducting an attended polysomnography (PSG) or so-called Type I study in a sleep laboratory [8]. A similar recording can be conducted at the patient's home (Type II study), in which the electrodes are attached by healthcare professionals at the sleep laboratory [9], [10]. The preparation process for both of the recordings is rather laborious, inconvenient for patients, and the electrodes are still prone to detach, which may lead to failed recording and retesting [10]. Revised guidelines for unattended portable sleep studies have been established to increase the cost-effectiveness of sleep studies [11]. Especially a Type III study has become common practice in many clinical units [9], [10], [12]. The Type III study is conducted on a portable device with a limited number of channels (4-7), and without electroencephalography (EEG). This type of study is typically used in home sleep apnea testing (HSAT). However, possible comorbid sleep disorders affecting the sleep architecture cannot be diagnosed with a Type III study. This is mainly due to the lack of EEG, a vital part of the accurate detection of the sleep architecture and cortical arousals. In addition to EEG, the currently used manual sleep staging is based on the information of electrooculography (EOG) and submental electromyography (chin-EMG) [13]. The identification of sleep stages is crucial as many of the diagnostic parameters rely on total sleep time (TST) or other metrics derived from the sleep architecture [2]. In fact, OSA severity might also be underestimated with the current home-based recordings lacking EEG because of an overestimation of TST, diluting the OSA severity index per hour of sleep [14].

EEG is not generally measured in HSAT studies as the clinical EEG sensors are not suitable for self-application [12]. Current EEG techniques require complicated skin preparation and electrode mounting, conducted by a professional sleep technologist [8]. This could be resolved using a reduced number of easily attachable electrodes.

EEG recordings with wearable electrode sets and devices using a reduced number of electrodes on the forehead area have been proposed for sleep studies [15]–[18]. Wearable technologies tend to have reduced reliability for detection of sleep parameters, compared to those derived from PSG recordings [19]. In addition, the technical quality of wearable devices or electrodes has been of concern [20]. Furthermore, beyond-wearable devices, i.e. ear-EEG and temporary tattoo electrodes for long-term sleep monitoring purposes have been introduced [21], [22]. However, among the most user-friendly and easily applicable electrodes are textile-based dry electrodes [23]. These can be easily integrated into flexible and comfortable garments, which allows a solid fixation of the electrodes e.g. on the forehead area. Textile electrodes are in general reusable and can maintain electrical conductivity after multiple washing cycles [24].

Textile-electrodes have been already proposed for various EEG recording applications [23], such as monitoring newborns [25] and brain-computer interfaces [26]. Recently, self-adhesive microstructure textile-electrodes have been developed for electrocardiography and EEG recordings and for measuring micro-and macro-level EEG sleep features [27], [28]. Despite the promising results, it is still unclear how forehead EEG signals acquired with textile-electrodes correspond to standard EEG and EOG signal derivations recommended by the American Academy of Sleep Medicine (AASM) [13]. Especially the differences between standard signals referenced to mastoid versus the forehead signals, e.g. Fp1-Fp2, are yet to be studied in detail.

Based on an in-laboratory pilot study of the FocusBand's (T 2 Green Pty Ltd, Windaroo, Australia) textile electrodes [29], we hypothesize that the signal quality of the electrodes might be different from that of the standard EEG signals in a nocturnal sleep study. This is due to the higher skin-electrode impedance of the textile electrodes and the susceptibility of the textile electrode-based forehead signals to contain also more EOG activity than standard EEG signals. In the present study, we aimed to compare the information content and signal quality of the nocturnal forehead EEG signals recorded using the textile electrodes to those of the standard EEG and EOG signals used in clinical sleep medicine. Specifically, we aimed to investigate the differences in signal amplitudes, waveforms, and frequency content of the compared signals during different sleep stages.

II. METHODS

A. SLEEP RECORDINGS

Ten healthy volunteers (70% men, aged from 24 to 38 years old) underwent an unattended overnight home EEG recording in a natural sleeping environment. All subjects gave informed consent prior to recording. The study protocol was reviewed and approved by The Research Ethics Committee of The Northern Savo Hospital District (849/2018).

The home EEG recordings were simultaneously performed with medical-grade silver/silver chloride (Ag/AgCl) electrodes (Neuroline 720 and 726, Ambu A/S, Copenhagen, Denmark) and the FocusBand textile electrodes (Fig. 1). The measurements were conducted using a Nox A1 portable PSG device (Nox Medical, Reykjavik, Iceland). Medical-grade Ag/AgCl cup electrodes (Neuroline 726) were used to record EEG signals at locations F4, C4, O2 and for reference potential at M1 and M2. Self-adhesive medical-grade Ag/AgCl electrodes (Neuroline 720) were used as a ground and to record EOG and chin-EMG signals at standard locations. Moreover, a neoprene-based headband, the FocusBand, was connected to the same PSG system to simultaneously record forehead EEG signals with three integrated silver oxide-based textile electrodes. The contact area of a single textile electrode is 875 mm^2 and the headband was positioned to cover standard Fp1, Fp2, and Fpz locations.

Medical-grade electrodes were attached according to current clinical practices and detailed instructions from Kuopio University Hospital. The skin was prepared for cup electrodes

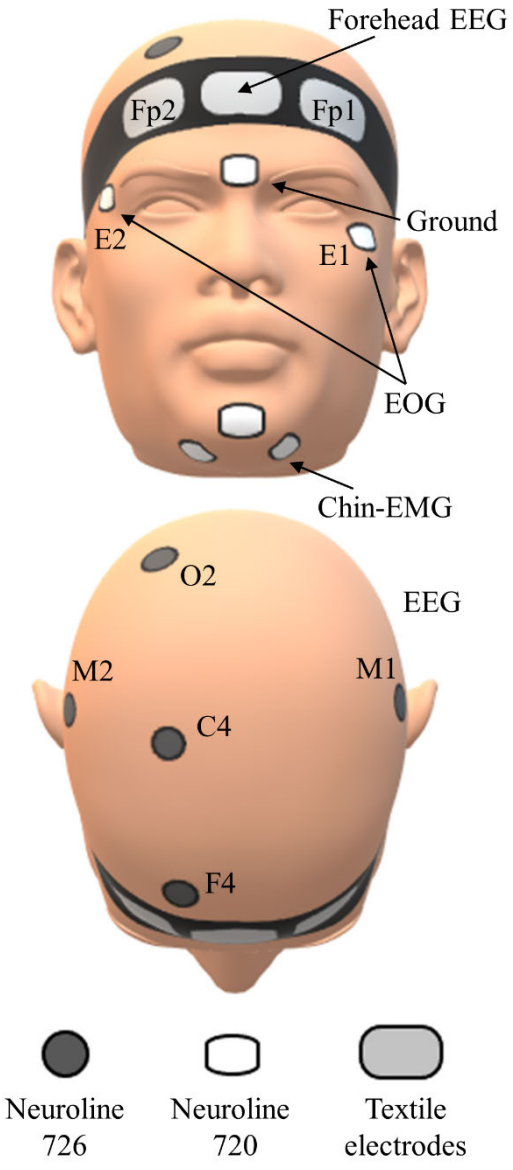

FIGURE 1. The electrode configurations used in the overnight EEG recordings. The Neuroline 726 electrodes (F4, C4, O2, M1, M2) were attached to the standard places in accordance with the 10-20 system. EEG = electroencephalography, EOG = electrooculography, EMG = electromyography.

using Nuprep skin prep gel (Weaver and Company, Aurora, CO, USA) and the cup electrodes were filled with Ten20 (Weaver and Company, Aurora, CO, USA) conductive paste. Furthermore, the cup electrodes were fixed with EC2 (Natus Medical Inc., Pleasanton, CA, USA) adhesive electrode cream and skin tape in the relevant locations, to attain reliable fixation. The skin was prepared for the self-adhesive clinical electrodes by gentle abrasion of the outermost skin layer and by cleansing with alcohol wipes. Skin preparation was not conducted for the FocusBand textile electrodes. Subjects were instructed to wear the headband so that it felt comfortable. Moreover, all electrodes were attached over 30 minutes prior to the start of the recording to enable stabilization of the skin-electrode interface [29]. The headband was cleaned with alcohol wipes after each recording. A new set of cup and self-adhesive clinical electrodes was applied for each subject.

The recordings were manually annotated by an expert sleep technologist at the Reykjavik University Sleep Institute. Sleep stages were identified based on the recorded standard PSG signals (i.e. EEG, EOG, and EMG), following the guidelines of the AASM manual, version 2.5 [13]. Noxturnal software (version 5.1, Nox Medical, Reykjavik, Iceland) was used for the annotation of the sleep stages and for visual inspection of the recorded signals.

Information related to annotated sleep stages is presented in Table 1. All recordings were considered successful, although a single subject reported that a cup electrode at the C4 location was detached during the recording night. In addition, another subject reported movement of the headband after loosening it during the night.

TABLE 1. Details of the sleep recordings ($n = 10$) presented as median (25-75% quartiles).

Total recording time (min)	449.4 (429.0–495.9)
Total sleep time (min)	390.5 (375.5–411.0)
Sleep latency (min)	16.5 (9.5–28.0)
Wake after sleep onset (min)	38.3 (26.0–48.0)
REM (%)	23.3 (22.1–25.8)
Stage N1 (%)	2.7 (1.3–3.6)
Stage N2 (%)	49.2 (43.6–51.8)
Stage N3 (%)	27.0 (21.9–32.3)
Sleep efficiency (%)	91.1 (85.3–93.1)
Arousal index (arousals/hour)	16.1 (14.3–21.1)

The relative durations of REM, N1, N2, and N3 sleep stages are presented as a percentage of the total sleep time. Sleep efficiency is the percentage of sleep during total recording time. Abbreviations: wakefulness (Wake), rapid eye movement sleep (REM), light sleep stage (N1), intermediate sleep stage (N2), and deep sleep (N3).

B. DATA ANALYSIS

All signals were recorded with a sampling rate of 256 kHz and stored with a rate of 200 Hz. The signals and the sleep stage annotations were exported from the Noxturnal software and imported to MATLAB (version 2019b, The MathWorks, Natick, MA, USA) for further analysis. The signals were first divided into 30-second epochs corresponding to the sleep staging. The epochs were further pooled based on the annotated sleep stage; wakefulness (Wake), rapid eye movement sleep (REM), light sleep (Stage N1), intermediate sleep (Stage N2), and deep sleep (Stage N3). This resulted a total of 1434 epochs of Stage Wake, 1854 epochs of Stage REM, 215 epochs of Stage N1, 3814 epochs of Stage N2, and 2020 epochs of Stage N3.

Preliminary investigations showed that textile electrode-based forehead signal (Fp1-Fp2) was most similar to E1-M2 and F4-M1 signals from all recorded EOG and EEG signal derivations. Based on the preliminary results, the Fp1-Fp2 signal recorded using the textile electrodes was chosen for a more detailed comparison against the standard EEG (F4-M1) and EOG (E1-M2) signals. These three signals were filtered

before amplitude and waveform analysis with a 0.3–35 Hz bandpass filter to exclude low-frequency drifting [13]. Filtering was conducted using a fifth-order type II Chebyshev filter with 40 dB stopband attenuation. For the amplitude analysis, the upper and lower envelopes of a signal were estimated with spline interpolation between the detected maximums and minimums. Furthermore, the mean of the upper and lower envelope was calculated to represent the upper and lower amplitude levels of the analyzed signal separately for each epoch. Relative differences between the compared signals' amplitude levels were calculated by subtracting the textile electrode signal's amplitudes from the wet electrode signal's amplitudes and dividing the result with the wet electrode signal's amplitudes. Finally, the distributions of the upper and lower amplitude levels between analyzed signals were compared in different sleep stages. Furthermore, the similarity between the compared signals' waveforms was quantified with dynamic time warping (DTW) [30]. Euclidean distance was chosen as the distance metric for the non-linear algorithm to preserve the units on voltage. Wilcoxon's signed-rank test was conducted to test the statistical significance of the differences in the amplitude and distance distributions. Due to the great number of epochs, a p-value less than 0.001 was considered to indicate statistical significance.

The information content and stability of the textile electrode signals were further compared with standard EEG signals utilizing frequency domain analysis. Power spectral densities (PSDs) for unfiltered 30-second signal patches were estimated using Welch's method with standard 50% overlap. All annotated epochs were included in the analysis, i.e. no segments with artifacts were removed. These frequency-domain segments were also pooled based on the annotated sleep stages. Time-synchronized difference spectrograms between the compared signals were computed in a subject-by-subject manner for different sleep stages to compare the overall correspondence of frequency content. This also allowed to evaluate the stability of the recording as non-stable segments with extensively high or low power spectral densities can be identified from the spectrograms. Moreover, we computed Magnitude-squared (MS) coherence estimates for the 30-second epochs to quantify how well the signals agreed on different frequencies. We computed the median coherences with interquartile ranges over all epochs divided into different sleep stages. The spectral estimation related to MS coherence estimates was conducted using Welch's method with 50% overlap and 0.1 Hz frequency resolution. To further investigate how a sleep-specific EEG feature appears in the signal recorded with the textile electrodes, we studied spindle coherence against the standard EEG and EOG signals recorded with the wet electrodes. Sleep spindles are defined as a train of distinct 11-16 Hz sinusoidal waves with duration over half a second. Most commonly the frequency of the sleep spindles is between 12 and 14 Hz. Therefore, spindle coherence was defined as the maximum coherence of the compared signals between 12 and 14 Hz and during the N2 sleep stage, where most of the spindle activity appears [13]. Statistical significance of the differences in spindle coherence distributions between textile electrode signals and standard wet electrode EOG or EEG signals was evaluated using Wilcoxon's signed-rank test.

III. RESULTS

A. AMPLITUDE COMPARISON

The Fp1-Fp2 signals recorded using textile electrodes had consistently lower ($p < 0.001$) amplitudes in all different sleep stages compared to the standard EEG (F4-M1) signals recorded with the clinical wet electrodes (Table 2, Fig. 2). However, the Fp1-Fp2 amplitudes were much closer

TABLE 2. Medians of upper and lower envelope amplitudes for filtered signals, recorded simultaneously with medical-grade wet electrodes (F4-M1 and E1-M2) and textile electrodes (Fp1-Fp2).

Stage	Lower envelope (μV)			Upper envelope (μV)		
	F4-M1	Fp1-Fp2	Difference (%)	F4-M1	Fp1-Fp2	Difference (%)
Wake	-16.7	-8.8	**90.3**	16.4	8.9	**84.4**
REM	-12.5	-5.6	**122.2**	12.0	5.6	**113.0**
N1	-13.0	-6.1	**114.3**	12.3	6.1	**101.1**
N2	-16.0	-8.0	**101.4**	15.4	8.0	**93.4**
N3	-22.3	-12.1	**83.9**	22.0	12.1	**81.9**
Average	-16.1	-8.1	101.8	15.6	8.2	94.3
	E1-M2	Fp1-Fp2	Difference (%)	E1-M2	Fp1-Fp2	Difference (%)
Wake	-12.3	-8.8	**40.2**	12.9	8.9	**45.4**
REM	-8.1	-5.6	**44.7**	8.0	5.6	**41.9**
N1	-8.0	-6.1	**31.7**	8.0	6.1	**29.9**
N2	-9.4	-8.0	**18.6**	9.4	8.0	**17.2**
N3	-13.3	-12.1	**9.6**	13.4	12.1	**10.6**
Average	-10.2	-8.1	28.5	10.3	8.2	28.6

Upper and lower envelope amplitudes were determined from each epoch by calculating the mean voltage of the envelope curve. Relative differences were calculated by subtracting the amplitudes of textile electrode signals from the wet electrode signal's amplitudes and by dividing the result with the wet electrode signal's amplitudes. Wilcoxon's signed rank test was used to test if the differences in the distributions of the amplitudes were statistically significant ($p < 0.001$, bolded typeface). Signals were bandpass (0.3 to 35 Hz) filtered prior to the analysis. Abbreviations: wakefulness (Wake), rapid eye movement sleep (REM), light sleep stage (N1), intermediate sleep stage (N2), and deep sleep (N3).

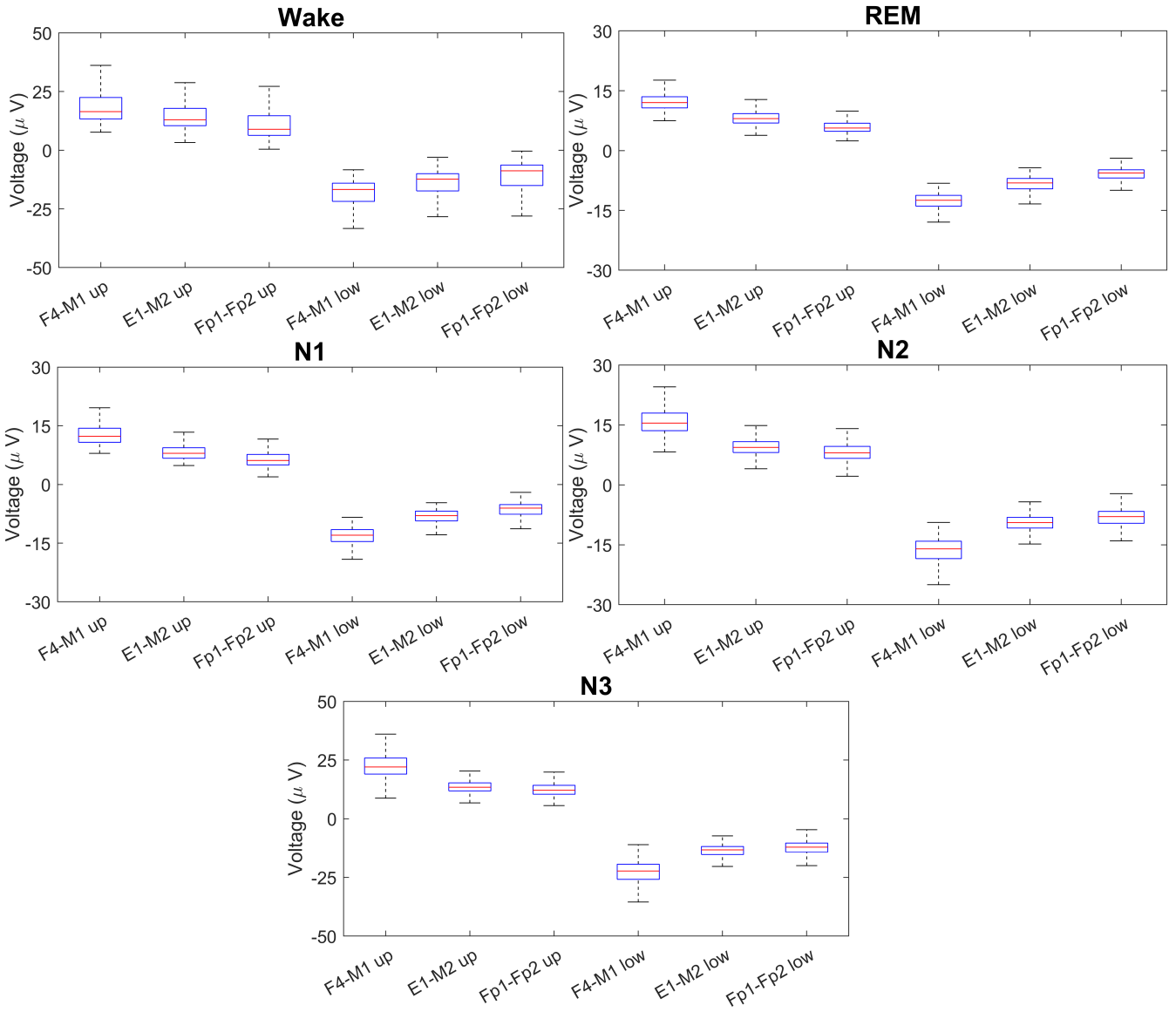

FIGURE 2. Box plot of the amplitude comparison between the standard polysomnography signals (F4-M1 and E1-M2) and signals recorded simultaneously with textile electrodes (Fp1-Fp2) during different sleep stages; wakefulness (Wake), rapid eye movement sleep (REM), light sleep (N1), intermediate sleep (N2), and deep sleep (N3). Amplitudes of the bandpass filtered (0.3–35 Hz) signals are estimated in an epoch-by-epoch manner as the mean of the lower (low) and upper (up) envelope curve of the signals. The red central line indicates the median and the box edges represent 25^{th} and 75^{th} percentiles. The whiskers indicate the most extreme data points which are not considered as outliers. Please note the different scaling of y-axis in the subplots.

to the amplitudes of EOG signals recorded with the clinical electrodes (E1-M2) albeit still being significantly lower ($p < 0.001$, Table 2). Especially in stages N2 and N3, the mean envelope amplitudes of Fp1-Fp2 signals were close to E1-M2 signals amplitudes. However, in stages Wake, REM, and N1 the amplitudes of E1-M2 signals were typically 30-45% higher than the amplitudes of signals recorded with the textile electrodes. The distributions of mean amplitudes of upper and lower envelopes showed consistent results with the findings observed using median values (Fig. 2, Table 2). An example of the recorded signals is presented in Figure 3, where the amplitude differences can also be seen.

B. SIGNAL WAVEFORM COMPARISON

Signals recorded with textile electrodes from the forehead area (Fp1-Fp2) displayed similar waveforms as both EOG (E1-M2) and EEG (F4-M1) (Fig. 3, Fig. 4). However, the DTW with Euclidean distance showed that EOG signals were significantly more similar ($p < 0.001$) to textile electrode signals compared to EEG signals during all sleep stages (Fig. 4). The median distance between the textile electrode signal and the standard EOG was significantly smaller ($p < 0.001$) in all sleep stages compared to the median distance between the textile electrode signal and the standard EEG.

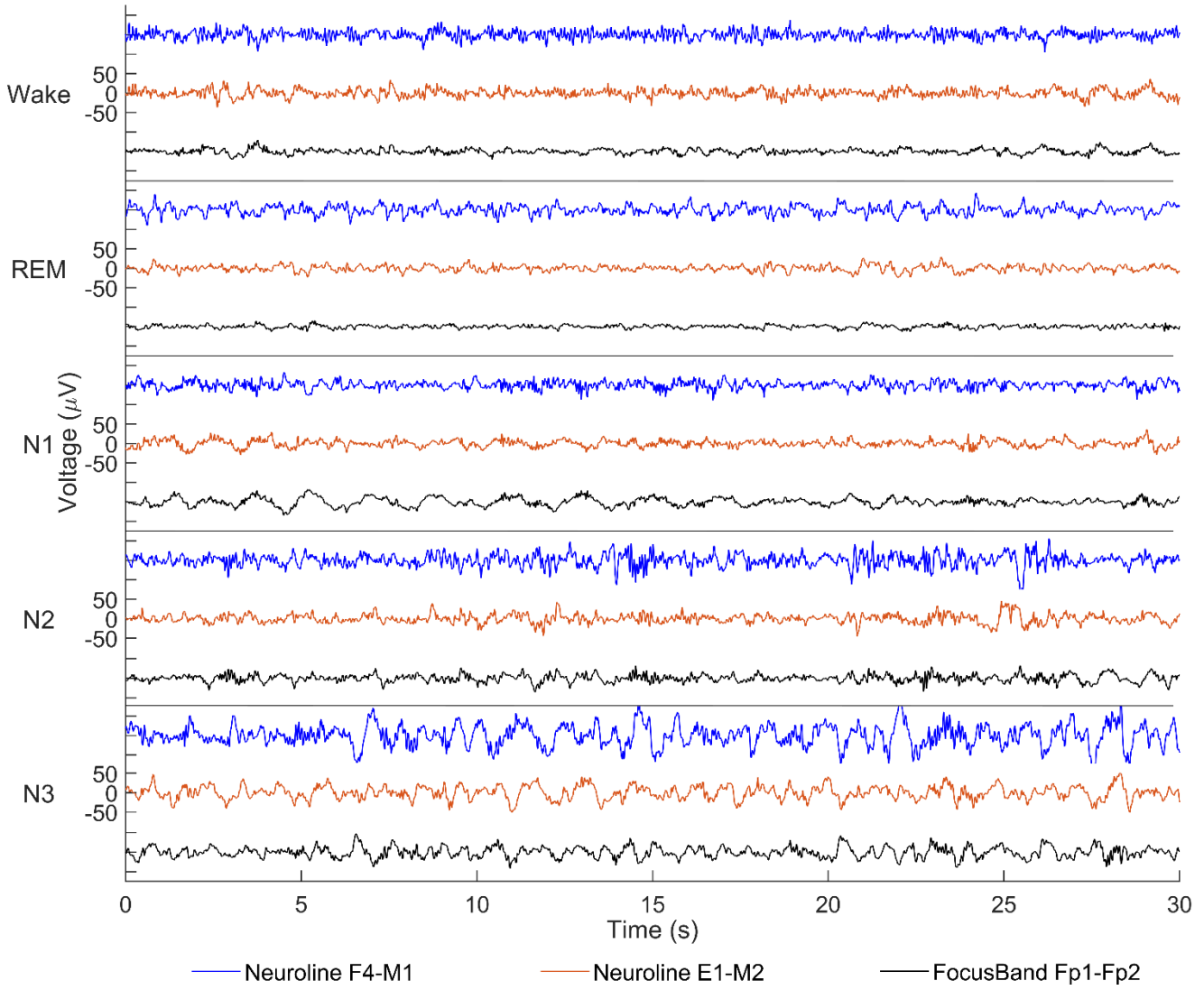

FIGURE 3. An example of the compared signals of volunteer #1 in different sleep stages; wakefulness (Wake), rapid eye movement sleep (REM), light sleep (N1), intermediate sleep (N2), and deep sleep (N3). All signals are recorded using the same medical-grade portable polysomnography device and bandpass (0.3 to 35 Hz) filtered. Similarities in the waveforms of the signals recorded using FocusBand textile electrodes (Fp1-Fp2) can be seen when compared with the waveforms of EEG (F4-M1) and EOG (E1-M2) signals recorded with medical-grade electrodes. Abbreviations: EEG = electroencephalography, EOG = electrooculography.

C. FREQUENCY CONTENT COMPARISON

Power spectral densities of the Fp1-Fp2 signals in different sleep stages were more consistent with those of the EOG signals (E1-M2) than those of the EEG signals (F4-M1) (Fig. 5, Fig. 6). This was seen especially on lower frequencies (<10 Hz) in the difference spectrograms (Fig. 5, Fig. 6). At higher frequencies, there were more differences in the powers of textile electrode signals and Neuroline electrode signals during wakefulness than other sleep stages (Fig. 5, Fig. 6).

Compared to the recorded EOG signals, the Fp1-Fp2 signals had a highly similar power content during N1 and N2 sleep stages and during deep sleep (N3) (Fig. 5). During REM sleep, there were more differences in the spectral powers of the compared signals on very low frequencies (<4 Hz) compared to other sleep stages (Fig. 5).

The Fp1-Fp2 signals showed systematically less spectral power than the standard EEG (F4-M1) signals. This was highlighted especially during N2 and N3 sleep stages (Fig. 6). The spectral power content of Fp1-Fp2 signals was most similar to F4-M1 signals power content during REM and N1 sleep stages. The difference spectrograms showed mainly positive differences between the compared signals' powers, except during the final segments of the Wake, REM, N1, and N2 sleep stages. However, this phenomenon was due to the reported movement of the headband by a single subject (indicated with a * marker in Fig. 5 and Fig. 6). Temporal variation in the agreement of the compared signals was

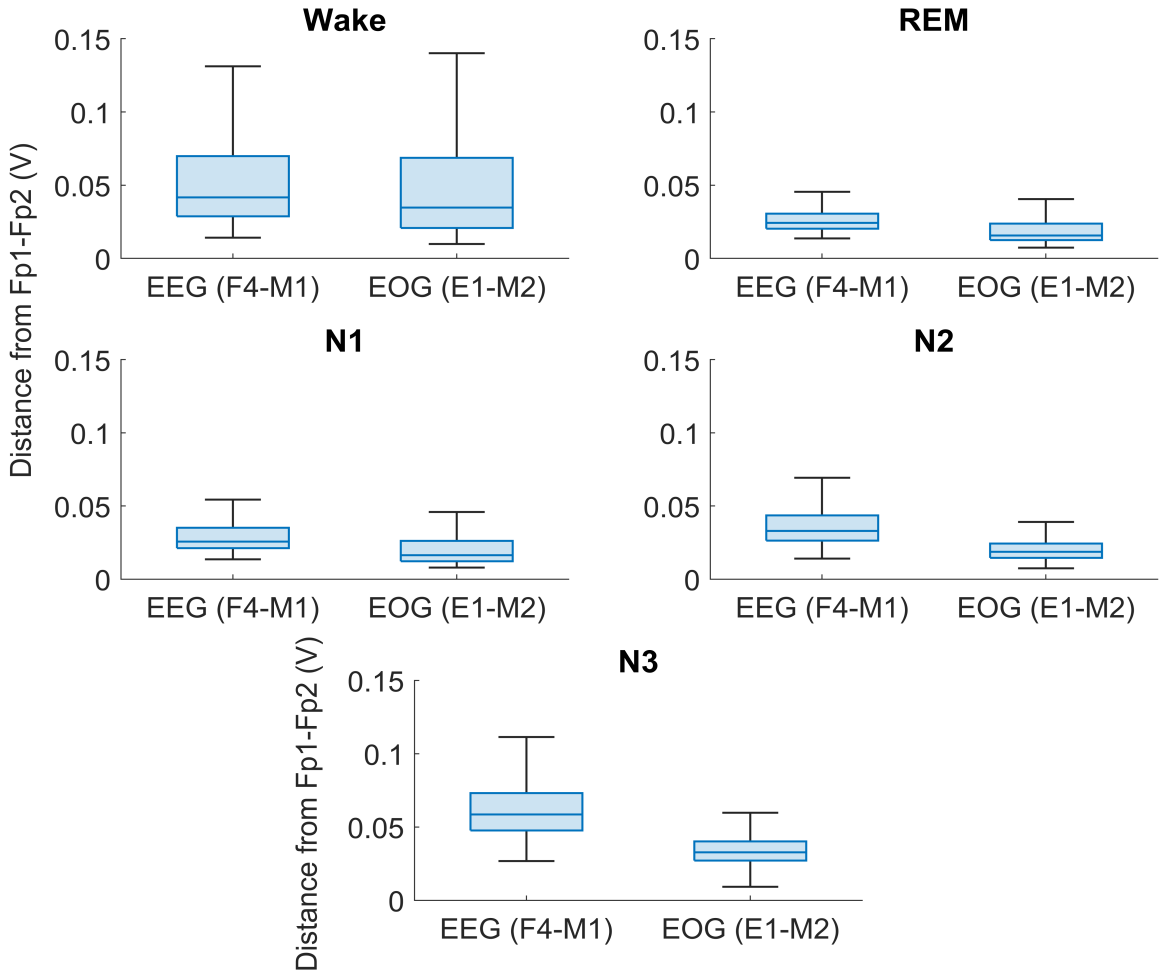

FIGURE 4. Box plot of the Euclidean distances in different sleep stages computed using dynamic time warping between textile electrode signal (Fp1-Fp2) and standard EEG (F4-M1) or EOG (E1-M2) signal. The median distance between the textile electrode signal and the standard EOG signal was significantly smaller ($p < 0.001$) in all sleep stages compared to median distance between the textile electrode signal and the standard EEG. The blue central line indicates the median and the box edges represent 25^{th} and 75^{th} percentiles. The whiskers indicate the most extreme data points which are not considered outliers. Abbreviations: EEG = electroencephalography, EOG = electrooculography, Wake = wakefulness, REM = rapid eye movement sleep, N1 = light sleep, N2 = intermediate sleep, N3 = deep sleep.

visible, even when the PSDs were grouped according to sleep stages (Fig. 5 and 6). The overall stability i.e., the presence of extensively high or low power spectral densities, was similar with both types of electrodes.

The coherence estimates of the 30-second signal epochs had extensive variance, but the overall coherence between the textile electrode signal and standard EEG signal was mainly similar to that between the textile electrode signal and standard EOG signal in different sleep stages (Fig. 7). However, there were statistically significant differences at specific frequencies in certain sleep stages. For example, at frequencies below 10 Hz during REM sleep, coherence between the textile electrode signal and the standard EOG signal was significantly ($p < 0.001$) higher than coherence between the textile electrode signal and the standard EEG signal. Conversely, the spindle coherence was significantly ($p < 0.001$) higher between the textile electrode forehead signal and standard EEG signal, when compared to spindle coherence between the textile electrode signal and standard EOG signal.

IV. DISCUSSION

In this study, we aimed to investigate one of the possible future solutions for self-applicable EEG measurements within a home environment. We studied the technical performance of a textile electrode-based headband that records forehead EEG signals with three integrated silver-oxide electrodes that do not require skin preparation or adhesive electrode mounting. The present study shows that the nocturnal forehead EEG signals recorded using the headband have a

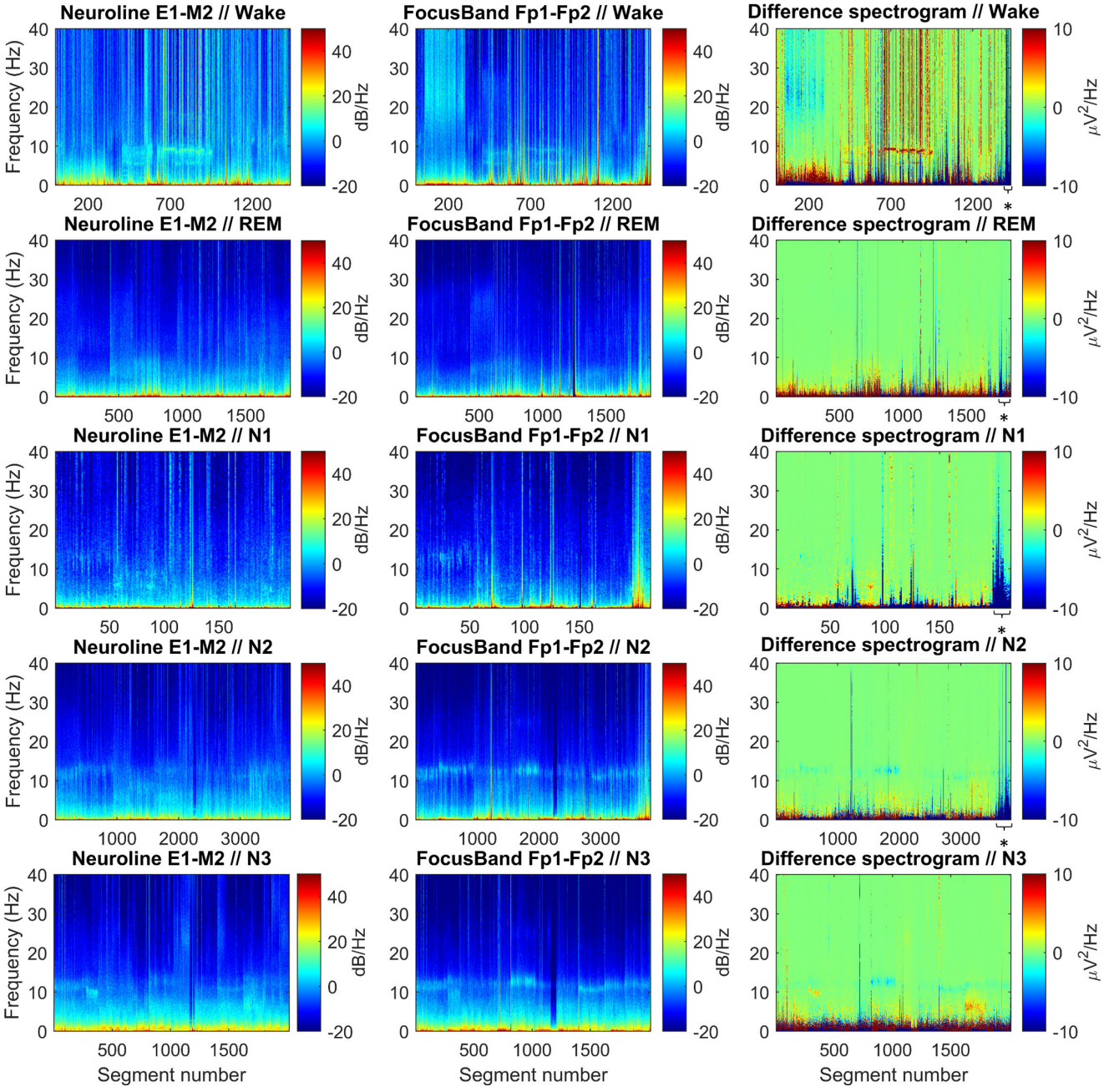

FIGURE 5. Spectrograms and difference spectrograms in different sleep stages for E1-M2 (Neuroline clinical wet electrodes) and Fp1-Fp2 (FocusBand's textile electrodes) signals. The difference spectrogram is computed by subtracting the textile electrode's signal powers from wet electrode's signal powers. A * marker indicates high differences in the power spectral densities resulted from a reported movement of the headband. Spectrograms are presented in units of dB/Hz (10 * $\log_{10}(\mu V^2/Hz)$) for visualization purposes, while the differences are calculated in units of $\mu V^2/Hz$. Abbreviations: Wake = wakefulness, REM = rapid eye movement sleep, N1 = light sleep, N2 = intermediate sleep, N3 = deep sleep.

high degree of similarities to standard EEG (F4-M1) and EOG (E1-M2) signals. One out of ten recordings (10%) partially (around 50% of the recorded night) failed due to the textile electrode headband. Similarly, one out of ten recordings (10%) failed due to a detached medical-grade cup electrode, but this was compensated with other electrodes. We found significant amplitude differences between the textile electrode-recorded signals and the standard PSG signals, but importantly, also similarities in signal waveforms and spectral content. Especially the F4-M1 signals had notably higher amplitudes than the signals obtained using the headband (Fp1-Fp2). Albeit the headband signals contained features of both EEG and EOG, the amplitudes, and the frequency content of the textile electrode-recorded signals were more consistent with the EOG (E1-M2) than with the EEG (F4-M1) signals. Furthermore, the spectrogram analysis

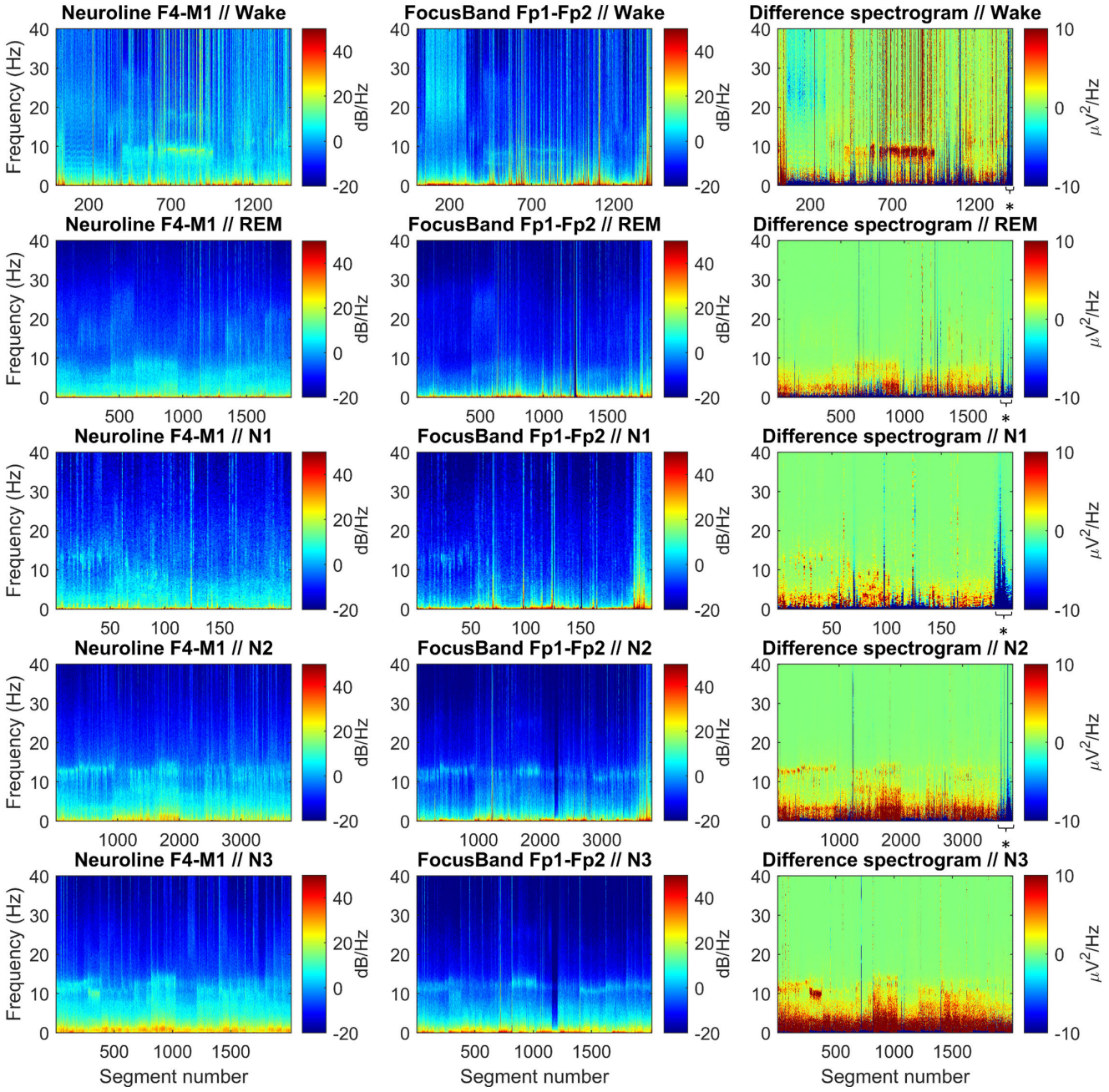

FIGURE 6. Spectrograms and difference spectrograms in different sleep stages for F4-M1 (Neuroline clinical wet electrodes) and Fp1-Fp2 (FocusBand's textile electrodes) signals. The difference spectrogram is computed by subtracting the textile electrode-based signal powers from wet electrode-based signal powers. A * marker indicates high differences in the power spectral densities resulted from a reported movement of the headband. Spectrograms are presented in units of dB/Hz (10 * $\log_{10}(\mu V^2/Hz)$) for visualization purposes, while the differences are calculated in units of $\mu V^2/Hz$. Abbreviations: Wake = wakefulness, REM = rapid eye movement sleep, N1 = light sleep, N2 = intermediate sleep, N3 = deep sleep.

showed good overall stability of the signals recorded with textile electrodes as well as the signals recorded with the standard clinical-grade Neuroline electrodes. As an exception, one of the subjects had a loosely fixed headband and reported movement of the electrodes, which was also seen in the spectrogram analysis as exceptionally high power spectral densities. Overall, these results suggest that the textile electrode-based headband has potential in home-based, patient-centered sleep disorder screening as an easily applicable EEG recording tool.

The amplitudes of the EEG signals recorded with FocusBand electrodes were consistently lower than those recorded with clinical electrodes. This difference most probably results from a different referencing of the signals. The standard signals are recorded against the electrical inactivity at the mastoid, whereas the textile electrode-based signal is fully

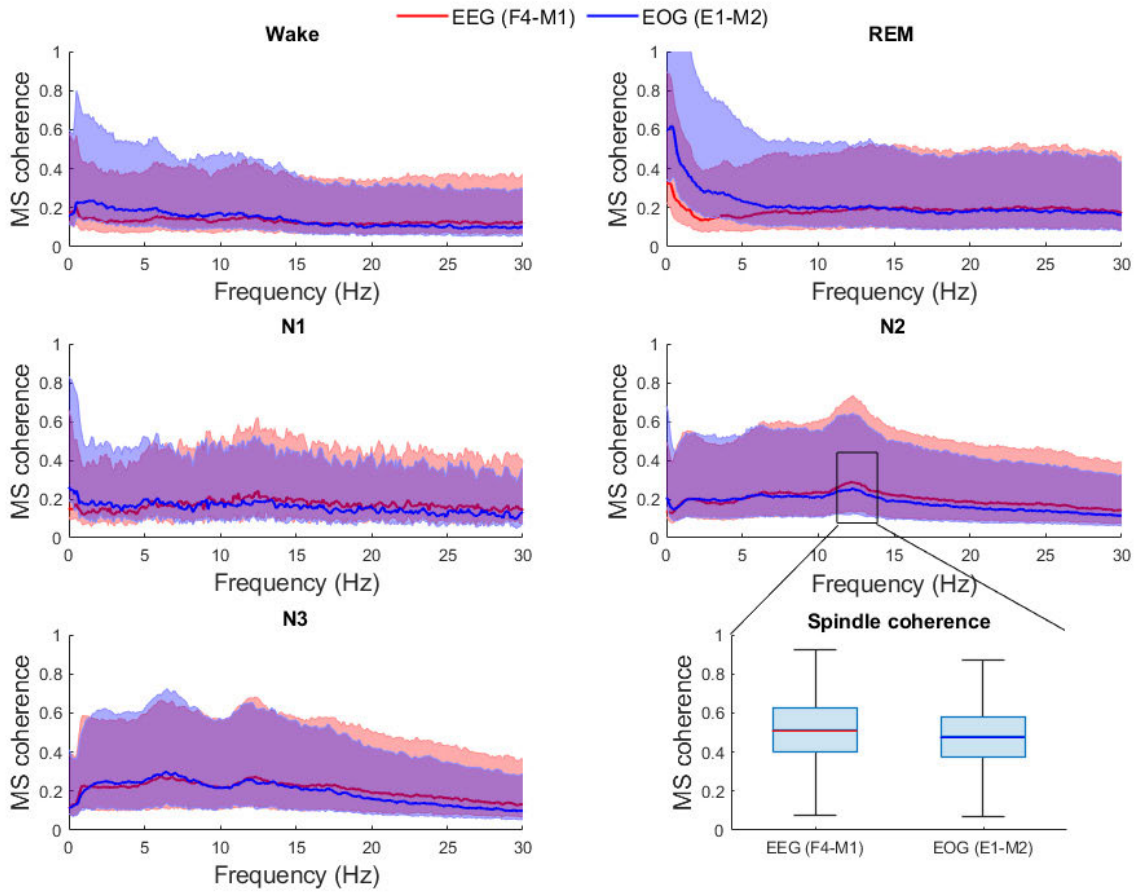

FIGURE 7. Magnitude-squared (MS) coherence estimates computed between textile electrode signals and standard wet electrode EEG (F4-M1) or EOG (E1-M2) signals. Red and blue lines represent the median coherence between textile electrode signal segments and EEG (F4-M1) segments or EOG (E1-M2) segments, respectively. Estimated 25-75% quartiles are marked with respective shaded colors and the overlapping part with a purple color. The box plot shows the spindle coherence between textile electrode signal and standard EEG or EOG signal. Abbreviations: EEG = electroencephalography, EOG = electrooculography, Wake = wakefulness, REM = rapid eye movement sleep, N1 = light sleep, N2 = intermediate sleep, N3 = deep sleep.

derived from the forehead, making the electrode-to-electrode distance relatively short. This can significantly affect the amplitudes. A similar finding of differences in amplitudes with other types of electrodes, and between the standard and forehead signals, supports this conclusion [18]. Furthermore, no major amplitude differences were reported in a study where the textile electrode signals were recorded using similar behind-the-ear referencing as used with the standard electrodes [27]. The amplitude difference between the compared signals might also be partially due to different electrode types. Referring to our previous in-laboratory testing, the textile electrodes have markedly higher skin electrode impedances than the wet electrodes, especially at lower (<10 Hz) frequencies [29]. This might attenuate the textile electrode-based signals resulting in lower amplitude levels compared to those of the wet electrodes. Moreover, many sleep-related EEG features, such as K complexes, sleep spindles, and slow-wave activity, differ in amplitude between brain regions [13]. The recording of these EEG microfeatures using textile electrodes has already been studied, suggesting insignificant differences between the amplitudes when the features are recorded using similar derivations between the electrode types [27]. Moreover, as the measured textile electrode-based signal has a lower amplitude, the signal-to-noise ratio generally remains worse with textile electrodes. Thus, further studies investigating the clinical importance of these differences and the effect on sleep microfeatures in EEG as well as on scoring of sleep stages and cortical arousals are needed. Regarding the amplitude difference, textile electrode-based forehead signals need additional amplification or digital processing. The present results display also sleep stage-specific differences in the amplitudes. The median envelope amplitudes of the textile electrode-based forehead signals were largest in stage N3 and decrease in a similar order as the amplitudes of standard EEG and EOG signals between the sleep stages. The largest relative differences in the compared signal amplitudes were seen during REM sleep, whereas the smallest relative difference was found during the N3 sleep stage. As the relative differences of the compared amplitudes are not systematic between the

sleep stages, the amplitude processing of the textile electrode-based forehead signals might need to be adjusted in a sleep-stage specific manner. A frequency-specific modulation of the signals is likely needed if the forehead signal amplitudes need to correspond to those of the standard derivations.

The results of the spectral analysis conducted in the present study are in line with the results of the amplitude comparison. The PSDs of the textile electrode-based forehead signals were more similar to those of the EOG (E1-M2) signals than the EEG (F4-M1) signals. This is reasonable, as the amplitude of the signal is directly related to the total power seen in the PSD analysis. Also, the recording site of the headband is prone to interference from the EOG. Thus, the forehead EEG signals contain lots of characteristics visible in EOG signals, and considering sleep staging, the standard rules set by the AASM for manual annotation are not directly applicable for these signals. However, the EOG information content of the textile electrode-based forehead signals could be included for automatic sleep staging of the signals. A similar solution has been tested before with different types of electrodes [31]. Nevertheless, the automatic sleep staging accuracy of the textile electrode-recorded forehead signals compared to Type II PSG-based manual sleep staging must be studied before any further conclusions.

Through the applied spectrogram approach, also the temporal correspondence was analyzed. The difference spectrograms showed temporal variation in the agreement of the PSDs even when these were pooled according to sleep stages. Generally, the PSDs at higher frequencies (>30 Hz) are low when artifacts are not present, as the information of the EEG and EOG signals is concentrated at lower frequencies. The difference spectrograms measure the absolute difference and thus, the highest differences were seen during N3 sleep, when the amplitude of the EEG is generally highest. Conversely, when the forehead EEG was compared to EOG, there were no systematic differences in the PSDs. Differences at very low frequencies (<2 Hz) were not systematic, which might be due to dc level drifting, a typical limitation of all biopotential measurements. However, during wake, high and non-systematic differences in the PSDs were seen also at the higher frequencies. This is probably a consequence of extensive movement of the subjects. Especially for healthy volunteers, this type of movement is more likely to happen when awake than during sleep. It might indicate that the Neuroline electrodes and FocusBand textile electrodes have different susceptibility to movement artifacts. More precisely, it can be speculated that this is due to the higher susceptibility of the dry skin-electrode contact to movement artifacts [32].

The spectrograms were constructed in a subject-by-subject order. Therefore, the high spectral power densities related to loosely fit textile electrodes on subject #10 can be identified. More precisely, these can be seen in the spectrograms during the last segments of Wake as well as REM, N1, and N2 sleep. The neoprene headband might be prone to this kind of artifact in unattended recordings. A feasibility study, where the success rate of the headband is considered, needs to be conducted to assess these aspects more accurately in the future.

Coherence estimates between the textile electrode signal and standard EEG and EOG signals gave insight into the agreement at different frequencies. Methodologically speaking, the coherence estimates (Fig. 7) are not as dependent on the absolute power spectral densities as the difference spectrograms (Fig. 5 and Fig. 6). Therefore, the agreement between the textile electrode signal and standard EEG or EOG signal was better than when using the difference spectrograms. Still, coherence estimates had extensive variance following that the median coherence was weak during all sleep stages and between both standard signals and the textile electrode signal. However, there were interesting differences in the coherence spectrums when comparing the forehead EEG to standard EEG and EOG. For example, during REM sleep the textile electrode signals agreed better on low (<10 Hz) frequencies with the EOG signal than with the EEG signals. This is reasonable when considering the placement of the textile electrodes close to the eyes. Furthermore, the spindle coherence, which was defined as maximum coherence between 12-14 Hz during N2 sleep segments, was higher between the textile electrode signal and EEG than between textile electrode signal and EOG. Based on these findings, it can be concluded that the forehead signals recorded using textile electrodes include sleep-specific features typical to EEG signals, although the overall agreement was better with the EOG signal.

One of the limitations of this study is that all the signals were recorded using the same portable PSG system and unipolar EEG channels. This measuring configuration was chosen to enable easy time synchronization and subject-friendly measurements. However, it is a limitation as the reference and ground electrodes are of a wet electrode type. This might slightly skew the derivation of the textile electrode-based signals. The problem related to a different type of reference electrodes has been tested in an earlier study using potential splitters and two synchronized amplifiers [27]. The results of the study suggested that the reference electrode type does not have a significant effect on the quality of the textile electrode signals. Furthermore, the overall stability of the textile electrode-based signals was similar to that of the signals recorded with wet electrodes. However, there might be artifacts present that can be confused with EEG information content and are not separable in the PSD analysis. For example, sweat artifacts have been previously considered as a concern, as the produced disturbance can easily be mixed up with slow-wave activity [27], [33]. This is a significant problem with all types of electrodes as individuals who suffer from e.g. sleep apnea, tend to perspire nocturnally more than healthy people [34], [35]. However, the electrode's tolerance against this type of artifact may be improved with a proper material selection [36]. With textile electrodes, a multilayered design of sweat-absorbable materials is proposed to stabilize the skin-electrode interface [37]. Moreover, the susceptibility of the textile electrodes to movement artifacts can be reduced

with sufficient contact pressure and electrode padding [32]. Considering the FocusBand, the flexible headband seems to be sufficient to maintain reliable skin contact over the whole night, if the neoprene strap is properly tightened. Finally, as the study population consisted of healthy young adults, the results of the present study cannot be fully generalized to patients with sleep disorders. Instead, this issue should be investigated separately later in appropriate patient cohorts.

V. CONCLUSION

Despite the significant amplitude difference, nocturnal forehead EEG signals recorded using textile electrode-based headband have prominent similarities to standard EOG and EEG signals. Moreover, the signals stay relatively stable over the whole night of the recording. Therefore, it can be assumed that the easily attachable headband has potential in home-based sleep disorder screening. However, based on the results of this study, the headband showed its potential for healthy young adults. Thus, more studies assessing the clinical feasibility of the headband and diagnostic usability of the recorded forehead signals in sleep-disordered patient cohorts of different age are needed.

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