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RESEARCH ARTICLE

Self-applied somnography: technical feasibility of electroencephalography and electro-oculography signal characteristics in sleep staging of suspected sleep-disordered adults

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Summary

Sleep recordings are increasingly being conducted in patients' homes where patients apply the sensors themselves according to instructions. However, certain sensor types such as cup electrodes used in conventional polysomnography are unfeasible for self-application. To overcome this, self-applied forehead montages with electroencephalography and electro-oculography sensors have been developed. We evaluated the technical feasibility of a self-applied electrode set from Nox Medical (Reykjavik, Iceland) through home sleep recordings of healthy and suspected sleep-disordered adults ($n = 174$) in the context of sleep staging. Subjects slept with a double setup of conventional type II polysomnography sensors and self-applied forehead sensors. We found that the self-applied electroencephalography and electro-oculography electrodes had acceptable impedance levels but were more prone to losing proper skin–electrode contact than the conventional cup electrodes. Moreover, the forehead electroencephalography signals recorded using the self-applied electrodes expressed lower amplitudes (difference 25.3%–43.9%, $p < 0.001$) and less absolute power (at 1–40 Hz, $p < 0.001$) than the polysomnography electroencephalography signals in all sleep stages. However, the signals recorded with the self-applied electroencephalography electrodes expressed more relative power ($p < 0.001$) at very low frequencies (0.3–1.0 Hz) in all sleep stages. The electro-oculography signals recorded with the self-applied electrodes expressed comparable characteristics with standard electro-oculography. In conclusion, the results support the technical feasibility of the self-applied electroencephalography and electro-oculography for sleep staging in home sleep recordings, after adjustment for amplitude differences, especially for scoring Stage N3 sleep.

KEYWORDS
home sleep apnea testing, portable diagnostics, sleep-disordered breathing

1 | INTRODUCTION

The increasing evidence of negative health consequences (Jose et al., 2005; Luyster et al., 2014; Zinchuk et al., 2018) and economic burden (AASM, 2016; Benjafield et al., 2019; Hillman et al., 2018; Mattila et al., 2022) originating from sleep-disordered breathing (SDB), including obstructive sleep apnea (OSA), is globally acknowledged. This epidemiological pressure has set the need for cost-effective and patient-centric technological solutions to be adopted into practices of diagnostics and therapeutics of SDB (Arnardottir et al., 2022; Cooksey & Balachandran, 2016). Thus, sleep medicine is now experiencing a technical revolution (Arnardottir et al., 2022).

One of the main objectives of adopting these new practices is to bring sleep studies from hospitals into patients' homes (Arnardottir et al., 2022; Korkalainen et al., 2021). Currently, attended polysomnography (PSG) in a sleep laboratory is the 'gold standard' measurement for diagnostics of SDB (AAST, 2012). Although a valid method for differential diagnostics of sleep disorders, PSG comes with several disadvantages (Hirshkowitz, 2016; Korkalainen et al., 2021). The limited availability, high costs, and in-person overnight supervision of PSG have made portable monitoring and home sleep apnea testing (HSAT) a more practical option for screening and diagnosing SDB (Berry et al., 2020; Bruyneel & Ninane, 2014; Ghegan et al., 2006).

Guidelines for portable monitoring were set in 1994 (Ferber et al., 1994) and updated in 2007 (Collop et al., 2007). Furthermore, guidelines for HSAT are set in the American Academy of Sleep Medicine (AASM) manual (Berry et al., 2020). Type III sleep studies, which include unattended portable monitoring of four to seven channels without electroencephalography (EEG), have become the standard practice in HSAT globally (Calero & Anderson, 2016; Cooksey & Balachandran, 2016; Fischer et al., 2012). Unfortunately, several studies indicate that sleep time cannot be quantified accurately with current HSAT because of the lack of EEG, causing for example underestimation of the severity of OSA (Bianchi & Goparaju, 2017; Ghegan et al., 2006; Kalevo et al., 2022; Lim & See, 2021). The seldom use of EEG in HSAT comes from the lack of easy-to-use sensors that could be applied to portable monitoring (Bruyneel & Ninane, 2014).

In recent years, a substantial amount of research has been conducted to develop sensors that could be self-applied by subjects at home (Korkalainen et al., 2021). In addition, many other novel sleep monitoring technologies, such as in-ear EEG, smart rings, and watches have been developed (Altini & Kinnunen, 2021; Cay et al., 2022; da Silva Souto et al., 2021; De Zambotti et al., 2019). However, these have not been widely adopted into clinical use, due to lack of clinical validation or possibly reduced accuracy in sleep staging. As a result, self-applied EEG electrode sets and devices intended for home sleep monitoring have been introduced (Arnal et al., 2020; Carneiro et al., 2020; Kainulainen et al., 2021; Miettinen, Myllymaa, Muraja-Murro, et al., 2018; Rusanen et al., 2021). These self-applied EEG setups have already shown promising results in increasing the accuracy of sleep staging in HSAT to a nearly comparable level with standard PSG (Jónsson et al., 2020; Kalevo et al., 2022; Leino et al., 2022; Levendowski et al., 2017; Rusanen et al., 2023). However, self-applied EEG is not used clinically and needs more rigorous validation.

In this study, we focused on the technical evaluation of a self-applied somnography (SAS) (Figure 1) developed by Nox Medical (Reykjavik, Iceland). Previous studies have considered the success rate and data quality of this setup in a large cohort study (Punjabi et al., 2022), as well as its potential for automatic sleep staging (Jónsson et al., 2020). Moreover, the frontal EEG signal characteristics of this frontal EEG setup have been compared with those of standard PSG in a paediatric cohort (Kainulainen et al., 2021). However, adult patients with suspected sleep disorders form a significantly different patient cohort. For example, a common symptom of patients with SDB is sweating (Arnardottir et al., 2013), which can drastically affect the technical quality of the recordings (Kalevo et al., 2020). Therefore, more comprehensive testing is needed to further validate the setup in an adult patient population.

Based on previous research, we hypothesised that the EEG and electro-oculography (EOG) signals recorded using this setup have slightly different characteristics than those recorded in PSG. However, the setup is hypothesised to be a technically feasible option for sleep recordings and sleep scoring when the differences in signal characteristics are acknowledged. We aimed to investigate this by comparing the frontal EEG and EOG to conventional EEG and EOG in signal amplitudes and frequency content during different sleep stages. Furthermore, we aimed to study the technical feasibility of the frontal and conventional electrodes in home-based EEG recordings by comparing the overnight skin–electrode contact impedances of both setups.

2 | METHODS

2.1 | Double-setup recordings

The data of the present study comprised simultaneous recordings with type II PSG and Nox SAS setup. These double-setup recordings were set up at the Reykjavik University Sleep Institute by trained staff, overseen by expert sleep technologists, and the subjects slept at home. The participants were recruited using an open online screening questionnaire for interested volunteers. Based on the screening, a balanced cohort of healthy subjects and those who were at risk of OSA, restless legs syndrome, or insomnia were recruited (Table 1). Data collection was given a favourable statement by the National Bioethics Committee of Iceland (21-070, March 16, 2021) and all subjects gave informed written consent before recording.

Two Nox A1 PSG devices (Nox Medical) were used for recording signals simultaneously from type II PSG and Nox SAS setups according to the clinical guidelines (Berry et al., 2020). The PSG setup included a recording of EEG, EOG, and chin muscle tone with electromyography (EMG). There was one ground (GND) electrode and two reference electrodes (M1 and M2) for EEG and EOG. The EEG was recorded with six channels (F4, C4, and O2 with the F3, C3, and O1 channels as backup), and the EOG was recorded with two channels

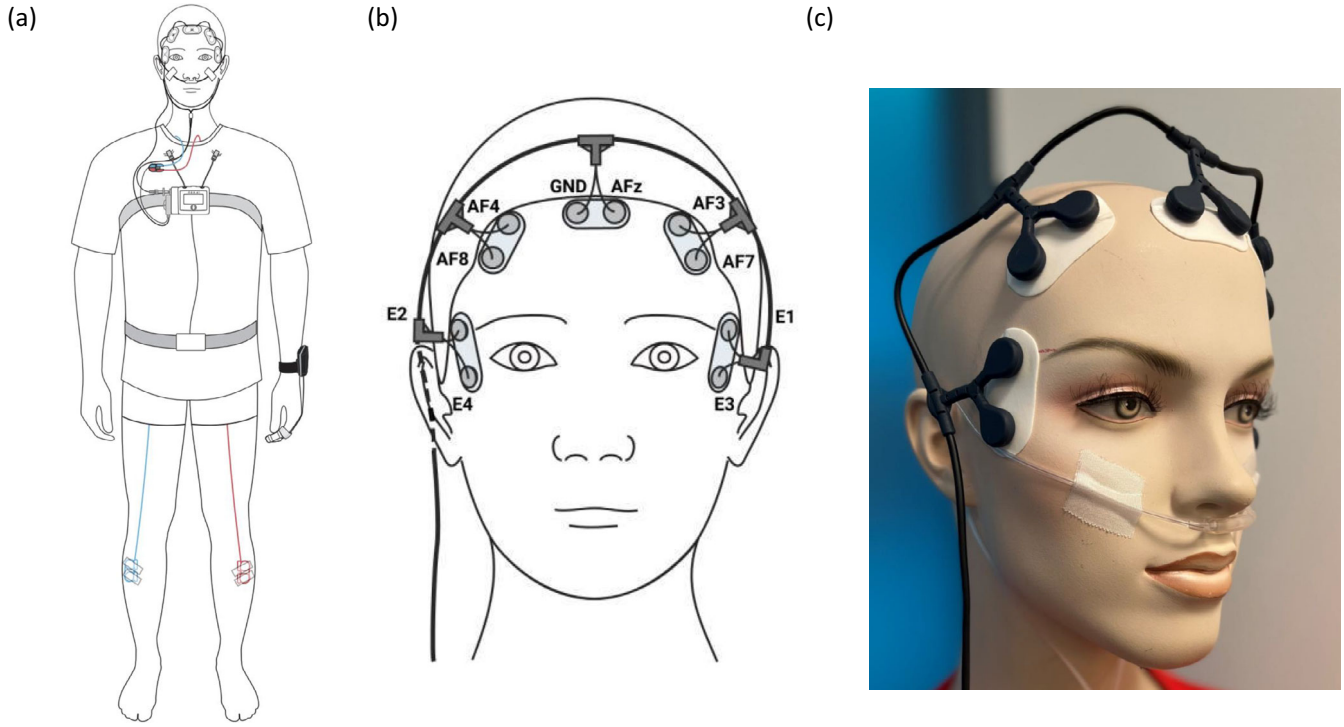

FIGURE 1 Illustration of the self-applied somnography (SAS) sensor setup from Nox Medical (Reykjavik, Iceland) in (a) full configuration, (b) the headset, and (c) image of the headset applied on a manikin. Illustrative figures of the sensors are courtesy of Nox Medical.

(E1 and E2). The chin EMG was recorded with two electrodes placed below the mandible, and one reference electrode placed above the mandible. Nox gold cup electrodes (Nox Medical) were used for all signals except the chin EMG. Chin EMG was recorded using Ambu Neuroline Cup or Ambu Neuroline 715 electrodes (Ambu A/S, Ballerup, Denmark) if the subject was bearded. The cup electrodes were filled with Ten20 (Weaver and Company, Aurora, CO, USA) conductive paste and the skin was prepared using Nuprep skin prep gel (Weaver and Company) before application. Electrodes applied in areas with hair were attached using Grass EC2+ Electrode cream (Natus Medical Incorporated, Pleasanton, CA, USA), whereas electrodes applied in hairless areas were attached using Leukosilk Tape (BNS Medical GmbH & Co, Hamburg, Germany).

In the SAS setup, there were four channels for frontal EEG (AF4, AF3, AF8, and AF7), four channels for EOG (E1, E2, E3, and E4), a reference electrode (AFZ), and a GND electrode, specially developed for the SAS setup (as shown in Figure 1). The EOG channels were referenced to AFZ or bipolarly and the EEG channels were referenced to the averaged EOG channel ([E3 + E4]/2). This was done to minimise eye movement artefacts in the EEG. Before applying the electrodes for the SAS setup, the skin was prepared by wiping it clean with alcohol. Leukosilk Tape was used for additional fixation of the SAS setup.

Only the Nox A1 device with the SAS setup was used to record the following signals: a three-lead electrocardiogram (ECG), leg movements with right and left leg EMG, nasal breathing with a nasal cannula, respiratory movements with thorax and abdomen respiratory inductance plethysmography belts, pulse and oxygen saturation with a finger probe oximeter, activity and body position with a three-dimensional acceleration sensor, and snoring/audio with a microphone. All of these were recorded similarly as in standard PSG. In addition, skin electrodermal activity was measured.

The PSGs were manually scored by an expert sleep technologist following the guidelines from the AASM Manual for the Scoring of Sleep and Associated Events Version 2.6 (Berry et al., 2020).

2.2 | Data analysis

In this study, we focused on comparing EEG and EOG signals recorded with the standard electrodes against those recorded with the self-applied electrodes frontally. The frontal derivations F4-M1 and F3-M2 of PSG EEG were chosen for comparison due to them having reasonable counterparts in SAS EEG: AF4-E3E4 and AF3-E3E4, respectively. Similarly, the E1-M2 and E2-M1 derivations of PSG EOG were compared with their closest counterparts i.e., E1-E4 and E2-E3, as well as E3-AFZ and E2-AFZ derivations of SAS EOG, respectively. All recordings were exported into European Data Format (EDF) files and signals were analysed with codes programmed in Python using a sampling rate of 200 Hz for the EEG and EOG signals.

We first analysed the impedances of the recorded channels to study the skin–electrode contact of electrodes and the technical quality of the signals for both PSG and SAS electrode setups. During the night, the impedance of each channel is sampled injecting

TABLE 1 Demographic information of the study population ($n = 174$).

Variable	Value
Number of patients (% male)	174 (46.6)
Age, years, mean (SD, range)	43.5 (14.5, 18–79)
BMI, kg/m^2, mean (SD)	29.1 (5.9)
AHI, events/h, mean (SD)	12.4 (12.8)
No OSA (AHI < 5 events/h), *n* (%)	66 (37.9)
Mild OSA (5 ≤ AHI < 15 events/h), *n* (%)	58 (33.3)
Moderate OSA (15 ≤ AHI < 30 events/h), *n* (%)	32 (18.4)
Severe OSA (AHI ≥ 30 events/h), *n* (%)	18 (10.3)
ODI, events/h, mean (SD)	11.4 (12.3)
Total sleep time, min, mean (SD)	423.1 (69.3)
Sleep stage, mean (SD)	
Wake, % of recording	21.6 (9.9)
Stage N1, % of recording	14.5 (6.7)
Stage N2, % of recording	28.9 (7.8)
Stage N3, % of recording	12.6 (4.6)
Stage REM, % of recording	17.8 (5.0)
Unscored, % of recording	4.6 (5.9)
Sleeping position, mean (SD)	
Upright, % of recording	5.3 (8.6)
Left, % of recording	20.1 (15.7)
Right, % of recording	21.1 (14.8)
Prone, % of recording	5.9 (8.5)
Supine, % of recording	47.5 (21.4)

Note: The % of recording was computed as the percentage of the analysed period of the total recording.
Abbreviations: AHI, apnea-hypopnea index; BMI, body mass index; ODI, oxygen desaturation index (AASM 2012 scoring criteria; desaturation ≥3%); OSA, obstructive sleep apnea; REM, rapid eye movement sleep; SD, standard deviation.

a small current of 200 Hz through the channels. Half of the channels are in the opposite phase compared to the other half. The impedance of each channel is then defined according to $Z_{CH} = (V_{CH} + 11 \times V_{REF} + (I_{PGND} + Z_{PGND}))/I_{CH}$, where V_{CH} and I_{CH} are the voltage and current of the measured channel, V_{REF} is the reference channel voltage, I_{PGND} is the residual current of the GND channel, and Z_{PGND} the GND electrode impedance. Normalised histograms of raw impedance values, 1 sample/s over the whole recording, were created for each channel and compared between SAS and PSG. Lower impedance levels (0–50 kΩ) were studied separately from high impedances (>200 kΩ). Lower impedances are thought to quantify the established skin–electrode contact. Higher impedances are assumed to represent the electrode losing contact. The statistical significance of differences in impedance distributions between PSG and SAS channels was investigated using the Kolmogorov–Smirnov test.

Following the impedance analysis, the signal characteristics of PSG and SAS signals were compared. The analysed signals were first synchronised timewise for each participant according to device clocks. The recordings were then split and pooled into 30-s segments (epochs) according to sleep stages scored using the PSG recording; wakefulness (Wake); non-rapid eye movement (NREM) sleep stages N1, N2, and N3; and rapid eye movement (REM) sleep. Pooling produced 41,303 epochs of Wake, 27,262 epochs of Stage N1 sleep, 54,224 epochs of Stage N2 sleep, 23,693 epochs of Stage N3 sleep, and 33,773 epochs of REM sleep. In addition, 8708 epochs (5% of all epochs) were left unscored between scored epochs due to the poor quality of the PSG recording. These epochs were excluded from the analysis.

We first quantified signal amplitudes for each channel using a similar envelope-based method as in previous studies (Kainulainen et al., 2021; Rusanen et al., 2021). All signals were filtered with a fifth-order type II Chebyshev bandpass (0.3–35 Hz) filter before amplitude analysis. As such, we wanted the envelope amplitudes to better represent the amplitudes used in sleep staging (Berry et al., 2020). Furthermore, we tested how increasing the lower cut-off frequency of the bandpass filter to 0.5 Hz for the SAS signals affected the amplitudes. This was performed due to previous experience of the SAS signals occasionally showing harmful low-frequency (<0.5 Hz) oscillations, which could be removed with the aforementioned adjustment in standard filtering settings. Differences in amplitude distributions between PSG and SAS channels were compared using Wilcoxon signed-rank test.

In addition to amplitude characteristics, we compared the frequency content of unfiltered SAS and PSG signals. By unfiltered signals, we mean the raw recorded signals that have been filtered only at the hardware level. Nox A1 unipolar EEG channels have an input range of ±3.2 mV with 0.1–85 Hz bandpass and alternating coupling. A comparison of frequency content was performed by estimating the power spectral densities (PSDs) for each 30-s epoch in a stationary manner using Welch's average periodogram method. A Hamming window and a 50% overlap between segments of length 1500 samples were applied with zero-padding to 0.02 Hz frequency resolution for smooth visualisations. Relative powers for low frequencies were computed from the PSD estimates using the composite Simpson's rule to estimate the integral at 0.3–1 Hz and dividing it with the total power at 0.3–40 Hz. The statistical significance of differences in PSDs and relative powers was compared using Wilcoxon signed-rank test. To ensure comparable results with a previous study (Kainulainen et al., 2021), differences in median PSDs between PSG and SAS signals were visualised by computing PSD ratios in each sleep stage.

3 | RESULTS

3.1 | Impedance results

The contact impedance values (<50 kΩ) showed that impedances for SAS AF4, AF8, E3, and E2 channels were on the same level (median difference 0.6–4.5 kΩ) as impedances for the counterpart PSG channels (Figure 2). However, significantly higher impedances (median differences 7.1–15.5 kΩ, $p < 0.001$) were observed for SAS AF3, AF7,

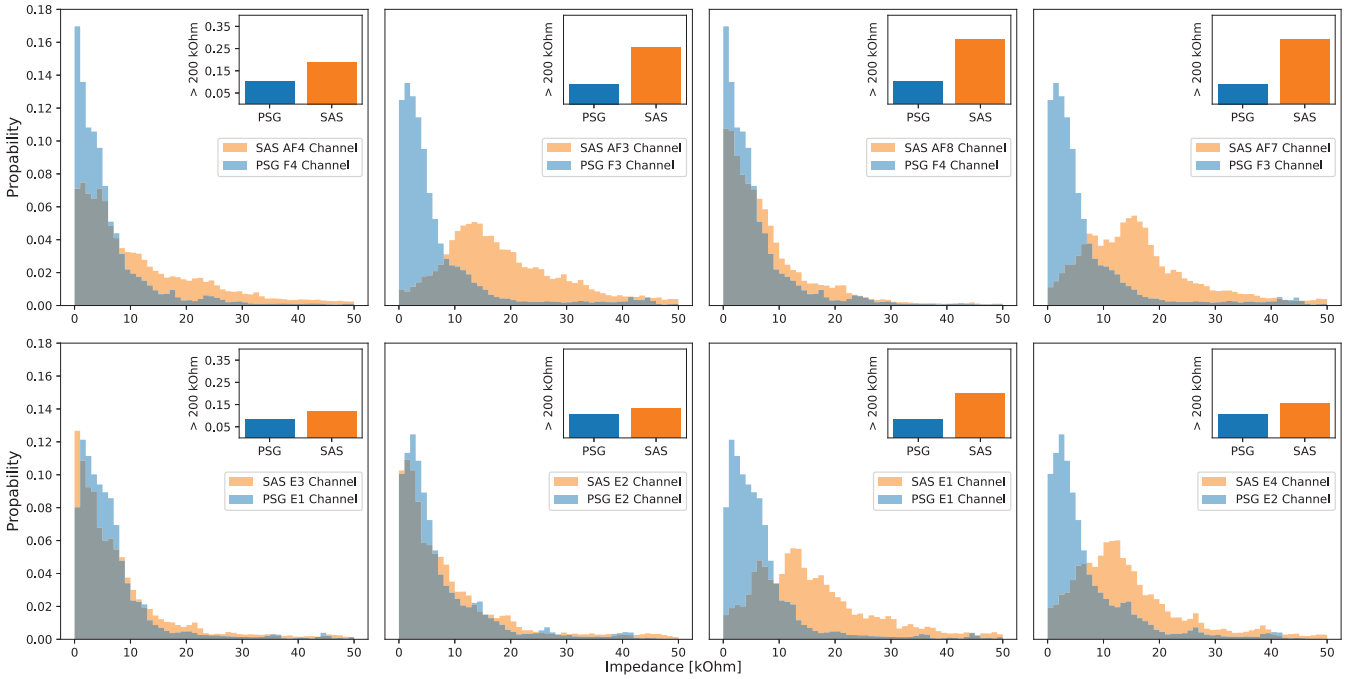

FIGURE 2 Normalised histograms of raw impedance values (1 sample/s) for all recordings ($n = 174$). Inset axes show the probability for high impedance values (>200 kΩ) between SAS and PSG channels. The width of each pin was set to 1 kΩ. PSG, polysomnography; SAS, self-applied somnography.

E1, and E4 channels compared to the counterpart PSG channels (Figure 2). In addition, SAS channels were more prone to have high impedance values (>200 kΩ) compared to PSG channels (Figure 2). The mean (range) probabilities for high impedances were 0.20 (0.12–0.29) for SAS channels and 0.10 (0.08–0.11) for PSG channels. Similar findings were observed when impedance distributions were compared in each sleep stage separately. Probabilities for high impedance values (>200 kΩ) were highest in stage Wake with both setups (Figures S1 and S2).

3.2 | Amplitude results

The results of the envelope analysis of EEG signals clearly showed that the SAS AF4-E3E4 signals had significantly lower ($p < 0.001$) median amplitudes compared to PSG F4-M1 amplitudes in all sleep stages (relative differences 31.6%–43.9%) when both signals were filtered with the standard settings (Table 2, Figure 3). Similar differences were observed on counter hemispheric derivations (AF3-E3E4 versus F3-M1; relative differences 25.3%–43.9%, $p < 0.001$). Instead, SAS E1-E4 median amplitudes were more similar to the PSG E1-M2 median amplitudes, although consistently lower (relative differences 4.5%–25.3%, $p < 0.001$; Table 2). SAS E2-E3 signal showed similar consistency and even lower amplitudes in each sleep stage when compared to PSG E2-M1 (relative differences 19.0%–33.0%, $p < 0.001$; Table S1). Moreover, SAS E3-AFZ and E2-AFZ median amplitudes differed ($p < 0.001$) from PSG E1-M2 and E2-M1 median amplitudes in all sleep stages (relative differences −9.5%–16.1% and 10.2%–38.7%, respectively; Table S2).

When amplitude analysis was performed using a bandpass filter of 0.5–35 Hz for the SAS signals, the amplitude differences of the compared EEG signals increased further (relative differences 35.6%–47.5% between F4-M1 and AF4-E3E4 signals; Table 3). Similarly, differences increased in the EOG comparison and were 13.5%–31.1% between E1-E4 and E1-M2 signals depending on the sleep stage and whether the upper or lower envelope amplitudes were compared (Table 3).

3.3 | Power spectral density comparison

In frequency content analysis, SAS and PSG EEG signals had similar overall morphology in median power spectral densities in all sleep stages (Figure 4). However, SAS signals (AF4-E3E4 and AF3-E3E4) expressed consistently less absolute power ($p < 0.001$) at frequencies ranging from 1 to 40 Hz compared to frontal PSG EEG signals. SAS EEG signals also expressed more relative power ($p < 0.001$) at the low frequencies (0.3–1 Hz, Figure S3) causing a rapid increase in the ratio (SAS/PSG) of median PSDs towards lower frequencies (Figure 4). Moreover, in sleep stages N2 and N3, SAS EEG signals expressed relatively higher peaks on median PSDs at frequencies ranging from 8 to 15 Hz than PSG EEG signals increasing the ratio of median PSDs closer to 1 (Figure 4). Hemispheric differences in the comparison of PSD ratios between the SAS and PSG EEG signals were negligible (Figure 4).

Envelope	Sleep stage	AF4-E3E4, μV	F4-M1, μV	Difference, %
Lower	Wake	−14.7	−21.9	32.9
	N1	−10.5	−18.7	43.9
	N2	−13.3	−23.6	43.8
	N3	−22.1	−36.6	39.5
	REM	−8.9	−15.6	42.6
Upper	Wake	15.1	21.7	30.6
	N1	10.4	17.4	39.9
	N2	13.1	22.3	41.0
	N3	21.9	35.0	37.5
	REM	8.9	14.5	38.3
Envelope	**Sleep stage**	**E1-E4, μV**	**E1-M2, μV**	**Difference, %**
Lower	Wake	−14.0	−18.7	25.3
	N1	−10.3	−12.1	15.0
	N2	−12.1	−13.7	12.0
	N3	−19.7	−20.9	5.5
	REM	−9.3	−11.2	16.6
Upper	Wake	14.3	19.0	24.7
	N1	10.6	12.1	12.6
	N2	12.3	13.5	9.4
	N3	19.9	20.8	4.5
	REM	9.7	11.2	13.5

TABLE 2 Median values of lower and upper amplitude distributions for 30-s electroencephalography (AF4-E3E4 and F4-M1) and electro-oculography (E1-E4 and E1-M2) signal segments as well as relative differences between self-applied somnography and polysomnography signals when all signals were filtered with the 0.3–35 Hz bandpass filter.

Note: The relative difference in median amplitudes between self-applied somnography (SAS) and polysomnography (PSG) signals was computed by subtracting SAS amplitude from PSG amplitude and dividing it by PSG amplitude. The statistical significance of differences in amplitude distributions was tested with Wilcoxon signed-rank test. All differences were statistically significant ($p < 0.001$). Abbreviation: REM, rapid eye movement.

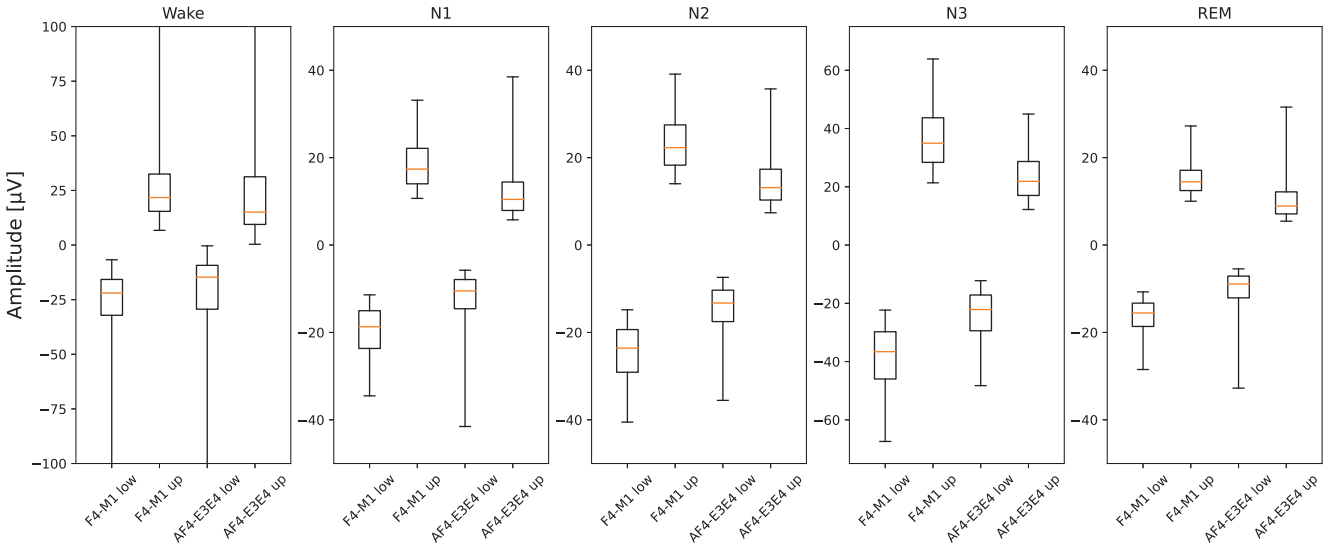

FIGURE 3 Box plot of upper (up) and lower (low) envelope amplitude levels for PSG F4-M1 channel and its counterpart SAS AF4-E3E4 channel. Amplitudes of the SAS EEG signal were significantly lower ($p < 0.001$) than amplitudes of the PSG EEG signal in all sleep stages. Box plot explanation; median (orange line), 25–75 percentile of the data (box edges), 5–95 percentile of the data (whiskers). EEG, electroencephalography; PSG, polysomnography; SAS, self-applied somnography. Please note different value ranges in *y*-axes.

TABLE 3 Median values of lower and upper amplitude distributions for 30-s electroencephalography (AF4-E3E4 and F4-M1) and electro-oculography (E1-E4 and E1-M2) signal segments as well as relative differences between self-applied somnography (SAS) and polysomnography (PSG) when SAS signals were filtered with the 0.5–35 Hz bandpass filter and PSG signals with the 0.3–35 Hz bandpass filter.

Envelope	Sleep stage	AF4-E3E4, μV	F4-M1, μV	Difference, %
Lower	Wake	−13.5	−21.9	38.3
	N1	−9.8	−18.7	47.4
	N2	−12.4	−23.6	47.5
	N3	−19.5	−36.6	46.9
	REM	−8.3	−15.6	46.4
Upper	Wake	14.0	21.7	35.6
	N1	9.7	17.4	43.9
	N2	12.2	22.3	45.3
	N3	18.9	35.0	45.8
	REM	8.4	14.5	42.2
Envelope	**Sleep stage**	**E1-E4, μV**	**E1-M2, μV**	**Difference, %**
Lower	Wake	−13.1	−18.7	30.1
	N1	−9.8	−12.1	19.6
	N2	−11.5	−13.7	15.9
	N3	−17.7	−20.9	15.3
	REM	−8.7	−11.2	21.8
Upper	Wake	13.4	19.0	29.4
	N1	10.0	12.1	17.0
	N2	11.7	13.5	13.5
	N3	17.8	20.8	14.6
	REM	9.1	11.2	18.7

Note: The relative difference in median amplitudes between SAS and PSG signals was computed by subtracting SAS amplitude from PSG amplitude and dividing it by PSG amplitude. The statistical significance of differences in amplitude distributions was tested with Wilcoxon signed-rank test. All differences were statistically significant ($p < 0.001$).
Abbreviation: REM, rapid eye movement.

Self-applied somnography EOG signals E1-E4 and E3-AFZ only slightly varied in absolute powers against the PSG E1-M2 signal (Figure 5). SAS EOG signals E2-E3 and E2-AFZ expressed smaller ($p < 0.001$) absolute powers compared to the PSG E2-M1 signal (Figure 5). PSD ratios between the compared SAS and PSG EOG signals showed variation in the agreement of median PSDs between analysed frequencies and sleep stages (Figure 5). In sleep stages N2 and N3, SAS EOG signals showed relatively higher peaks at 8–15 Hz than standard PSG EOG, increasing the PSD ratio at these frequencies (Figure 5). Overall, the SAS E1-E4 and E2-E3 signals had PSD ratios closer to 1 than SAS signals with AFZ reference (E3-AFZ and E2-AFZ) when compared against their PSG EOG counterpart (Figure 5). In addition, no clear differences were observed in the relative powers at 0.3–1.0 Hz between SAS and PSG EOG (Figure S3).

4 | DISCUSSION

In the present study, we evaluated the technical feasibility of a self-applied frontal EEG and EOG setup for home-based sleep recordings. Technical testing was conducted by comparing the self-applied setup (SAS) with the PSG setup in an extensive cohort of adults with suspected sleep disorders as well as healthy subjects. The main focus was on comparing the skin–electrode impedances and signal characteristics of the two setups. We found that the SAS EEG and EOG signals expressed similar or higher contact impedances against the reference PSG signals. Moreover, the self-adhesive electrodes of the frontal setup were more susceptible to losing contact than standard PSG electrodes. The EEG signals recorded with the frontal setup had notably smaller peak-to-peak amplitudes and less absolute power than the PSG counterparts at frequencies ranging from 1 to 40 Hz. The SAS and PSG setups showed comparable EOG signal characteristics. Apart from the detected differences, the overall technical feasibility and morphology of the SAS EEG and EOG signals were on a good level and could be suitable for sleep staging of home PSGs. The findings from previous studies support that, too (Jónsson et al., 2020; Kainulainen et al., 2021; Punjabi et al., 2022).

Differences in contact impedances of the SAS and PSG electrodes were notable. These could have originated from different designs of the electrodes. Conventional cup electrodes use externally added wet gel between the electrode and the skin whereas the SAS electrodes are pre-gelled and solid. In addition, the skin was prepared with an adhesive scrub for the PSG electrode, whereas for the SAS electrodes, it was only wiped with an alcohol pad. Nevertheless, higher contact

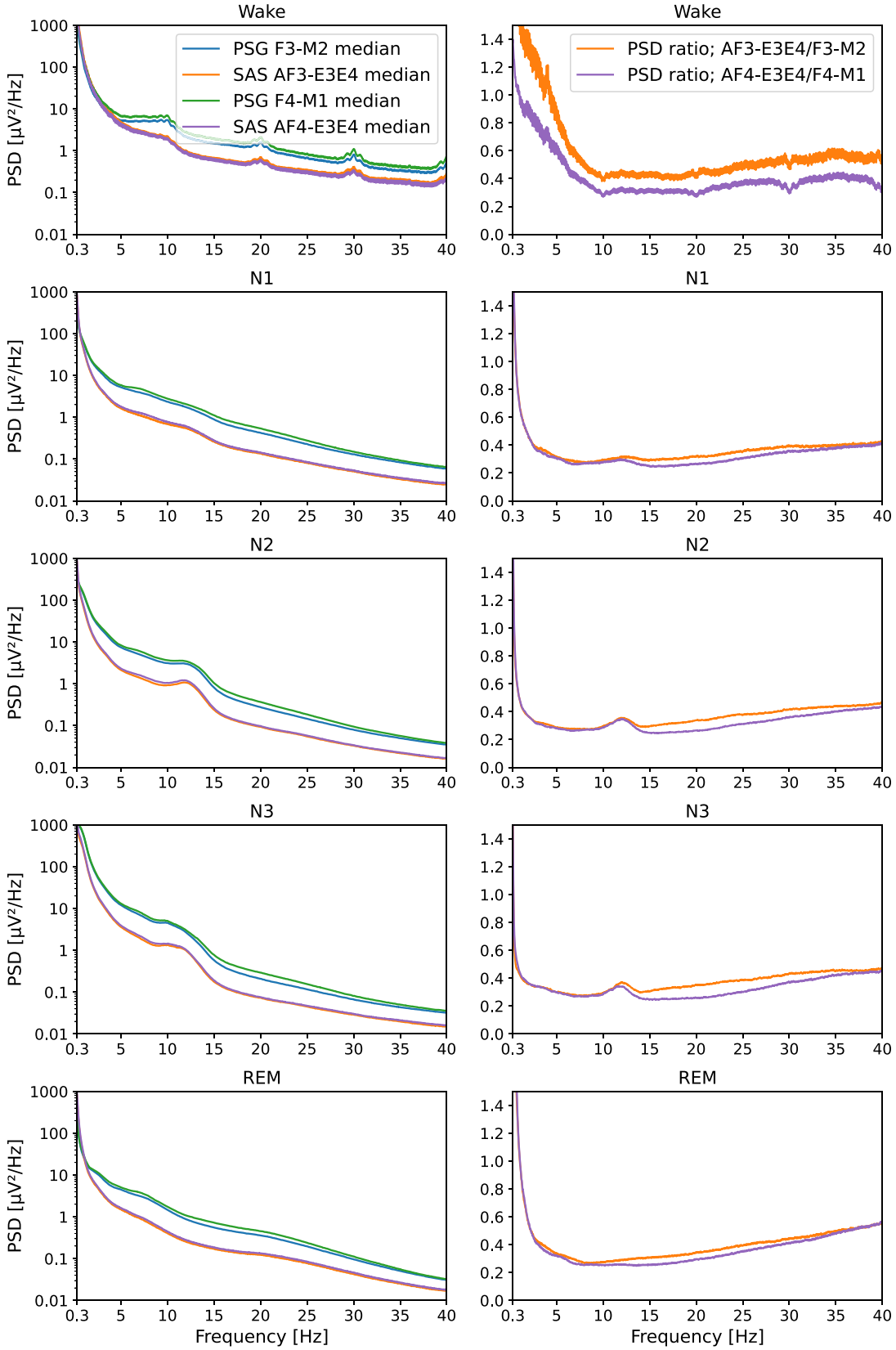

FIGURE 4 Median power spectral densities (PSDs) and PSD ratios for simultaneously recorded frontal type II PSG EEG (F3-M2 and F4-M1) and SAS EEG (AF3-E3E4 and AF4-E3E4) signals. EEG, electroencephalography; PSG, polysomnography; REM, rapid eye movement; SAS, self-applied somnography (Nox Medical).

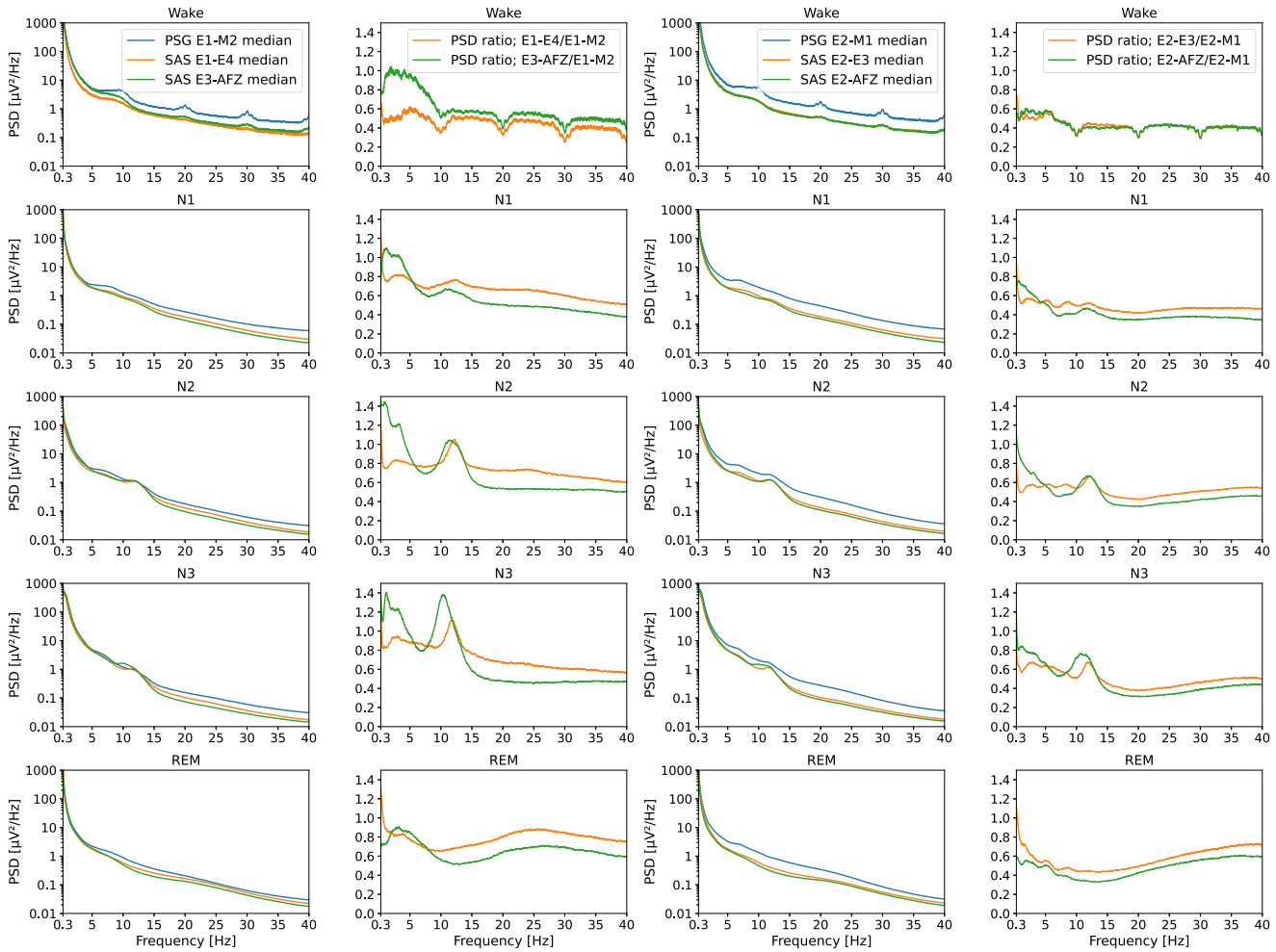

FIGURE 5 Median power spectral densities (PSDs) and PSD ratios for simultaneously recorded type II PSG EOG (E1-M2 and E2-M1) and self-applied somnography (Nox Medical) EOG (E1-E4, E3-AFZ, E2-E3, and E2-AFZ) signals. EOG, electro-oculography; PSG, polysomnography; REM, rapid eye movement; SAS, self-applied somnography (Nox Medical).

impedances can increase the noise in the recorded signals (Kappenman & Luck, 2010). A generally accepted contact impedance limit for diagnostic recordings is 5 kΩ (Berry et al., 2020). However, this limit may be unnecessarily strict with modern biopotential amplifiers that have high input impedances (>5 MΩ in Nox A1 device). Therefore, the contact impedances of the SAS electrodes (median impedances <26 kΩ) can be considered technically feasible for clinical PSGs.

The SAS electrodes were more prone to lose the proper skin–electrode contact. i.e., showing >200 kΩ impedances. This raises concern, especially because the electrodes were fixed by trained staff and with additional skin tape. In the intended end-use setting where patients self-apply the electrodes, more electrode-contact failures could be expected. However, a previous study showed that out of 1013 studies, only 46 (4.5%) failed due to the self-applied EEG producing <3 h of good-quality signal (Punjabi et al., 2022). This could be explained by the SAS setup comprising backup channels if one of the four channels for EEG or EOG fails. Therefore, contemporary self-adhesive electrodes may already be suitable for diagnostic sleep recordings.

Impedance results also expressed a small imbalance in the raw impedance values between the SAS channels (AF4, AF8, E3, E2 vs. AF3, AF7, E1, E4). This is due to the Nox A1 impedance protocol, where the GND electrode impedance is first measured at the start of the recording and assumed to be constant over the recording. If the GND electrode impedance changes during the night, it adds positive or negative bias to the recorded impedance depending on whether the channel current is in phase or inverse phase with the residual current. Therefore, the AF3, AF7, E1, and E4 channels might have a small positive bias, and the AF4, AF8, E3, and E2 channels a small negative bias from true impedance values, both originating from the changes in Z_{PGND}. However, this does not affect the conclusions based on differences between SAS and PSG channel impedances, as impedances are measured and defined similarly for both setups.

Despite the relatively small differences in the impedances, the amplitude characteristics differed between PSG and SAS. Differences in the amplitudes of the EEG signals are unlikely to fully originate from the higher resistance of the skin–electrode interface with SAS electrodes. These might rather originate from different referencing of the

SAS signals. The SAS EEG signals are referenced against the average of E3 and E4 channels due to the absence of standard mastoid reference electrodes. The shorter distance between electrodes, possibly higher contact impedances, and referencing to EOG activity affect the SAS EEG signals by dampening the total power of the signals. The reduced peak-to-peak amplitudes and suppressed overall power can negatively affect the signal-to-noise ratio of the SAS EEG signals. Incorporating mastoid reference electrodes could be one option to tackle challenges related to reduced peak-to-peak amplitudes. The sites around the mastoids are free from the activity of neural sources and are suitable for electrode self-application (Kalevo et al., 2022).

Moreover, the reduced amplitudes of EEG signals directly affect the scoring of different sleep stages. When following the AASM scoring criteria, Stage N3 is scored with a 75 μV peak-to-peak amplitude limit for the 0.5–2 Hz slow-wave activity in frontal EEG (F4-M1 and F3-M2 signals; Berry et al., 2020). Furthermore, characteristic low-amplitude, mixed-frequency EEG is used in the detection of REM and Stage N1 sleep (Berry et al., 2020). Therefore, scoring criteria where amplitudes play a major factor need to be adjusted to overcome these differences when SAS EEG is used. Based on the average of 40% amplitude difference between SAS and PSG frontal EEG signals, we recommend reducing the Stage N3 scoring limit to 45 μV (= [1–0.40] × 75 μV) if a standard passband of 0.3–35 Hz is used in filtering. This amplitude recommendation on adult subjects is consistent with that made in a paediatric cohort (Kainulainen et al., 2021). However, if the lower cut-off frequency is increased to 0.5 Hz excluding harmful low-frequency artefacts, the amplitude limit for scoring Stage N3 should be decreased further, e.g., to 40.5 μV (= [1–0.46] × 75 μV) with the filter design used in this study.

Although the SAS EEG signals showed decreased power compared with PSG EEG during sleep, the differences were relatively constant in median PSDs at >1 Hz frequencies. Instead, PSD comparison showed a rapid increase in the SAS/PSG ratio towards lower frequencies. Furthermore, the median PSD estimates during stage Wake showed peaks likely originating from distorted and artefactual unfiltered signals. This is typical for overnight ambulatory recordings, where Stage Wake includes anything from the subject lying in the bed eyes closed to evening activities outside the bed. These differences could be addressed using digital processing of the signals. In the AASM technical specifications, bandpass filtering with lower and upper limits of 0.3 and 35 Hz is recommended as routine processing of PSG EEG (Berry et al., 2020). Increasing the recommended cut-off frequency of a high-pass filter from 0.3 to 0.5 Hz when using SAS signals in sleep staging could be beneficial to exclude the inconsistency in power content at the lower frequencies. With a filter design of fast cut-off between pass- and stopbands and with ripples on the stopband side, this should not drastically affect the clinically relevant 0.5–2 Hz slow waves. However, the amplitude results showed that increasing the lower cut-off frequency to 0.5 Hz decreased the SAS EEG and EOG signal amplitudes even further. Therefore, the amplitude limit for scoring Stage N3 sleep should be adjusted depending on the used filter. The practical relevance of this needs to be confirmed in further studies, e.g., comparing the manual scoring of slow-wave sleep using different filter configurations in a blinded setting.

Self-applied somnography EOG signals exhibited similar absolute powers but small differences in the morphology of the median PSDs when compared to the PSG EOG signals. Especially at the high-alpha and spindle frequencies (8–15 Hz) during sleep Stages N2 and N3, SAS EOG showed relatively higher peaks in median PSDs when compared to standard PSG EOG. Thus, SAS EOG signals might express more neural activity, e.g., spindles, than standard EOG signals. The bipolar EOG derivations E1-E4 and E2-E3 were more comparable to standard EOG, and therefore, it is recommended to use these as the primary EOG channels, when scoring sleep using the SAS signals. EOG derivations with AFZ as the reference can be used as backup derivations but may contain more frontal EEG activity than the standard PSG EOG signals, as referenced against the M1 and M2 channels. Overall, the differences in absolute powers between SAS and PSG EOG were small, and therefore the signals appear technically comparable. However, further studies comparing the clinically relevant microfeatures, such as spindles and K-complexes, between these two electrode setups are still needed.

This study highlights that developing a reliable and self-applied EEG setup for home-based PSGs is a challenging task that requires rigorous research and development. Self-applied electrodes are usually placed below the hairline, in the frontal/facial region of the head (Korkalainen et al., 2021). Furthermore, the materials and design of the electrode are different from the traditional cup electrodes. Options such as self-adhesive pre-gelled electrodes and different dry electrode materials have been used (Arnal et al., 2020; Carneiro et al., 2020; Miettinen, Myllymaa, Westeren-Punnonen, et al., 2018; Rusanen et al., 2021). These modifications to the standard setup are necessary due to practical reasons and for increasing the ability to self-apply, although they produce another problem; the recorded signals differ in information content from that of the standard PSG (Kainulainen et al., 2021; Rusanen et al., 2021). Therefore, recommendations and notes related to adjusting the AASM scoring rules when scoring sleep using EEG and EOG recorded with the SAS setup are listed in Table 4.

The present study has limitations that need to be addressed before concluding the results. Most importantly, this study was a technical comparison, and the clinical relevance of the results remains unconfirmed. A separate study considering the diagnostic feasibility of the SAS setup and the recommendations following the results of this study should be conducted in the future. Moreover, we did not study the effects of typical symptoms, comorbid medical conditions, and medication of patients with suspected sleep disorders on the quality of EEG and EOG. Due to the study design, we believe this does not affect the obtained results, but further studies investigating the SAS performance in different comorbid diseases and respective medications are warranted. Another limitation is that the amplitude results depend on the used filtering settings. The design of a filter can drastically affect the characteristics of the signal. Therefore, technical differences in the digital and analogue processing of the signals in different PSG software/devices can affect the applicability of the

TABLE 4 Additional recommendations and notes for scoring of sleep when using electroencephalography and electro-oculography signals recorded with the self-applied somnography (SAS) setup (Nox Medical).

Signal type	Derivation	Primary/backup channel	Recommendations/notes	Evidence
EEG	AF4-E3E4	Primary	Lower the Stage N3 sleep peak-to-peak amplitude limit to 40–45 μV depending on the used pre-processing Increasing the cut-off frequency of the high-pass filter to 0.5 Hz can exclude harmful low-frequency artefacts but decreases amplitudes	Figure 3 Table 2 Table 3 Figure 4
	AF3-E3E4	Primary		
	AF7-E3E4	Backup		
	AF8-E3E4	Backup		
EOG	E1-E4	Primary	Increasing the cut-off frequency of the high-pass filter to 0.5 Hz can exclude harmful low-frequency artefacts but decreases amplitudes	Table 2 Table 3 Figure 5 Tables S1 and S2
	E2-E3	Primary		
	E3-AFZ	Backup		
	E2-AFZ	Backup		

Abbreviations: EEG, electroencephalography; EOG, electro-oculography.

present results and recommendations. Moreover, the SAS and PSG signals were recorded on separate devices and had to be synchronised based on device clocks. Small delays can therefore exist between the synchronised recordings but are considered insignificant due to the methodology of the analysis where 30-s segments of the signals are used, and results are averaged over a large number of samples. If we acknowledge these limitations and the differences in the characteristics of the SAS EEG and EOG signals, this study supports the SAS frontal setup being a technically feasible solution for the recording of EEG and EOG in self-applied diagnostic sleep recordings.

AUTHOR CONTRIBUTIONS

Conceptualisation: Matias Rusanen, Erna Sif Arnardottir, Timo Leppänen, Sami Myllymaa, and Samu Kainulainen. *Methodology*: Matias Rusanen, Henri Korkalainen, Heidur Gretarsdottir, Kristin Anna Olafsdottir, Tiina Siilak and Samu Kainulainen. *Analysis*: Matias Rusanen and Samu Kainulainen. *Writing—Original Draft*: Matias Rusanen, Henri Korkalainen, Samu Kainulainen and Timo Leppänen. *Writing—Review and Editing*: all authors.

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CONFLICT OF INTEREST STATEMENT

Erna Sif Arnardottir declares lecture honoraria from Nox Medical, Jazz Pharmaceuticals, Linde Healthcare, Alcoa – Fjardaral, Wink Sleep, Apnimed, and Vistor. She is also a member of the Philips Sleep Medicine and Innovation Medical Advisory Board. All other authors declare no conflict of interest relevant to this study.

DATA AVAILABILITY STATEMENT

The data includes sensitive medical information and therefore is not made publicly available.

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REFERENCES

Altini, M., & Kinnunen, H. (2021). The promise of sleep: A multi-sensor approach for accurate sleep stage detection using the oura ring. *Sensors*, *21*(13), 1–21. https://doi.org/10.3390/s21134302

American Academy of Sleep Medicine. (2016). *Hidden health crisis costing America billions*. 1–25. https://aasm.org/resources/pdf/sleep-apnea-economic-crisis.pdf

American Association of Sleep Technologist. (2012). *Standard polysomnography*. *July*, 1–19. https://www.aastweb.org/Portals/0/Docs/Resources/Guidelines/AAST%20PSG%20Guideline%20Final.pdf

Arnal, P. J., Thorey, V., Debellemaniere, E., Ballard, M. E., Hernandez, A. B., Guillot, A., Jourde, H., Harris, M., Guillard, M., Van Beers, P., Chennaoui, M., & Sauvet, F. (2020). The Dreem headband compared to polysomnography for electroencephalographic signal acquisition and sleep staging. *Sleep*, *43*(11), 1–13. https://doi.org/10.1093/sleep/zsaa097

Arnardottir, E. S., Janson, C., Bjornsdottir, E., Benediktsdottir, B., Juliusson, S., Kuna, S. T., Pack, A. I., & Gislason, T. (2013). Nocturnal

sweating – A common symptom of obstructive sleep apnoea: The Icelandic sleep apnoea cohort. *BMJ Open*, *3*(5), e002795. https://doi.org/10.1136/bmjopen-2013-002795

Arnardottir, E. S., Islind, A. S., Óskarsdóttir, M., Ólafsdóttir, K. A., August, E., Jónasdóttir, L., Hrubos-Strøm, H., Saavedra, J. M., Grote, L., Hedner, J., Höskuldsson, S., Ágústsson, J. S., Jóhannsdóttir, K. R., McNicholas, W. T., Pevernagie, D., Sund, R., Töyräs, J., & Leppänen, T. (2022). The sleep revolution project: The concept and objectives. *Journal of Sleep Research*, *31*(4), 1–8. https://doi.org/10.1111/jsr.13630

Benjafield, A. V., Ayas, N. T., Eastwood, P. R., Heinzer, R., Ip, M. S. M., Morrell, M. J., Nunez, C. M., Patel, S. R., Penzel, T., Pépin, J. L. D., Peppard, P. E., Sinha, S., Tufik, S., Valentine, K., & Malhotra, A. (2019). Estimation of the global prevalence and burden of obstructive sleep apnoea: A literature-based analysis. *The Lancet Respiratory Medicine*, *7*(8), 687–698. https://doi.org/10.1016/S2213-2600(19)30198-5

Berry, R. B., Brooks, R., Gamaldo, C. E., Harding, S. M., Lloyd, R. M., Quan, S. F., Troester, M. M., & Vaughn, B. V. (2020). *The AASM manual for the scoring of sleep and associated events*. https://www.aasmnet.org

Bianchi, M. T., & Goparaju, B. (2017). Potential underestimation of sleep apnea severity by at-home kits: Rescoring in-laboratory polysomnography without sleep staging. *Journal of Clinical Sleep Medicine*, *13*(4), 551–555. https://doi.org/10.5664/jcsm.6540

Bruyneel, M., & Ninane, V. (2014). Unattended home-based polysomnography for sleep disordered breathing: Current concepts and perspectives. *Sleep Medicine Reviews*, *18*(4), 341–347. https://doi.org/10.1016/j.smrv.2013.12.002

Calero, K., & Anderson, W. M. (2016). Home portable sleep testing has gone global. *Journal of Clinical Sleep Medicine*, *12*(1), 7–8. https://doi.org/10.5664/jcsm.5380

Carneiro, M. R., De Almeida, A. T., & Tavakoli, M. (2020). Wearable and comfortable e-textile headband for long-term Acquisition of Forehead EEG signals. *IEEE Sensors Journal*, *20*(24), 15107–15116. https://doi.org/10.1109/JSEN.2020.3009629

Cay, G., Ravichandran, V., Sadhu, S., Zisk, A. H., Salisbury, A. L., Solanki, D., & Mankodiya, K. (2022). Recent advancement in sleep technologies: A literature review on clinical standards, sensors, apps, and AI methods. *IEEE Access*, *10*, 104737–104756. https://doi.org/10.1109/ACCESS.2022.3210518

Collop, N. A., Anderson, W. M., Boehlecke, B., Claman, D., Goldberg, R., Gottlieb, D. J., Hudgel, D., Sateia, M., Schwab, R., & Portable Monitoring Task Force of the American Academy of Sleep Medicine. (2007). Clinical guidelines for the use of unattended portable monitors in the diagnosis of obstructive sleep apnea in adult patients. *Journal of Clinical Sleep Medicine*, *3*(7), 737–747.

Cooksey, J. A., & Balachandran, J. S. (2016). Portable monitoring for the diagnosis of OSA. *Chest*, *149*(4), 1074–1081. https://doi.org/10.1378/chest.15-1076

da Silva Souto, C. F., Pätzold, W., Wolf, K. I., Paul, M., Matthiesen, I., Bleichner, M. G., & Debener, S. (2021). Flex-printed ear-EEG sensors for adequate sleep staging at home. *Frontiers in Digital Health*, *3*(June), 1–10. https://doi.org/10.3389/fdgth.2021.688122

De Zambotti, M., Cellini, N., Goldstone, A., Colrain, I. M., & Baker, F. C. (2019). Wearable sleep technology in clinical and research settings. *Medicine and Science in Sports and Exercise*, *51*(7), 1538–1557. https://doi.org/10.1249/MSS.0000000000001947

Ferber, R., Millman, R., Coppola, M., Fleetham, J., Murray, C. F., Iber, C., Mccall, V., Nino-murcia, G., Pressman, M., Sanders, M., Strohl, K., Bernhard, V., & Williams, A. (1994). ASDA standards of practice portable recording in the assessment of obstructive sleep apnea. *Sleep*, *17*(4), 378–392. https://doi.org/10.1093/sleep/17.4.378

Fischer, J., Dogas, Z., Bassetti, C. L., Berg, S., Grote, L., Jennum, P., Levy, P., Mihaicuta, S., Nobili, L., Riemann, D., Puertas Cuesta, F. J., Raschke, F., Skene, D. J., Stanley, N., & Pevernagie, D. (2012). Standard procedures for adults in accredited sleep medicine centres in Europe. *Journal of Sleep Research*, *21*(4), 357–368. https://doi.org/10.1111/j.1365-2869.2011.00987.x

Ghegan, M. D., Angelos, P. C., Stonebraker, A. C., & Gillespie, M. B. (2006). Laboratory versus portable sleep studies: A meta-analysis. *Laryngoscope*, *116*(6), 859–864. https://doi.org/10.1097/01.mlg.0000214866.32050.2e

Hillman, D., Mitchell, S., Streatfeild, J., Burns, C., Bruck, D., & Pezzullo, L. (2018). The economic cost of inadequate sleep. *Sleep*, *41*(8), 1–13. https://doi.org/10.1093/sleep/zsy083

Hirshkowitz, M. (2016). Polysomnography challenges. *Sleep Medicine Clinics*, *11*(4), 403–411. https://doi.org/10.1016/j.jsmc.2016.07.002

Jónsson, S. Æ., Gunnlaugsson, E., Finssonn, E., Loftsdóttir, D. L., Ólafsdóttir, G. H., Helgadóttir, H., & Ágústsson, J. S. (2020). RESTNET: A robust end-to-end DEEP LEARNING approach to sleep staging of self applied SOMNOGRAPHY studies. *Sleep*, *43*, A171. https://doi.org/10.1093/sleep/zsaa056.444

Jose, M., Santiago, J., Alvar, G. N., Marin, J. M., Carrizo, S. J., Vicente, E., & Agusti, A. G. N. (2005). Long-term cardiovascular outcomes in men with obstructive sleep apnoea-hypopnoea with or without treatment with continuous positive airway pressure: An observational study. *Lancet*, *365*(9464), 1046–1053. https://doi.org/10.1016/S0140-6736(05)71141-7

Kainulainen, S., Korkalainen, H., Sigurdardottir, S., Myllymaa, S., Serwatko, M., Sigurdardottir, S. P., Clausen, M., Leppanen, T., & Arnardottir, E. S. (2021). Comparison of EEG signal characteristics between polysomnography and self applied Somnography setup in a pediatric cohort. *IEEE Access*, *9*, 110916–110926. https://doi.org/10.1109/ACCESS.2021.3099987

Kalevo, L., Miettinen, T., Leino, A., Kainulainen, S., Korkalainen, H., Myllymaa, K., Töyräs, J., Leppänen, T., Laitinen, T., & Myllymaa, S. (2020). Effect of sweating on electrode-skin contact impedances and artifacts in EEG recordings with various screen-printed Ag / AgCl electrodes. *IEEE Access*, *8*, 50934–50943. https://doi.org/10.1109/ACCESS.2017

Kalevo, L., Miettinen, T., Leino, A., Westeren-Punnonen, S., Sahlman, J., Mervaala, E., Toyras, J., Leppanen, T., Myllymaa, S., & Myllymaa, K. (2022). Self-applied electrode set provides a clinically feasible solution enabling EEG recording in home sleep apnea testing. *IEEE Access*, *10*, 60633–60642. https://doi.org/10.1109/ACCESS.2022.3178189

Kappenman, E. S., & Luck, S. J. (2010). The effects of electrode impedance on data quality and statistical significance in ERP recordings. *Psychophysiology*, *47*(5), 888–904. https://doi.org/10.1111/j.1469-8986.2010.01009.x

Korkalainen, H., Nikkonen, S., Kainulainen, S., Dwivedi, A. K., Myllymaa, S., Leppänen, T., & Töyräs, J. (2021). Self-applied home sleep recordings: The future of sleep medicine. *Sleep Medicine Clinics*, *16*(4), 545–556. https://doi.org/10.1016/j.jsmc.2021.07.003

Leino, A., Korkalainen, H., Kalevo, L., Nikkonen, S., Ahlberg, J., Sahlman, J., & Miettinen, T. (2022). Deep Learning enables accurate automatic sleep staging based on ambulatory forehead EEG. *IEEE Access*, *10*, 26554–26566. https://doi.org/10.1109/ACCESS.2022.3154899

Levendowski, D. J., Ferini-Strambi, L., Gamaldo, C., Cetel, M., Rosenberg, R., & Westbrook, P. R. (2017). The accuracy, night-to-night variability, and stability of frontopolar sleep electroencephalography biomarkers. *Journal of Clinical Sleep Medicine*, *13*(6), 791–803. https://doi.org/10.5664/jcsm.6618

Lim, W. Y., & See, K. C. (2021). Predictors of obstructive sleep apnea misclassification when using total bed time versus total sleep time. *Scientific Reports*, *11*(1), 1–7. https://doi.org/10.1038/s41598-021-90818-y

Luyster, F. S., Kip, K. E., Buysse, D. J., Aiyer, A. N., Reis, S. E., & Strollo, P. J. (2014). Traditional and nontraditional cardiovascular risk factors in comorbid insomnia and sleep apnea. *Sleep*, *37*(3), 593–600. https://doi.org/10.5665/sleep.3506

Mattila, T., Hasala, H., Kreivi, H.-R., Avellan-Hietanen, H., Bachour, A., Herse, F., Leskelä, R.-L., Toppila-Salmi, S., Erhola, M., Haahtela, T., & Vasankari, T. (2022). Changes in the societal burden caused by sleep apnoea in Finland from 1996 to 2018: A national registry study. *The Lancet Regional Health – Europe*, *16*, 100338. https://doi.org/10.1016/j.lanepe.2022.100338

Miettinen, T., Myllymaa, K., Muraja-Murro, A., Westeren-Punnonen, S., Hukkanen, T., Töyräs, J., Lappalainen, R., Mervaala, E., Sipilä, K., & Myllymaa, S. (2018). Screen-printed ambulatory electrode set enables accurate diagnostics of sleep bruxism. *Journal of Sleep Research*, *27*(1), 103–112. https://doi.org/10.1111/jsr.12536

Miettinen, T., Myllymaa, K., Westeren-Punnonen, S., Ahlberg, J., Hukkanen, T., Toyras, J., Lappalainen, R., Mervaala, E., Sipila, K., & Myllymaa, S. (2018). Success rate and technical quality of home polysomnography with self-applicable electrode set in subjects with possible sleep bruxism. *IEEE Journal of Biomedical and Health Informatics*, *22*(4), 1124–1132. https://doi.org/10.1109/JBHI.2017.2741522

Punjabi, N. M., Brown, T., Aurora, R. N., Patel, S. R., Stosor, V., Cho, J. H.-J., Helgadóttir, H., Ágústsson, J. S., D'Souza, G., & Margolick, J. B. (2022). Methods for home-based self-applied polysomnography: The multicenter AIDS cohort study. *SLEEP Advances*, *3*(1), 1–8. https://doi.org/10.1093/sleepadvances/zpac011

Rusanen, M., Huttunen, R., Korkalainen, H., Myllymaa, S., Töyräs, J., Myllymaa, K., Olafsdottir, K. A., Leppänen, T., Arnardottir, E. S., & Kainulainen, S. (2023). Generalizable deep learning-based sleep staging approach for ambulatory textile electrode headband recordings. *IEEE Journal of Biomedical and Health Informatics*, *27*, 1869–1880. https://doi.org/10.1109/JBHI.2023.3240437

Rusanen, M., Kainulainen, S., Korkalainen, H., Kalevo, L., Myllymaa, K., Leppanen, T., Toyras, J., Arnardottir, E. S., & Myllymaa, S. (2021). Technical performance of textile-based dry forehead electrodes compared with medical-grade overnight home sleep recordings. *IEEE Access*, *9*, 157902–157915. https://doi.org/10.1109/access.2021.3128057

Zinchuk, A. V., Jeon, S., Koo, B. B., Yan, X., Bravata, D. M., Qin, L., Selim, B. J., Strohl, K. P., Redeker, N. S., Concato, J., & Yaggi, H. K. (2018). Polysomnographic phenotypes and their cardiovascular implications in obstructive sleep apnoea. *Thorax*, *73*(5), 472–480. https://doi.org/10.1136/thoraxjnl-2017-210431

SUPPORTING INFORMATION

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