Delft University of Technology

From lab to living room: evaluating the signal quality of water-based EEG

A study of EEG cap performance in controlled and uncontrolled settings

Master thesis

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TU Delft

From lab to living room: evaluating the signal quality of water-based EEG

A study of EEG cap performance in controlled and uncontrolled settings

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TU Delft

Preface

This master thesis concludes my academic journey at the TU Delft. It represents my interests in healthcare, data analysis, research, the mysterious organ that is the brain and finally, (working with) people. Being able to perform not only one, but two experiments myself was one of my favourite parts of this master research. I also greatly enjoyed being able to incorporate three different coding languages into the project.

I would like to thank Mark and Arthur for supervising me throughout my thesis. Thank you both for taking the time out of your busy schedules to meet with me and thank you for encouraging me to keep asking for help. Thanks for all of the feedback rounds, and for the enthusiasm for this project.

Furthermore, I would especially like to thank Jamie, for our 'quick' talks in the lab that helped me make choices about the project that otherwise I still would've doubted about. Also a big thank you for the talks about after my graduation. I am certain I will end up with a great first job, thanks to your inspiration and motivation. And I will have to, since you will watch me closely on LinkedIn.

Additionally, I would also like to thank everyone that participated in my experiments and that had to either wash all of the gel out of their hair in a washbasin on campus, or let me into their homes and take over their entire dining table or desk. Or both.

I want to express my appreciation to the NMClab master students, for their help during all of the Monday lunch meetings. For giving their advice on how to perform a pilot experiment or a certain figure, but also for hearing me out about my cat or what I did that weekend.

I extend my gratitude to dr. Selene Pirola for being part of my graduation committee.

Finally, I wish to thank my family and friends for their support during my studies. A special thanks to Karien and Elise for the coffee breaks that seemed to never end. Thanks to Martijn for reading the very first version with no idea about brains or EEGs. Thanks to Marloes for the online support that spans even 80 kms. Thanks to Maaike for always remembering what I struggled with, thanks to Indira for the mutual understanding, thanks to Annelot for your critical feedback and thanks to Joris for a nice thesis-exchange. Also thanks to Jayanti and Paul for accidentally being perfect models. Moreover, thanks to Alex for enduring my rants even though you were also in the middle of a graduation project, for pushing me to get out of my comfort zone and for always being there.

I really enjoyed the past few months and I look forward to what the future brings!

Jennifer Juch
Utrecht, September 2024

Abstract

Scalp electroencephalography (EEG) is a widely used, noninvasive tool to assess brain activity. EEG is valuable for various neurological conditions like epilepsy and migraine. Conventional EEG uses gel-based electrodes, ensuring good signal quality but requiring complex setup and the removal of patients from their natural environments. Portable EEG devices with water electrodes offer easier home measurements but pose signal quality concerns. This thesis aims to evaluate the signal quality of water electrodes compared to gel electrodes and investigate the feasibility of home-based EEG measurements.

Three experimental protocols were used: resting-state (RS), single visual evoked potentials (SVEP) and 12 Hz steady-state visual evoked potentials (SSVEP). These signals were measured 1) in the lab, once with a gel cap and once with a water cap; and 2) at home with a water cap. Signal quality was assessed with the artefact proportion, the signal-to-noise ratio (SNR), the relative 12 Hz SSVEP power, the relative alpha band eyes-closed RS power, the presence of the Berger effect, and the SVEP waveforms.

In the lab setting, the water and gel cap showed similar signal quality as illustrated by a similar SNR, relative alpha power, alpha band presence and SVEP waveform. However, an increase in artefacts and slight decrease in relative 12 Hz power and SVEP amplitude show remaining shortcomings of the water cap compared to the gel cap. When comparing the water cap between lab and home settings, the performance closely matches. This is demonstrated by the similar SNR, relative 12 Hz power, alpha presence and SVEP waveform and amplitude. Differences were a decrease in artefacts and an increase in relative alpha band power for the signal measured at home.

Provided that the limitations of the water cap can be mitigated by further developments, the otherwise relatively comparable signal quality between the gel and water caps suggests that water-based EEG systems could be a viable alternative to traditional gel-based systems. Furthermore, the positive home study results suggest that home-based EEG measurements could be a viable alternative to lab-based studies with the help of a water electrode EEG cap.

Contents

Preface i

Abstract ii

1 Introduction 1

2 Materials and Methods 3
- 2.1 Participants . . . 3
- 2.2 Experimental setup . . . 3
- 2.3 Experimental protocol . . . 4
- 2.4 Data preprocessing and analysis . . . 5
  - 2.4.1 EEG preprocessing . . . 5
  - 2.4.2 Main outcomes . . . 6
  - 2.4.3 Statistical analysis . . . 7

3 Results 8
- 3.1 Participants . . . 8
- 3.2 Signal quality metrics . . . 8
  - 3.2.1 Artefact proportion . . . 8
  - 3.2.2 SNR . . . 9
  - 3.2.3 Relative 12 Hz power . . . 10
  - 3.2.4 Relative alpha band power . . . 10
  - 3.2.5 Alpha band presence . . . 11
  - 3.2.6 SVEP presence . . . 11

4 Discussion and Conclusion 12
- 4.1 Interpretation of results . . . 12
  - 4.1.1 Artefact proportion . . . 12
  - 4.1.2 SNR . . . 13
  - 4.1.3 Relative 12 Hz power . . . 13
  - 4.1.4 Relative alpha band power . . . 14
  - 4.1.5 Alpha band presence . . . 14
  - 4.1.6 SVEP presence . . . 14
- 4.2 Limitations . . . 15
- 4.3 Future recommendations . . . 15

A Supplementary results 21

List of Figures

2.1 Experimental setup 4
2.2 10-20 Electrode placement. 4
2.3 Preprocessing pipeline 5
2.4 Schematic overview of the three protocol types: length of the data, creating epochs and resulting data. 6

3.1 Signal of one participant before and after artefact deletion 9
3.2 Signal-to-noise ratio (SNR) and impedance 9
3.3 Relative 12 Hz power, PSD and violin plot 10
3.4 Relative alpha power, PSD and violin plot 10
3.5 SVEP response of two participants. 11

A.1 Deviant artefacted signal for one participant. 22
A.2 SNR for every participant. 22
A.3 The four signals with the highest and lowest SNR 23
A.4 SNR for all three occipital channels. 23
A.5 RS PSD plots for participants corresponding to SVEP plots. 23

List of Tables

2.1 Inclusion criteria 3

3.1 Participant characteristics 8
3.2 Overview of results 11

1

Introduction

Scalp electroencephalography (EEG) is a technique to assess brain function used for a plethora of applications in both clinical and research settings. EEG quantifies brain activity by recording the electric current contributions generated by neuronal activity through electrodes placed on the surface of the scalp [1]. This method is a commonly used, noninvasive, cost-effective neuroimaging technique [2, 3]. In clinical settings, scalp EEG is often used to monitor and evaluate abnormal brain activity. For instance, it serves as a diagnostic as well as evaluation and management tool for patients with epilepsy [4, 5]. A comprehensive epilepsy center may perform >4000 EEG studies per year [6]. Since epilepsy and migraine patients share pathophysiological mechanisms such as epileptiform discharges, EEG can be valuable for migraine patients as well [7]. It is primarily valuable in the ongoing research for a biomarker to improve characterisation of migraine attacks and to identify novel drug targets [8, 9].

The conventional approach to measuring EEG signals in a clinical environment uses electrodes that are in contact with the scalp through electrolytic gel. The gel bridges the ionic current flow from the scalp and the electron flow in the electrode and it increases adhesion of the electrode to the scalp [10]. This ensures good signal quality and lessens the susceptibility to artefacts. However, conventional clinical gel EEG measurements are not comfortable for patients, require a long set-up procedure and a technician to apply the gel [11]. This approach also removes them from their natural environment, which is inconvenient for the patient and might influence migraine attack susceptibility [8]. Often, hospital EEG recordings allow measurements of short duration only, or require costly resources such as 24-hour nursing, beds and meals [12]. At home EEG measurements offer convenient and comfortable measurements, lower hospital costs and can allow for a continuous measurement, increasing the change of correctly identifying predictive patterns [5, 13, 14].

The step towards measuring EEG at patient's homes has come closer with the emergence of small, user-friendly portable EEG devices with dry or water electrodes [15, 16]. Dry electrodes demonstrate short setup time and increased patient comfort because they eliminate the need for gel application [17, 18]. It is even stated that dry electrodes can be used with comparable quality to gel electrodes [19]. However, more studies show that dry electrodes demonstrate a decrease in signal quality partly due to a significant contribution of noise and a higher susceptibility to movement artifacts [20–24]. Water electrodes use saline water or an electrolyte liquid as conductive solution, and thus they offer the short setup time, increased patient comfort and user-friendliness of dry electrodes, while also retaining some of the signal quality [15, 25]. However, it was also concluded that water electrodes contain a higher noise distribution in the signal than gel electrodes [26]. Since there is a wide variety of portable water-based EEG devices, their potential seems promising but a generalization of signal quality remains difficult [16].

Aside from the signal quality that portable wet electrode EEG devices offer, an uncontrolled home environment might also impact the quality of the EEG measurements. Specifically, in a home environment, the magnitude and frequency of artefacts tend to increase, possibly due to experimental error or subject motion [14, 27, 28]. In one study for instance, half of the home EEG recordings had to be deleted due

to artefacts [29]. In another, home recordings were reported to be of inferior quality to recordings in a clinical environment [30]. The authors of the latter study hypothesised the absence of both medical-technical assistance and clinical conditions to be the cause of reduced quality [30]. On the other hand, portable home-used EEG devices have been reported to successfully collect comparable EEG to conventional clinically-collected EEG as well [15, 31–34]. Therefore, home measurements prove to be promising, provided that the mobile EEG device used grants EEG of sufficient signal quality and that the quality is thoroughly evaluated [8].

This signal quality evaluation should encompass factors relevant to both clinical and research settings. In these settings, EEG is evaluated in the time and frequency domain. Therefore, not only artefacts but also noise which impacts the frequency distribution of the signal can have a negative impact on epilepsy diagnoses or the characterisation of migraines [5, 8]. Thus, it is important for new mobile EEG devices to be assessed on multiple aspects of the signal they produce. Metrics like the number of artefacts, the signal-to-noise ratio (SNR) or the strength of single visual-evoked potentials (SVEPs) can indicate the quality of the signal in the time domain [35–38]. Whereas the power of a steady-state visual evoked potential (SSVEP) on the stimulation frequency, or the resting-state (RS) eyes-closed power around the alpha band can give signal quality insights in the frequency domain [39–42].

Home EEG measurements are desired because they offer convenient, comfortable and continuous measurements. Fortunately, they can be realised with the emergence of portable EEG devices, such as those with water electrodes. However, a generalization of the signal quality offered by these devices remains difficult. To be able to correctly interpret the signal that results from home measurements, it is important to optimise signal quality and minimize the amount of artefacts in the EEG signal resulting from water electrode EEG caps. Therefore, the aim of this thesis is to determine the signal quality of EEG signals measured with a water electrode EEG cap, compared to a conventional gel electrode EEG cap. That same water cap will then be evaluated in a home environment to compare its performance to the that of the water cap in the controlled environment of the first study.

2

Materials and Methods

2.1. Participants

Participants were recruited through the network of the author (e.g. WhatsApp and study association Variscopic). Inclusion criteria for participants included age, head circumference and health requirements. The exact requirements can be found in Table 2.1. Prior to the EEG measurements, each subject was asked to read the participant information and all participants provided consent to the experiment. The experimental protocol was approved by the Human Research Ethics Committee of the TU Delft on 17-04-2024 (HREC 4068).

2.2. Experimental setup

All EEG recordings of the lab study were performed in a sound-proof room in the EEGlab on campus (Faculty of Mechanical Engineering, TU Delft). Figure 2.1 shows the experimental setup in a schematic diagram where all materials are connected, and in pictures displaying the experiment in the lab and at home. Two different caps were used for the experiments; a gel cap and a water cap (Infinity Gel Headcap and Infinity Water Headcap, TMSi, Oldenzaal, the Netherlands). For both caps, 32 Ag/AgCl electrodes were used. The electrodes were arranged according to the 10-20 system (Figure 2.2). The gel cap required gel to be applied to the head of the participants to assure good electrode-scalp contact. The water cap required sponges saturated with a salt water solution to be applied to the cap. The EEG headcap was connected to the amplifier (APEX, TMSi, Oldenzaal, The Netherlands) which sampled the data at a frequency of 1000 Hz. A ground electrode was also connected to the amplifier, which was secured to the participant through either a wristband (water cap) or a stick-on electrode positioned on

Inclusion topic	Criterion
Age	$\geq$ 18 years
Head circumference	$\geq$ 54 cm
Mental disorders	No
• Severe depression	
• Panic disorders	
• Schizophrenia or psychiatric disorders	
Epileptic disorders	No
Severe visual impairment	No
Malignancy in medical history	No
Periodic pain attacks	No
Migraine or trigeminal autonomic cephalgia	No personal or first-degree family history
Tension-type headaches	No more than one in three months

Table 2.1: Inclusion criteria

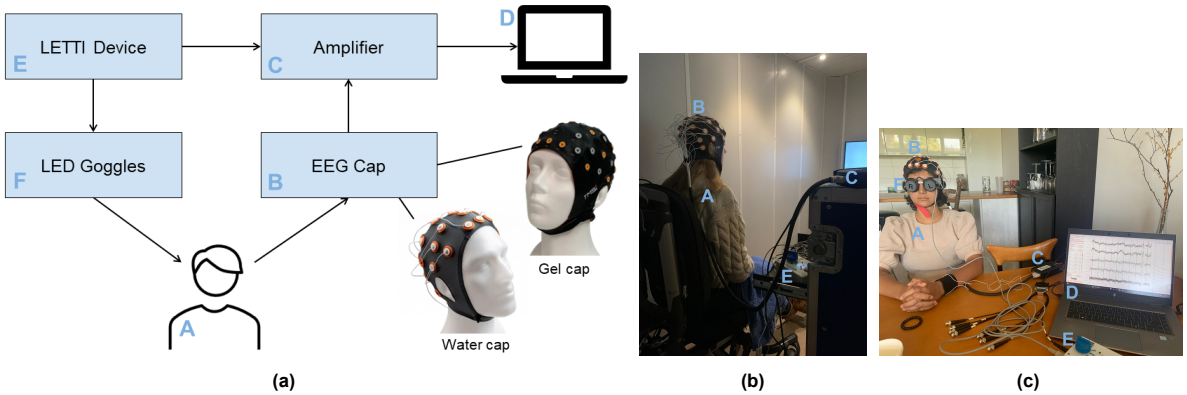

Figure 2.1: Experimental setup: (a) Schematic diagram illustrating the connections between the participant, EEG cap, LED goggles, LETTI device and amplifier. The LED goggles, controlled by the LETTI device, provide visual stimuli while the EEG cap (gel or water) transmits brain signals to the APEX amplifier. Event triggers corresponding to the visual stimuli are also recorded by the amplifier, which is connected to a laptop for data acquisition. (b) Real-life example showing the setup in a sound-proof room at the EEGlab, collecting resting-state data. (c) Real-life example collecting data with visual stimulation at a participant's home.

the wrist (gel cap).

For visual stimulation in order to collect the SVEP and SSVEP data, each participant wore LED goggles with red LEDs (Synergy Plinth; Medelec International, Pleasanton, CA, USA) which were connected to a stimulating device (LETTI, TU Delft, The Netherlands). The LED goggles were fixed to the EEG cap with tape to assure safe fixation. The LETTI device controlled the LED goggles corresponding to the experimental protocol. The LETTI device was connected to the APEX amplifier to record event triggers at the times of the flashes.

The materials used in the home study were equal to those in the lab study, except for the screen portraying a cross being replaced by a piece of paper.

2.3. Experimental protocol

Before recording the EEG data, an impedance measurement was performed once. A maximum of 15 minutes was spent to reach an impedance value lower than $10k\Omega$. The EEG data were then recorded for three different experimental conditions to collect three different kinds of data (RS, SVEP and SSVEP). To mitigate the effect of fatigue, the order of these conditions was randomized for every participant [44]. The procedures for each experimental condition are described below.

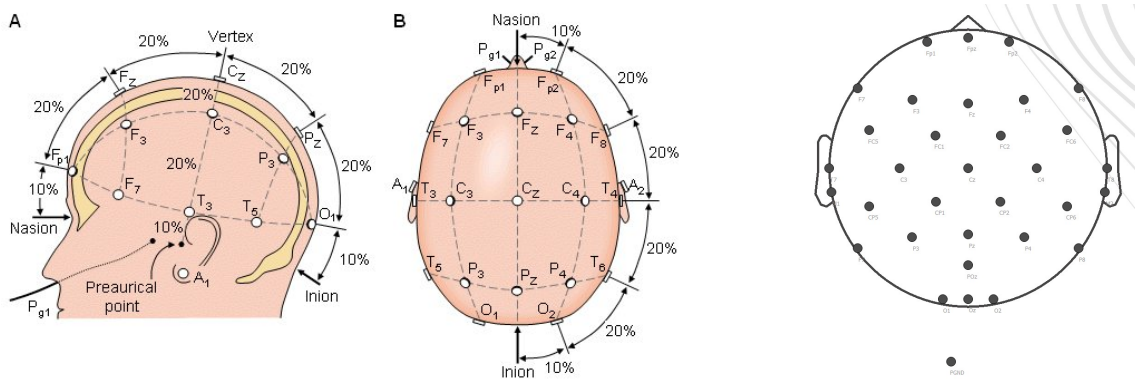

Figure 2.2: 10-20 Electrode placement. (a) Anatomical guide including the Nasion and Inion, adapted from [43]; (b) Digital guide as it appeared during the experiments.

- During the six minute RS EEG recording, each participant was asked to open their eyes for one minute and then close their eyes for one minute and repeat this three times, for a total of six minutes. The LETTI device indicated when one minute had passed with an auditory stimulus. When the participant opened their eyes, they were instructed to look at an 'X' on a screen in front of them, to limit eye movement artefacts.
- During the four-minute SSVEP recording, visual stimulation was applied through the goggles with a frequency of 12 Hz. This frequency was chosen to not interfere with the alpha or beta band already present. Ten blocks of ten seconds of visual stimulation were applied. In between the stimulation blocks, there was a rest period where no stimulation was applied. The duration of the rest period was randomized for each block between 10 and 15 seconds.
- During the five minute SVEP recording, continuous visual stimulation was applied through the LED goggles. The time between each flash was randomized between 700 and 900 ms and a total of 300 flashes was applied.

2.4. Data preprocessing and analysis

2.4.1. EEG preprocessing

All data was processed and analyzed using MATLAB R2022b (The Mathworks, Inc., Natick, MA, USA). A schematic overview of the preprocessing pipeline can be found in Figure 2.3. Figure 2.4 shows the epoching process of all data types.

All data were filtered with a Finite Impulse Response (FIR) filter using signal processing toolbox EEGlab [45]. The filter cut-off parameters varied per experimental protocol.

- To minimize non-neural low-frequency drift, a high-pass filter with a cutoff at 0.5 Hz was implemented for the RS data. A low-pass filter of 100 Hz was applied to eliminate high-frequency noise, since the highest EEG frequency band does not reach above 100 Hz. To reduce the effect of line noise, a 50 Hz and 100 Hz notch filter were also used. Afterwards, six events were extracted, of which three eyes open and three eyes closed events. The signal was then epoched into 5s segments. To eliminate eye opening/closing artefacts, the first and last epoch of every event was deleted. This produced 30 epochs of 5s for both types of data.
- The SVEP data were filtered between 0.1 Hz to minimize drift and 40 Hz to accentuate recognizable SVEP components and minimize the effect of (line) noise. Each event indicated the brain response to a single flash resulting in 300 events in total. The signal was then epoched into 300 segments of 0.8s, from 0.1s before the flash to 0.7s after.
- A high-pass filter with a cutoff at 1 Hz was used for the SSVEP data. To retain as much information as possible, a low-pass filter with a cutoff at 300 Hz was implemented. To minimize the effect of line noise and its harmonic frequencies, notch filters at 50, 100, 150, 200, 250 and 300 Hz were applied to the data. Afterwards, 10 events were created; each event represented one 10s block of flashes. Before epoching, the first 2 seconds of every 10s block were deleted to eliminate any stimulation onset artefacts. Then, the remaining 10 times 8s data were epoched into 20 4s segments.

For all data types, flat epochs were deleted and epochs which contained artefacts were detected and deleted manually. This was done with the help of an SD threshold. The $mean \pm 3 \cdot SD$ was shown to aid the researcher in manually detecting artefacts. Epochs with samples excessively exceeding

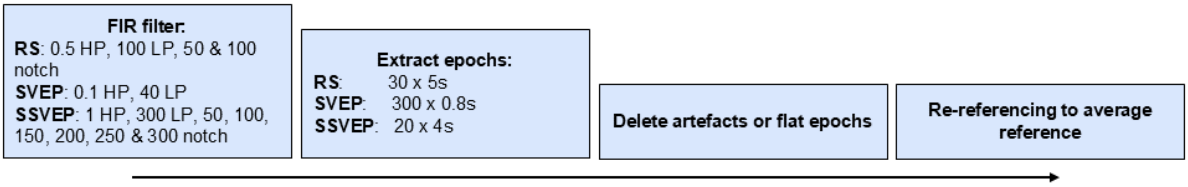

Figure 2.3: Preprocessing pipeline, from left to right. FIR = finite impulse response, RS = resting state, SVEP = single visual evoked potential, SSVEP = steady-state visual evoked potential, HP = high-pass, LP = low-pass.

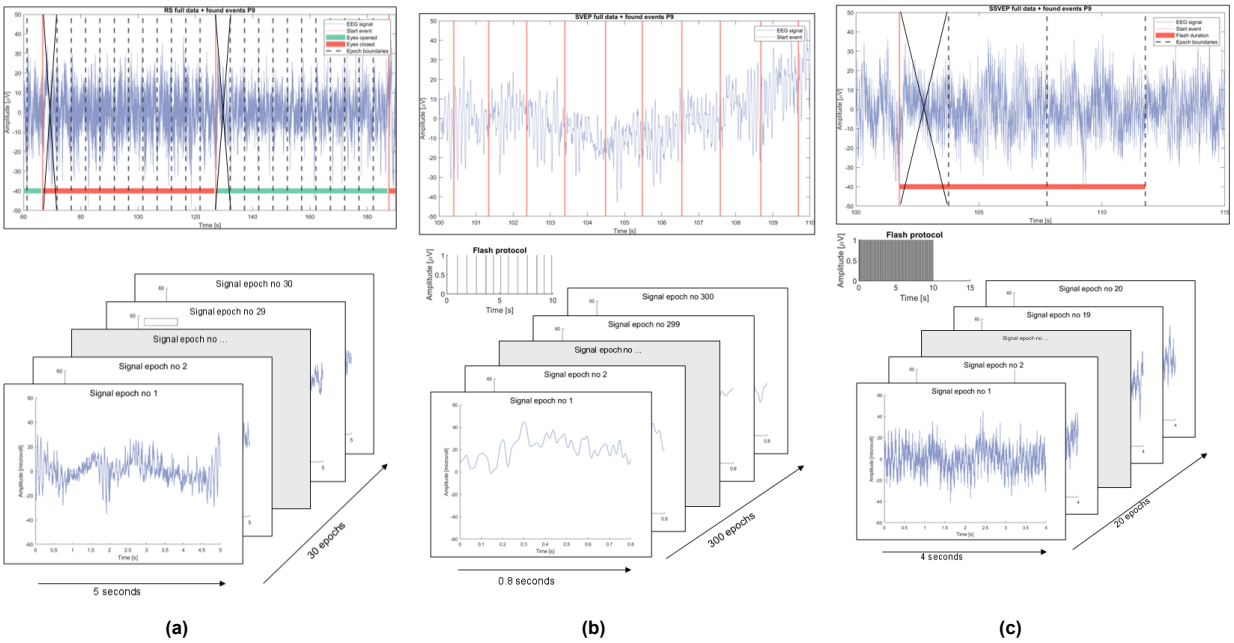

Figure 2.4: Schematic overview of the three protocol types: length of the data, creating epochs and resulting data. (a): resting-state, 180 seconds eye opened and 180 seconds eye closed data total. (b): single visual evoked potential, 300 seconds data total. (c): steady-state visual evoked potential, 100 seconds data total.

the threshold or epochs with a big fraction of samples exceeding the threshold were marked as artifact-contaminated and were deleted accordingly. If a channel retained less than 20% of epochs after this detection step, the entire channel was removed from further analysis. Finally, the data was re-referenced to an average reference.

2.4.2. Main outcomes

The EEG signal quality was assessed through various signal quality metrics. The number of artifact-contaminated epochs was extracted from all data types. From the SSVEP data, a signal-to-noise ratio (SNR) and relative 12 Hz power were calculated. From the RS data, a relative alpha band power was calculated and the alpha band presence was detected. Finally, from the SVEP data, an averaged SVEP waveform response was formed.

Artefacts

The number of epochs that were deleted following the artefact detection were noted for every channel. Since all following outcome measures are based on the signal from the occipital channels (O1, Oz, O2), the number of deleted epochs due to artefacts for these channels were averaged per participant.

SNR

To define the ratio between signal and noise on the EEG signal during visual stimulation, the signal-to-noise ratio (SNR) was calculated on the SSVEP signal. The SSVEP signal, $x(k)$, was averaged over all epochs (P) to reach a steady-state response (SSR, $\hat{x}(k)$). The SNR was then calculated for every channel by dividing the SSR power by the variance across all epochs:

$$\hat{x}(k) = \frac{1}{P}\sum_{p=1}^{P} x^{[p]}(k) \tag{2.1}$$

$$SNR = \frac{\hat{E}_x}{\hat{\sigma}_x^2} = \frac{\sum_{k=1}^{N} \hat{x}(k)^2}{\sum_{k=1}^{N} \frac{1}{P-1} \sum_{p=1}^{P} (x^{[p]}(k) - \hat{x}(k))^2} \tag{2.2}$$

This formula was adapted from Vlaar et al. [46].

Furthermore, the impedance data was presented alongside the SNR results to explore any potential relationship between the two measures.

Relative 12 Hz power

To be able to determine how well the brain's response to a 12 Hz stimulation can be recorded, a relative 12 Hz power was calculated. To reach this quality metric, first the power spectral density of the epochs was formed. This was done with the Matlab function *pwelch*, with a Hanning window and 50% overlap. The relative 12 Hz power was calculated according to the formula described by Tautan et al. [47, 48]:

$$\textit{Relative power} = \frac{mean(PSD_{\textit{band of interest}})}{mean(PSD_{\textit{signal band}-\textit{band of interest}})} \tag{2.3}$$

The band of interest was defined as the stimulation frequency and its harmonic frequency: 11-13 and 23-25 Hz. The signal band was chosen as the same band taken by Tautan et al., between 4-30 Hz.

Relative alpha band power

Similar to the relative 12 Hz power, the relative alpha band power was calculated on the power spectral density of the RS data. Equation 2.3 was used with a band of interest around the alpha band of 8-12 Hz, and the same signal band of 4-30 Hz was chosen. The relative alpha band power was calculated on the eyes closed data only, since the alpha wave is only expected when the subject closes their eyes.

Alpha band presence

For every participant, it was determined whether there was a significant difference in alpha band power between the eyes closed and eyes opened conditions. The number of participants with this distinguishable Berger effect were counted. For both caps and the home environment, a significant participant fraction was calculated, by dividing the number of participants with a Berger effect by the total number of participants.

SVEP presence

To determine the ability to produce a distinguished single-flash brain response in the time domain, average SVEP waveforms were produced. This was done by averaging the signal over all remaining epochs for every channel. Then, for every participant, the response was averaged over the three occipital channels.

2.4.3. Statistical analysis

The number of artefacts, SNR, relative 12 Hz power and relative alpha power were tested for normal distribution by analysing histograms, q-q plots and by using the Anderson-Darling test. In the case of normally divided data, these metrics were compared between gel and water cap with a paired t-test through the *ttest* function in Matlab. If the data was not normally divided, the metrics were compared between caps with the non-parametric Wilcoxon signed rank test, using the *signrank* function. Normally distributed data would be presented as the mean $\pm$ standard deviation (SD). Non-normally distributed data would be presentated as median $\pm$ interquartile range (IQR; 25th and 75th percentiles).

These same metrics were compared between the lab and home environment using a linear mixed-effects model to account for hybrid paired/unpaired data, because participants that took part in the lab study might have also participated in the home study, but this was not obligatory. The home data was set as the dependent variable, the location (lab or home) as the independent variable and the participant numbers as cluster variable. The Matlab function *lme* was used.

All tests were performed using a two-tailed significance level of 95% (α = 0.05).

3

Results

3.1. Participants

Seventeen healthy participants participated in the lab study (nine female, median age $\pm$ interquartile range (IQR) (years) 24 $\pm$ 3 yrs, see Table 3.1). Three participants wore the large size cap and 14 participants wore the medium size cap. Twenty healthy participants participated in the home study (8 female, median age $\pm$ IQR (years) 24 $\pm$ 2 yrs, see Table 3.1). Four participants wore the large size cap and 16 participants wore the medium size cap. Of the 17 people participating in the lab study, 9 participants also took part in the home study.

Characteristics	Lab study (n=17)	Home study (n=20)
Age, years, median $\pm$ IQR	24 $\pm$ 3	24 $\pm$ 2
Gender, female, $n(\%)$	9 (53%)	8 (40%)
Cap size, m, $n(\%)$	14 (82%)	16 (80%)

Table 3.1: Participant characteristics for both lab and home study. IQR = interquartile range.

3.2. Signal quality metrics

An overview of all signal quality metric results can be found in Table 3.2. All data was found to be not normally divided, therefore all metrics were compared between the gel and water cap using the non-parametric Wilcoxon signed rank test.

3.2.1. Artefact proportion

Figure 3.1 shows an artifact-contaminated signal before and after the artifact was deleted. Three participants showed a highly deviant signal only for the water cap in the lab. An example epoch of this signal can be found in Figure A.1. There appeared to be a 100% artefacted signal with no way of retrieving any underlying EEG signals. Therefore, all epochs were deleted for these participants.

There was no significant difference in the number of epochs with artefacts between the gel cap and the water cap measured in the lab for the RS and SVEP data. However, there was a significant higher number of epochs with artefacts with the water cap for the SSVEP data (median $\pm$ IQR, SSVEP gel: 0 $\pm$ 0%, water: 10 $\pm$ 37.5%, p=0.008, r=-0.74; RS gel: 7.14 $\pm$ 11.6%, water: 7.14 $\pm$ 33%, p=0.06, r=-0.5; SVEP gel: 0 $\pm$ 0.92%, water: 4.33 $\pm$ 8.75%, p=0.09, r=-0.47). There was no significant difference in artefacted epochs between the water caps measured in the lab and at home setting for the RS and SVEP data. However, there was a significant higher number of epochs with artefacts in the lab for the SSVEP condition (SSVEP home: 0 $\pm$ 5%, p=0.004; RS home: 10.7 $\pm$ 16%, p=0.1; SVEP home: 4 $\pm$ 9.2%, p=0.72).

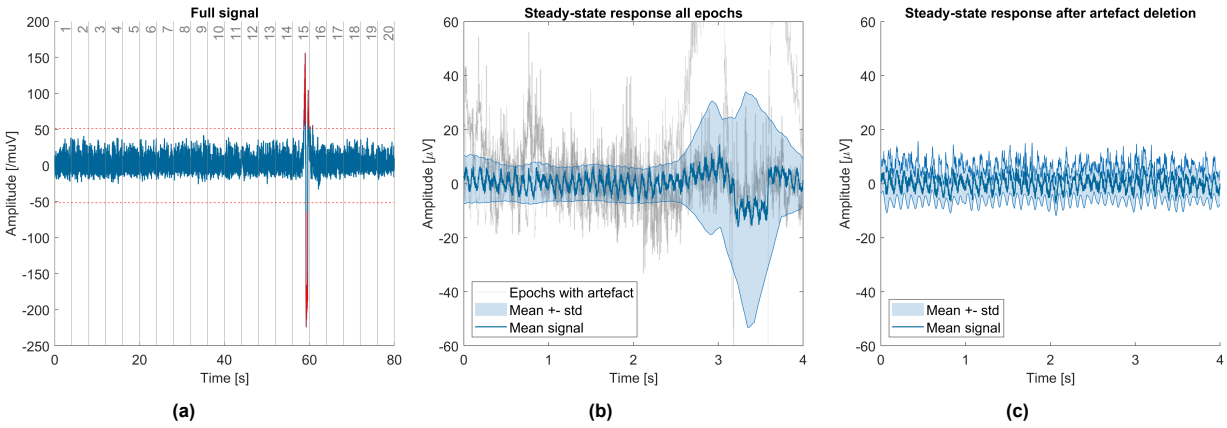

Figure 3.1: Signal before and after artefact deletion. (a) the full signal, all epochs. The artefact detection guideline is shown with the red dashed line; (b) signal averaged over all epochs (steady-state response, SSR); (c) the steady-state response after artifact-contaminated epochs have been deleted.

3.2.2. SNR

The SNR for the gel and water caps in both the lab and home environments is illustrated in Figure 3.2, averaged over channels O1, Oz, and O2. The SNR was not significantly lower for the water cap than the gel cap (median $\pm$ IQR: gel -8.1 $\pm$ 2.8 dB, water -8.9 $\pm$ 1.1 dB, p=0.062,r=0.41). The water cap at home showed no statistically significant difference in SNR to the water cap in the lab (median $\pm$ IQR: -9.2 $\pm$ 4.1 dB, p=0.89). The SNR of each participant separately can be found in Appendix A, Figure A.2. Within Figure 3.2b, the scatterplot also shows the influence of the impedance values on the SNR of the signal. Each scatterpoint indicates one participant and is colored according to the averaged occipital channel impedance value. The lighter the color, the lower and therefore 'better' the impedance. Figure 3.2a shows the distribution of the average SNR over all 32 channels. For the gel and water cap, the SNR is reported highest near the occipital channels and the distribution appears fairly left-right symmetrical. This also seems to be the case for the water cap at home.

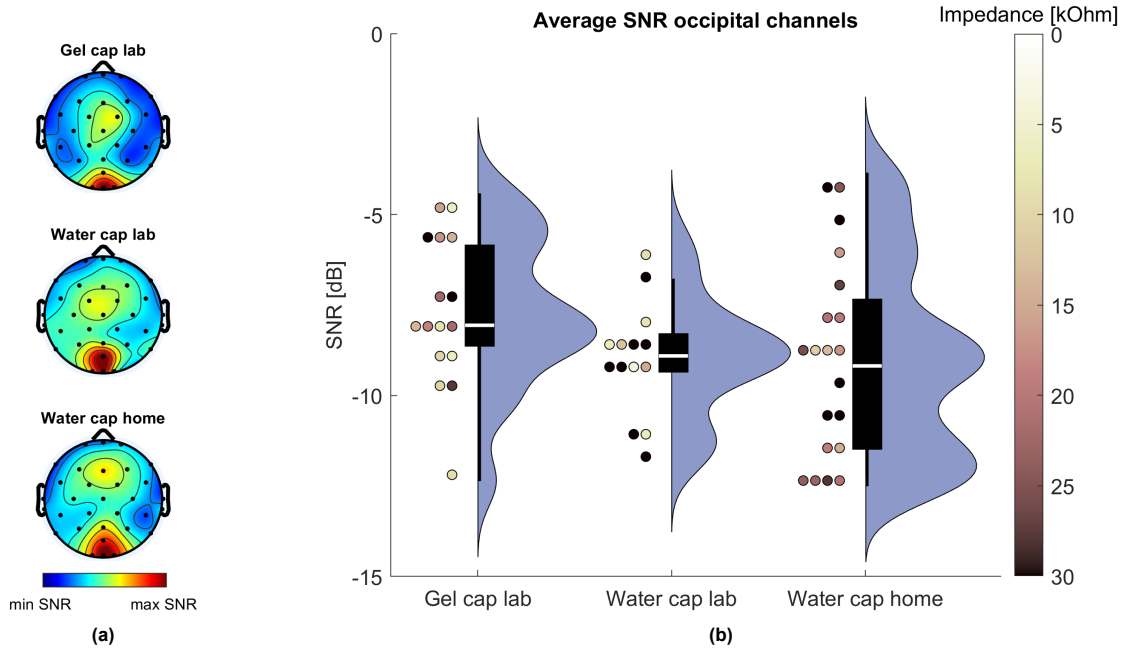

Figure 3.2: Signal-to-noise ratio (SNR) and impedance. The SNR is averaged over all three occipital channels. (a) distribution of SNR over all channels. (b) SNR violin- and boxplot for gel cap, water cap in the lab and water cap at home. The scatterplot shows every participant and is colored according to the averaged occipital channel impedance values.

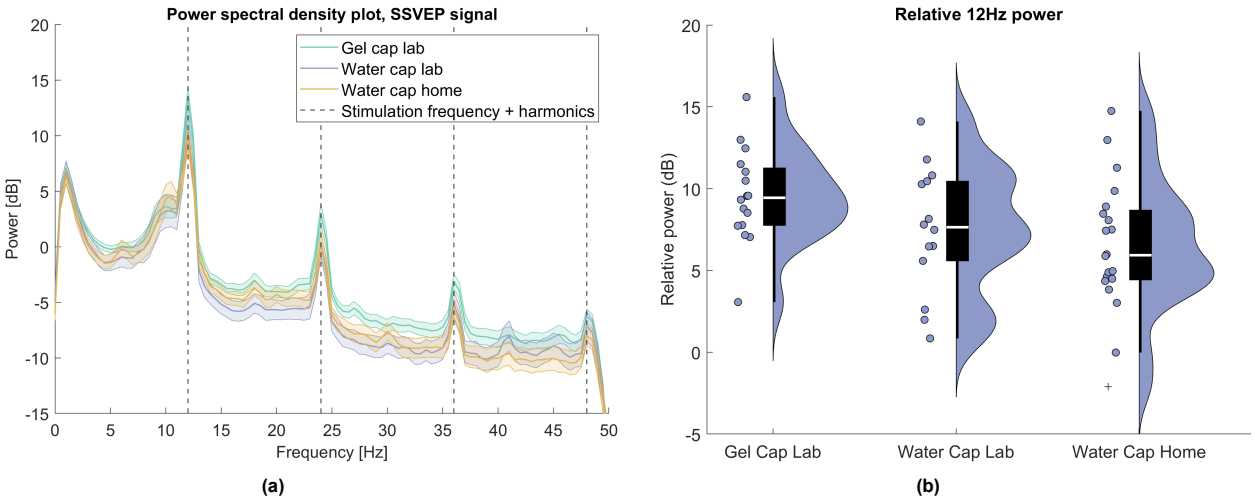

Figure 3.3: Relative 12 Hz power, steady-state visual evoked potential (SSVEP) data. (a) Power spectral density (PSD): every line represents the average PSD of all participants and the shaded area the standard error. (b) Violin plot: the white line shows the median, the edges of the box the inter-quartile range (IQR) and the whiskers show the highest and lowest samples.

3.2.3. Relative 12 Hz power

Figure 3.3a shows the SSVEP power spectral density (PSD) plots for the gel and water cap, where peaks around 12, 24, 36 and 48 Hz can be seen, as well as a response in the 1-2 Hz range. The average 12 Hz power was 19.6, 8.2 and 15.1 dB for the gel cap, water cap in the lab and water cap at home, respectively. Figure 3.3b shows the participant distribution of relative 12 Hz power for both caps and conditions. The gel cap reached the highest median relative power, which was significantly higher than the water cap in the lab (median $\pm$ IQR: gel cap, 9.3 $\pm$ 3.6 dB; water cap lab, 7.3 $\pm$ 4.9 dB; p=0.02, r=0.56). There was no difference in relative 12 Hz power between the lab and home environments (median $\pm$ IQR: 5.5 $\pm$ 3.9 dB; p=0.47).

3.2.4. Relative alpha band power

Contrary to the PSD of the SSVEP data, the RS data shows higher power across the 10 Hz range for the water cap than for the gel cap, as can be seen in Figure 3.4a. Also in all plots, a small response around 1 Hz is visible. Additionally, somewhat visible in every plot is a very small peak at 41 Hz. The gel cap also seems to display some responses in eyes opened and eyes closed data around 17, 27 and 36 Hz. Average alpha power was 23.5, 38.1 and 41.3 dB for the gel cap, water cap in the lab and

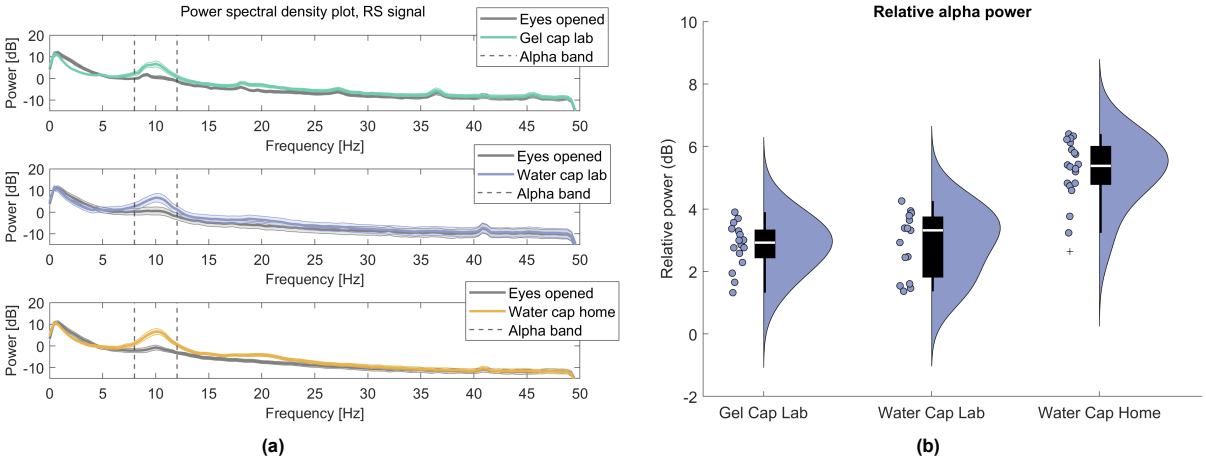

Figure 3.4: Relative alpha power, RS data. (a) Power spectral density (PSD): every line represents the average PSD of all participants and the shaded area the standard error. (b) Violin plot: the white line shows the median, the edges of the box the inter-quartile range (IQR) and the whiskers show the highest and lowest samples.

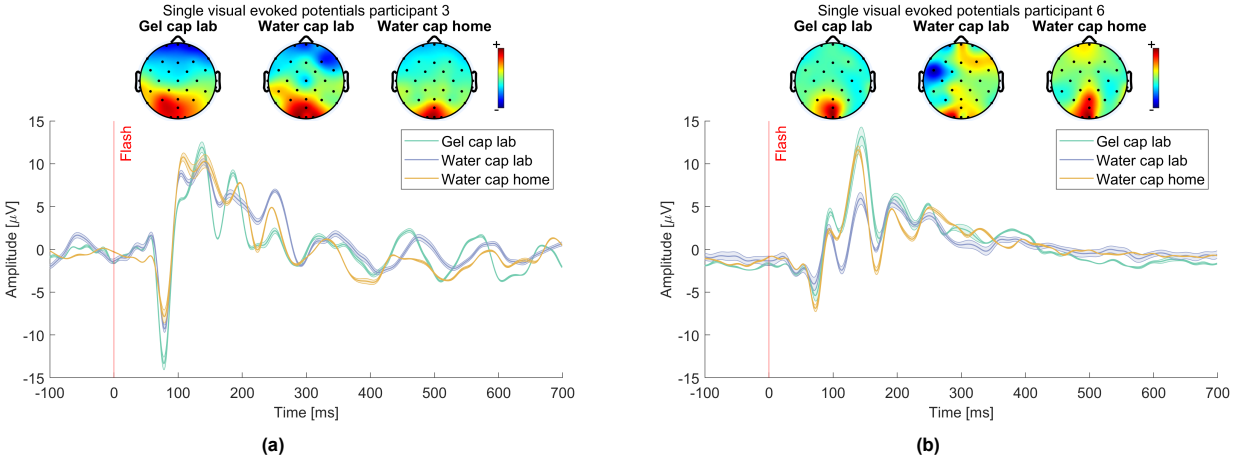

Figure 3.5: SVEP response of two participants, averaged over occipital channels. The shaded area represents the standard error. The amplitude of the signal at the latency of the first positive peak is shown above. (a) participant 3; (b) participant 6.

water cap at home, respectively. Figure 3.4b shows the participant distribution of relative alpha power for both caps. There was almost no difference in relative alpha power between the gel and water cap in the lab (median $\pm$ IQR: gel cap, 2.9 $\pm$ 0.9 dB; water cap lab, 3.3 $\pm$ 1.9 dB; p=0.43, r=0.05). The water cap did reach a significantly higher relative alpha power for the at home measurements than the lab measurements (median $\pm$ IQR: 5.4 $\pm$ 1.2 dB; p=0.00).

3.2.5. Alpha band presence

15 participants (88%) showed a significant alpha power difference between eyes opened and eyes closed conditions when wearing the gel cap, whereas this number of participants was 12 (86%) for the water cap in the lab. All participants that did not show a significant Berger effect with the gel cap also did not show a significant difference for the water cap in the lab. For the at home measurements, 19 participants showed a significant alpha power rise when their eyes were closed (95%).

3.2.6. SVEP presence

Nine participants completed the lab study as well as the home study, making SVEP signal comparisons between gel and water cap and between lab and home setting possible. The averaged SVEP response across the three occipital channels (O1, Oz, O2) for two participants is shown in Figure 3.5. Between caps nearly no difference can be seen, apart from a slightly higher amplitude for the gel cap signal. For both participants, a negative peak is visible just before t=100 ms, as well as a positive peak around t=150 ms. After t=300 ms, the signal doesn't reach a stationary baseline but instead seems to display a wave-like form.

	Lab study		**Home study**	
Quality metric	Gel cap lab	Water cap lab	Water cap lab	Water cap home
Artefacts: SSVEP	0 $\pm$ 0%	10 $\pm$ 37.5%	10 $\pm$ 37.5%	0 $\pm$ 5%
Artefacts: RS	7.14 $\pm$ 11.6%	7.14 $\pm$ 33%	7.14 $\pm$ 33%	10.7 $\pm$ 16%
Artefacts: SVEP	0 $\pm$ 0.92%	4.33 $\pm$ 8.75%	4.33 $\pm$ 8.75%	4 $\pm$ 9.2%
SNR	-8.1 $\pm$ 2.8 dB	-8.9 $\pm$ 1.1 dB	-8.9 $\pm$ 1.1 dB	-9.2 $\pm$ 4.1 dB
Relative 12Hz power	9.3 $\pm$ 3.6 dB	7.3 $\pm$ 4.9 dB	7.3 $\pm$ 4.9 dB	5.5 $\pm$ 3.9 dB
Relative alpha power	2.9 $\pm$ 0.9 dB	3.3 $\pm$ 1.9 dB	3.3 $\pm$ 1.9 dB	5.4 $\pm$ 1.2 dB
Alpha band	15 (88%)	12 (86%)	12 (86%)	19 (95%)

Table 3.2: Overview of results. Metrics are reported as the median $\pm$ IQR. A result is colored if there has been found a significant difference between caps or locations. Red for a lower signal quality, green for a higher signal quality.

4

Discussion and Conclusion

The aim of this thesis is to determine the signal quality of EEG signals measured with a water electrode EEG cap, compared to the commonly used gel electrode EEG cap. The secondary aim of this thesis is to determine the signal quality of this water cap when measured at home, compared to a conventional lab setting.

The water and gel cap showed similar signal quality in the lab, as illustrated by a similar SNR, relative alpha power, alpha band presence and SVEP waveform. However, an increase in artefacts and slight decrease in relative 12 Hz power and SVEP amplitude show remaining shortcomings of the water cap compared to the current golden standard. Provided that these water cap limitations can be mitigated by further developments, the otherwise relatively comparable signal quality between the gel and water caps suggests that water-based EEG systems could be a viable alternative to traditional gel-based systems.

The signal quality of the water cap in the home setting closely matched the signal quality obtained in the lab. This is demonstrated by the similar SNR, relative 12 Hz power, alpha presence and SVEP waveform and amplitude. Surprisingly, the water cap even performed better at home than in the lab for some occasions, as indicated by the decrease in artefacts and increase in relative alpha band power. This comparable signal quality between lab and home environments suggest that home measurements could be a reliable alternative to traditional clinical settings. This is particularly relevant for conditions such as epilepsy and migraine, where continuous monitoring could provide valuable insights into research findings, disease patterns and treatment efficacy without the need for frequent hospital visits.

4.1. Interpretation of results

4.1.1. Artefact proportion

The higher proportion of artefacts observed in the SSVEP water cap data in the lab setting may be attributed to the anomalous signals recorded in three participants, for whom all epochs were excluded. Thus for three participants, an artefact proportion of 100% was noted. Although the precise source of these artefacts remains unidentified, they may have been caused by issues with the amplifier or its connection. Further investigation is warranted to clarify the origin of these artefacts, particularly since they were only observed during the lab experiments, not at home. Given the inability to remove the artefacts or reconstruct the affected signals, the decision was made to exclude this data to prevent it from biasing other quality metrics. The anomalous signals are not likely a direct effect of the water cap properties, therefore including them would skew the data unfairly. Apart from the three inexplicable artefacted participants, a higher number of artifacts in the water cap data might also be explained by the use of a saline solution as conductive material instead of electrolytic gel. This technique leads to a higher electrode-skin impedance and thus water caps are suspected to be more prone to artefacts [49]. Studies also assessing data manually for artefact removal reported similar artefact rates of 11 to 25% for mobile EEG devices [36, 42, 50].

The decreased artefact proportion in the home measurements compared to the lab setting might also be explained by the three participants with anomalous signals, which only occurred during the lab study. Interestingly, for the SSVEP data, these findings differ from the hypothesis that for home measurements, the frequency of artefacts tends to increase [14, 27, 28]. The artefact susceptibility might be caused mainly by the water electrodes, since a gel cap study found the artefact proportion to decrease when measuring at home (11% in the lab, 7% at home) [32].

4.1.2. SNR

One of the main findings of this study is the SNR, which does not differ significantly between caps or environments.

The SNR remains a difficult concept to define on EEG signals, since its definition requires a pure 'signal' and pure noise source that are only possible to obtain from phantom subjects where the input signal is known and not from human participants. Therefore, the true brain signal and noisy attribution must be approximated and this is done in different ways in the field. Some studies defined the preprocessed signal as the true signal after using techniques such as filtering and the use of algorithms like artifact subspace reconstruction (ASR) or independent component analysis (ICA) [37–39, 42]. However, because most studies use slightly different preprocessing steps, the results are not comparable between them. Thus, it is not surprising that SNR values reported in these studies span a broad range from -50 to 16 dB.

The SNR definition that was used in this study was also employed by Vlaar et al. Average SNR values in that study were reported to be a little lower, nonetheless in the same order of magnitude (-11 to -21 dB) [46]. To give insight into this SNR metric, the best and worst performing signals are displayed in Figure A.3. For each cap in the lab, and for the home measurement, the signals of the participants with the 20% highest and lowest SNR scores were averaged. This figure shows that the signals with a higher SNR indeed seem to be of better quality than those with a low SNR.

The distribution of SNR over all channels appears relatively symmetrical in this study and is clearly highest in the occipital channels (O1, Oz, O2). Since the SNR is calculated on EEG data following visual stimulation, this result is to be expected [21] and corresponds with other visual-evoked SNR results in literature [40]. There was no big difference in SNR between occipital channels, as can be seen in Figure A.4.

Electrode impedance is often used as a proxy for the quality of recorded clinical EEG signals. For conventional EEG devices, the standard is often $10k\Omega$ and under, but in practice can range up to $50k\Omega$ [51–53]. A low impedance has often been found to lead to high signal quality [54, 55]. However, looking at the distribution of the impedance levels in Figure 3.2b, there seems to be no correlation between impedance and SNR for this study. This could be because there truly is no correlation, which has sometimes been described in literature as well [47, 53]. The finding could also be explained by the limitations of this study.

4.1.3. Relative 12 Hz power

Over the course of the SSVEP PSD, the gel cap shows a higher power than the water cap for almost every frequency, not only at the stimulating frequency and its harmonics, but also the frequency bins in between. This may be due to the ionic bridging of the electrolytic gel used. This gel might provide a stronger signal to the electrodes, which would lead to higher power across the frequency spectrum. Therefore, this suggests that the gel cap may be more effective at capturing both the desired neural signals and possibly other background signals, leading to a generally higher power spectrum.

Often a definition of EEG SNR has been made in the frequency domain, which might better be called a 'relative power'. The metric is calculated by assigning certain frequency bands to the true signal and noise, and dividing the power over the signal frequencies by the noise frequency power [40, 47, 51]. These frequency bands of interest differ between studies, dependent on study design. For example, an SSVEP study might define the signal frequency band around the stimulation frequency, whereas for RS data this can be around the alpha band. Another variable parameter in this definition is the width of the band of interest and 'noise' frequency band. This metric can be insightful, because in most cases, the higher the level of noise in an EEG channel, the lower the SSVEP response [26]. However, one

argument against this metric is the existence of true brain signal at the frequencies assigned as noise. For example, for RS data, all frequencies outside the alpha band contain not only noise but also theta, beta or gamma waves that might be present during that time. This is why it is not truly a 'signal-to-noise ratio' and it is included here as the relative 12 Hz power.

The gel cap showed a higher relative 12 Hz power than the water cap, which indicates a better performance of the gel cap at establishing the brains response to visual stimulation at 12 Hz. This finding suggests that the gel cap provides a stronger or clearer signal at the stimulation frequency. However, it remains a question whether the lower water cap response means that it is truly insufficient at detecting frequency-specific neural responses.

A similar relative 12 Hz power of 7 dB for gel electrodes was found by Heijs et al [40]. One study that employed similar calculation methods reached a lower relative stimulation power of 0.37-2.85 dB for both dry and gel electrodes [48]. The difference with this study's results might be attributed to their lower SSVEP stimulation frequency of 6 Hz, which could lead to a lower response. Or it could be due to their use of only six electrodes, none of them being located near the Oz electrode, while the response to visual stimulation is expected the highest near the occipital area (O1, Oz, O2 electrodes) [21, 40].

4.1.4. Relative alpha band power

Between gel and water cap, there was no significant difference in relative alpha band power. This result indicates a similar capability at detecting the alpha wave during RS eyes closed measurements. Surprisingly, the home experiment resulted in a higher relative alpha band power than the water cap lab experiment. This increase in performance was not expected, since the home environment was suspected to introduce more noise and artefacts. It should be noted that both water cap measurements but especially the home measurement displayed a great range of samples. A known element of the Berger effect is that there can be a great difference between participants in magnitude of alpha power [56]. Since the home study partly contained different participants than the lab study, perhaps the increase in relative alpha power could be due to the fact that the home participants themselves were more susceptible to the Berger effect.

Average eyes-closed (not relative) alpha power was found to be similar to the values found in this study for gel electrodes in the lab and at home (30-40 dB) [32]. The relative alpha power values found in this study are 40% lower than those found in literature of 5-10 dB [47, 48]. This difference could be attributed to the relatively low amount of participants in these studies (6 and 11), therefore the values might not be representative of the true mean relative alpha band powers of a population. Apart from the mean power values, no range was reported, therefore no accurate estimation of the population could be made.

4.1.5. Alpha band presence

Considering Figure 3.4a, the difference in alpha band power between eyes opened and eyes closed conditions is clearly visible. This effect also portrays in the participant proportion with a significant difference between eyes opened and eyes closed conditions. This signal quality metric could be seen as the bare minimal needed performance from an EEG device. Almost all participants showed this difference. In literature this effect has also been studied. Generally, a two-fold increase in alpha power is to be expected when a subject closes their eyes [32, 57].

4.1.6. SVEP presence

The SVEP waveform seems similar between both caps and both lab and home environment for most participants. The seemingly higher amplitude in the gel cap data suggests an ability of the gel cap to better capture the response of the brain to visual stimuli. An interesting occurrence is the wave in the signal that seems to appear after stimulation, clearly visible in Figure 3.5a around t=300ms. The signal does not seem to return to the baseline signal, that is, the signal before the stimulus. This finding is out of line with SVEPs usually found in literature [37]. One hypothesis of the cause of this wave is the alpha wave that appears when the subject has closed their eyes, which was the case for the duration of this experiment. This fits with the theory that the alpha wave experiences a 'phase-reset' due to the stimulation and therefore is more prominently visible after stimulation than before [58]. A finding supporting this hypothesis is the SVEP waveform of one participant which does not seem to display this

alpha-like wave after stimulation, see Figure 3.5b. When examining the RS data, it is found that this particular participant does not display such a clear alpha peak in the PSD as most other participants, as can be seen in Figure A.5b.

For most participants, the amplitude of the signal at the latency of the first positive peak is highest near the occipital channel and appears relatively symmetrical, which is in line with previous findings [37].

4.2. Limitations

The observed differences in number of artifacts may be partially due to the three participants with unusually large artifacts in the data, of which all epochs had to be deleted. Whether the cause of these artifacts was linked to the properties of the water cap, or something else entirely like amplifier or lab equipment, is unclear. Additionally, the process of manual artifact deletion could have contributed to this variability, as it was performed by a non-expert and lacked standardized criteria. Although the guideline of $mean \pm 3 \cdot SD$ was employed for artifact detection, this approach may not be entirely robust. To enhance the reliability and consistency of artifact removal, future studies should consider having the data reviewed by a professional with expertise in EEG signal processing, or employing a clear and robust guideline for artefact detection. Alternatively, advanced techniques such as ASR or ICA could be employed for more automated artifact or noise removal. However, these algorithms may not perform identically on different datasets. This raises the concern that any observed differences in signal quality might reflect variations in algorithmic performance rather than true differences between the EEG caps.

One factor potentially influencing the signal quality and therefore the SNR is the electrode-skin impedance before the EEG measurement. The overall impedance of this study appeared to be higher than the clinical standard in some cases, both for the gel and water cap and especially in the home setting. This might have lowered the signal quality for both caps, but it could also have skewed the data unfairly if the higher impedance was not a result of the water cap properties or home environment directly. Additionally, the impedance was only measured once, before all three experiments, of which the order was randomized. The SNR is calculated on the SSVEP data, which could be measured directly after impedance measurement or at the end of the experiments which could be as much as 15 minutes after the impedance measurement. Therefore the actual impedance at the time of measurement is not exactly known. It is also hypothesized that for both the gel and water cap, impedance improved during the experiments. Therefore, in reality, this value could have been different than noted in this study. However, as this study was not designed to investigate the effect of impedance on signal quality, follow-up research should be done to confirm this hypothesis. Research has already been done to investigate this, but conclusions differ [47, 53–55]. A follow-up study could be done by measuring impedance before every experiment and after all experiments are finished, or by making use of continuous impedance measurements, which are gaining in popularity together with mobile EEG devices [59].

In the field of EEG measurements, research has been done towards the influence of hair types on electrode-skin impedance and EEG signal quality [37]. In some cases, researchers even employ screening criteria for hair characteristics to avoid a negative impact on signal quality [60]. In this study, no such screening criteria were employed and hair type of the participants was not recorded. In the lab study, paired comparisons could be made because every participant underwent the experiment with both caps. The comparison between water cap in the home setting and the water cap in the lab setting, however, might have been impacted by inter-participant differences, like hair type. This effect might be of even more importance for the water cap, because the scalp-electrode bridge made by the saline solution might be hindered by certain hair types. Since for the gel electrodes, the gel was inserted with the help of a blunt-tipped needle, this bridge was more easily formed. Therefore hair type might have had less effect on the results.

4.3. Future recommendations

In this study, LED goggles were used to deliver visual stimuli. While the portability of the goggles and the LETTI device facilitated home measurements, participants were required to close their eyes during the experiments. This likely contributed to the appearance of alpha waves, which may have influenced the SVEP and possibly the SSVEP. Future studies might consider reducing the brightness of the flashes to allow participants to keep their eyes open or exploring alternative visual stimulation methods, such

as a TV screen or a checkerboard pattern.

This study used several performance metrics to assess signal quality, each highlighting different aspects of the EEG signal. Future research could focus on developing an optimal method for integrating these metrics, potentially by assigning different weights to each metric. These weights might vary depending on the research focus; for instance, placing greater emphasis on relative 12 Hz power when frequency response is a critical factor, or placing a smaller weight on the relative alpha power when a paired participant comparison is not possible.

The present study evaluated signal quality exclusively in healthy participants. It would be valuable to extend this research to populations with specific neurological conditions, such as migraine or epilepsy, to determine if signal quality metrics differ in these groups compared to healthy individuals. This would also provide crucial insights into whether the water cap's performance relative to the gel cap varies depending on the population being studied.

For future research involving home-based EEG measurements, it may also be beneficial to incorporate additional factors that could facilitate the transition to home environments. These factors could include assessing the user-friendliness of the devices and the feasibility of patients conducting the experiments independently. This can lead to further steps towards quantified signal quality of home EEG measurements, to assess and improve the possibility of home measurements for better epilepsy treatment efficacy and migraine attack characterisation.

References

[1] G. Buzsáki, C. A. Anastassiou, and C. Koch, "The origin of extracellular fields and currents — EEG, ECoG, LFP and spikes," *Nature Reviews Neuroscience*, vol. 13, pp. 407–420, June 2012.

[2] L. Zhao, Y. Zhang, X. Yu, H. Wu, L. Wang, F. Li, M. Duan, Y. Lai, T. Liu, L. Dong, and D. Yao, "Quantitative signal quality assessment for large-scale continuous scalp electroencephalography from a big data perspective," *Physiological Measurement*, vol. 44, p. 035009, Mar. 2023. Publisher: IOP Publishing.

[3] M. X. Cohen, "Where Does EEG Come From and What Does It Mean?," *Trends in Neurosciences*, vol. 40, pp. 208–218, Apr. 2017.

[4] M. J. Casale, L. V. Marcuse, J. J. Young, N. Jette, F. E. Panov, H. A. Bender, A. E. Saad, R. S. Ghotra, S. Ghatan, A. Singh, J. Y. Yoo, and M. C. Fields, "The Sensitivity of Scalp EEG at Detecting Seizures—A Simultaneous Scalp and Stereo EEG Study," *Journal of clinical neurophysiology : official publication of the American Electroencephalographic Society*, vol. 39, pp. 78–84, Jan. 2022.

[5] S. J. M. Smith, "EEG in the diagnosis, classification, and management of patients with epilepsy," *Journal of Neurology, Neurosurgery & Psychiatry*, vol. 76, pp. ii2–ii7, June 2005. Publisher: BMJ Publishing Group Ltd.

[6] "EEG (Electroencephalography) | Barnes-Jewish Hospital. Available at: https://www.barnesjewish.org/Medical-Services/Neurology-Neurosurgery/Epilepsy/Diagnosing-Epilepsy/EEG-Electroencephalography. Accessed september 11.."

[7] M. A. Rogawski, "Migraine and Epilepsy—Shared Mechanisms within the Family of Episodic Disorders," in *Jasper's Basic Mechanisms of the Epilepsies* (J. L. Noebels, M. Avoli, M. A. Rogawski, R. W. Olsen, and A. V. Delgado-Escueta, eds.), Bethesda (MD): National Center for Biotechnology Information (US), 4th ed., 2012.

[8] T. C. van den Hoek, M. van de Ruit, G. M. Terwindt, and E. A. Tolner, "EEG Changes in Migraine—Can EEG Help to Monitor Attack Susceptibility?," *Brain Sciences*, vol. 14, p. 508, May 2024. Number: 5 Publisher: Multidisciplinary Digital Publishing Institute.

[9] M. Ashina, G. M. Terwindt, M. A.-M. Al-Karagholi, I. d. Boer, M. J. Lee, D. L. Hay, L. H. Schulte, N. Hadjikhani, A. J. Sinclair, H. Ashina, T. J. Schwedt, and P. J. Goadsby, "Migraine: disease characterisation, biomarkers, and precision medicine," *The Lancet*, vol. 397, pp. 1496–1504, Apr. 2021. Publisher: Elsevier.

[10] J. G. Webster and A. J. Nimunkar, *Medical Instrumentation: Application and Design*. John Wiley & Sons, June 2020. Google-Books-ID: 1ovgDwAAQBAJ.

[11] V. Mihajlović, B. Grundlehner, R. Vullers, and J. Penders, "Wearable, Wireless EEG Solutions in Daily Life Applications: What are we Missing?," *IEEE Journal of Biomedical and Health Informatics*, vol. 19, pp. 6–21, Jan. 2015. Conference Name: IEEE Journal of Biomedical and Health Informatics.

[12] S. R. Benbadis, "What type of EEG (or EEG-video) does your patient need?," *Expert Review of Neurotherapeutics*, vol. 15, pp. 461–464, May 2015. Publisher: Taylor & Francis _eprint: https://doi.org/10.1586/14737175.2015.1029918.

[13] A. Biondi, P. Laiou, E. Bruno, P. F. Viana, M. Schreuder, W. Hart, E. Nurse, D. K. Pal, and M. P. Richardson, "Remote and Long-Term Self-Monitoring of Electroencephalographic and Noninvasive Measurable Variables at Home in Patients With Epilepsy (EEG@HOME): Protocol for an Observational Study," *JMIR Research Protocols*, vol. 10, p. e25309, Mar. 2021.

[14] J. Askamp and M. J. A. M. van Putten, "Mobile EEG in epilepsy," *International Journal of Psychophysiology*, vol. 91, pp. 30–35, Jan. 2014.

[15] A. Biondi, V. Santoro, P. F. Viana, P. Laiou, D. K. Pal, E. Bruno, and M. P. Richardson, "Noninvasive mobile EEG as a tool for seizure monitoring and management: A systematic review," *Epilepsia*, vol. 63, pp. 1041–1063, May 2022.

[16] G. Niso, E. Romero, J. T. Moreau, A. Araujo, and L. R. Krol, "Wireless EEG: A survey of systems and studies," *NeuroImage*, vol. 269, p. 119774, Apr. 2023.

[17] B. A. Taheri, R. T. Knight, and R. L. Smith, "A dry electrode for EEG recording," *Electroencephalography and Clinical Neurophysiology*, vol. 90, pp. 376–383, May 1994.

[18] G. Gargiulo, P. Bifulco, R. A. Calvo, M. Cesarelli, C. Jin, and A. van Schaik, "A mobile EEG system with dry electrodes," in *2008 IEEE Biomedical Circuits and Systems Conference*, pp. 273–276, Nov. 2008. ISSN: 2163-4025.

[19] A. J. Casson, "Wearable EEG and beyond," *Biomedical Engineering Letters*, vol. 9, pp. 53–71, Feb. 2019.

[20] E. Huigen, A. Peper, and C. A. Grimbergen, "Investigation into the origin of the noise of surface electrodes," *Medical and Biological Engineering and Computing*, vol. 40, pp. 332–338, May 2002.

[21] V. Mihajlović, G. Garcia-Molina, and J. Peuscher, "Dry and Water-Based EEG Electrodes in SSVEP-Based BCI Applications," in *Biomedical Engineering Systems and Technologies* (J. Gabriel, J. Schier, S. Van Huffel, E. Conchon, C. Correia, A. Fred, and H. Gamboa, eds.), (Berlin, Heidelberg), pp. 23–40, Springer, 2013.

[22] K. E. Mathewson, T. J. L. Harrison, and S. A. D. Kizuk, "High and dry? Comparing active dry EEG electrodes to active and passive wet electrodes," *Psychophysiology*, vol. 54, no. 1, pp. 74–82, 2017. _eprint: https://onlinelibrary.wiley.com/doi/pdf/10.1111/psyp.12536.

[23] E. Kutafina, A. Brenner, Y. Titgemeyer, R. Surges, and S. Jonas, "Comparison of mobile and clinical EEG sensors through resting state simultaneous data collection," *PeerJ*, vol. 8, p. e8969, May 2020. Publisher: PeerJ Inc.

[24] T. Neumann, A. K. Baum, U. Baum, R. Deike, H. Feistner, M. Scholz, H. Hinrichs, and B. Robra, "Assessment of the technical usability and efficacy of a new portable dry-electrode EEG recorder: First results of the HOMEONE study," *Clinical Neurophysiology*, vol. 130, pp. 2076–2087, Nov. 2019.

[25] L.-D. Liao, I.-J. Wang, S.-F. Chen, J.-Y. Chang, and C.-T. Lin, "Design, Fabrication and Experimental Validation of a Novel Dry-Contact Sensor for Measuring Electroencephalography Signals without Skin Preparation," *Sensors*, vol. 11, pp. 5819–5834, June 2011. Number: 6 Publisher: Molecular Diversity Preservation International.

[26] V. Mihajlović, G. Garcia-Molina, and J. Peuscher, "To what extent can dry and water-based EEG electrodes replace conductive gel ones?: A Steady State Visual Evoked Potential Brain-Computer Interface Case Study," *BIODEVICES 2012 - Proceedings of the International Conference on Biomedical Electronics and Devices*, Jan. 2012.

[27] K. T. Sweeney, T. E. Ward, and S. F. McLoone, "Artifact Removal in Physiological Signals—Practices and Possibilities," *IEEE Transactions on Information Technology in Biomedicine*, vol. 16, pp. 488–500, May 2012. Conference Name: IEEE Transactions on Information Technology in Biomedicine.

[28] A. S. Oliveira, B. R. Schlink, W. D. Hairston, P. König, and D. P. Ferris, "Proposing Metrics for Benchmarking Novel EEG Technologies Towards Real-World Measurements," *Frontiers in Human Neuroscience*, vol. 10, May 2016. Publisher: Frontiers.

[29] I. P. Martins, M. Westerfield, M. Lopes, C. Maruta, and R. Gil-da Costa, "Brain state monitoring for the future prediction of migraine attacks," *Cephalalgia*, vol. 40, pp. 255–265, Mar. 2020. Publisher: SAGE Publications Ltd STM.

[30] U. Baum, A.-K. Baum, R. Deike, H. Feistner, B. Markgraf, H. Hinrichs, B.-P. Robra, and T. Neumann, “Feasibility assessment of patient-controlled EEG home-monitoring: More results from the HOMEONE study,” *Clinical Neurophysiology*, vol. 140, pp. 12–20, Aug. 2022.

[31] S. Troller-Renfree, S. Morales, S. Leach, M. Bowers, R. Debnath, W. Fifer, N. Fox, and K. Noble, “Feasibility of assessing brain activity using mobile, in-home collection of electroencephalography: methods and analysis,” *Developmental Psychobiology*, vol. 63, no. 6, 2021.

[32] K. B. Mikkelsen, Y. R. Tabar, C. B. Christensen, and P. Kidmose, “EEGs Vary Less Between Lab and Home Locations Than They Do Between People,” *Frontiers in Computational Neuroscience*, vol. 15, Feb. 2021. Publisher: Frontiers.

[33] F. M. Barbey, F. R. Farina, A. R. Buick, L. Danyeli, J. F. Dyer, M. N. Islam, M. Krylova, B. Murphy, H. Nolan, L. M. Rueda-Delgado, M. Walter, and R. Whelan, “Neuroscience from the comfort of your home: Repeated, self-administered wireless dry EEG measures brain function with high fidelity,” *Frontiers in Digital Health*, vol. 4, p. 944753, 2022.

[34] R. J. Sugden, V.-L. L. Pham-Kim-Nghiem-Phu, I. Campbell, A. Leon, and P. Diamandis, “Remote collection of electrophysiological data with brain wearables: opportunities and challenges,” *Bioelectronic Medicine*, vol. 9, p. 12, June 2023.

[35] A. Delorme, D. Truong, R. Martinez-Cancino, C. Pernet, S. Sivagnanam, K. Yoshimoto, R. Poldrack, A. Majumdar, and S. Makeig, “Tools for importing and evaluating BIDS-EEG formatted data,” *International IEEE/EMBS Conference on Neural Engineering, NER*, vol. 2021-May, pp. 210–213, 2021.

[36] A. Giannadou, M. Jones, M. Freeth, A. Samson, and E. Milne, “Investigating neural dynamics in autism spectrum conditions outside of the laboratory using mobile electroencephalography,” *Psychophysiology*, vol. 59, no. 4, 2022.

[37] T. Lees, N. Ram, M. Swingler, and L. Gatzke-Kopp, “The effect of hair type and texture on electroencephalography and event-related potential data quality,” *Psychophysiology*, 2023.

[38] T. Shivaraja, R. Remli, N. Kamal, W. Wan Zaidi, and K. Chellappan, “Assessment of a 16-Channel Ambulatory Dry Electrode EEG for Remote Monitoring,” *Sensors*, vol. 23, no. 7, 2023.

[39] Z. Shi, B. Jiang, S. Liang, J. Zhang, D. Suo, J. Wu, D. Chen, G. Pei, and T. Yan, “Claw-shaped flexible and low-impedance conductive polymer electrodes for EEG recordings: Anemone dry electrode,” *Science China Technological Sciences*, vol. 66, no. 1, pp. 255–266, 2023.

[40] J. Heijs, R. Havelaar, P. Fiedler, R. van Wezel, and T. Heida, “Validation of soft multipin dry eeg electrodes,” *Sensors*, vol. 21, no. 20, 2021.

[41] D. Liu, Q. Wang, Y. Zhang, X. Liu, J. Lu, and J. Sun, “A study on quality assessment of the surface EEG signal based on fuzzy comprehensive evaluation method,” *Computer Assisted Surgery*, vol. 24, no. sup1, pp. 167–173, 2019.

[42] T. Radüntz, “Signal quality evaluation of emerging EEG devices,” *Frontiers in Physiology*, vol. 9, no. FEB, 2018.

[43] R. Shriram, M. Sundhararajan, and N. Daimiwal, “EEG Based Cognitive Workload Assessment for Maximum Efficiency,” *IOSR Journal of Electronics and Communication Engineering (IOSR-JECE)*, Aug. 2012.

[44] M. Hraar, “RANDOM.ORG - Sequence Generator,” 1998.

[45] A. Delorme and S. Makeig, “EEGLAB: an open source toolbox for analysis of single-trial EEG dynamics including independent component analysis,” *Journal of Neuroscience Methods*, vol. 134, pp. 9–21, Mar. 2004.

[46] M. P. Vlaar, T. Solis-Escalante, J. P. A. Dewald, E. E. H. van Wegen, A. C. Schouten, G. Kwakkel, F. C. T. van der Helm, J. de Munck, C. Meskers, M. Saes, L. Haring, C. Winters, A. Andringa, D. Hoevenaars, I. de Castro Fernandes, S. Zandvliet, A. Daffertshofer, J. Yao, Y. Yang, M. van de

Ruit, K. Kalogianni, L. Filatova, and on behalf of the 4D-EEG consortium, "Quantification of task-dependent cortical activation evoked by robotic continuous wrist joint manipulation in chronic hemiparetic stroke," *Journal of NeuroEngineering and Rehabilitation*, vol. 14, p. 30, Apr. 2017.

[47] A.-M. Tăutan, V. Mihajlović, Y.-H. Chen, B. Grundlehner, J. Penders, and W. Serdijn, "Signal quality in dry electrode EEG and the relation to skin-electrode contact impedance magnitude," *BIODEVICES 2014 - 7th Int. Conference on Biomedical Electronics and Devices, Proceedings; Part of 7th International Joint Conference on Biomedical Engineering Systems and Technologies, BIOSTEC 2014*, pp. 12–22, 2014.

[48] A.-M. Tăutan, W. Serdijn, V. Mihajlović, B. Grundlehner, and J. Penders, "Framework for evaluating EEG signal quality of dry electrode recordings," *2013 IEEE Biomedical Circuits and Systems Conference, BioCAS 2013*, pp. 186–189, 2013.

[49] Q. Liu, L. Yang, Z. Zhang, H. Yang, Y. Zhang, and J. Wu, "The Feature, Performance, and Prospect of Advanced Electrodes for Electroencephalogram," *Biosensors*, vol. 13, p. 101, Jan. 2023.

[50] J. Kam, S. Griffin, A. Shen, S. Patel, H. Hinrichs, H.-J. Heinze, L. Deouell, and R. Knight, "Systematic comparison between a wireless EEG system with dry electrodes and a wired EEG system with wet electrodes," *NeuroImage*, vol. 184, pp. 119–129, 2019.

[51] R. Zerafa, T. Camilleri, O. Falzon, and K. Camilleri, "A comparison of a broad range of EEG acquisition devices–is there any difference for SSVEP BCIs?," *Brain-Computer Interfaces*, vol. 5, no. 4, pp. 121–131, 2018.

[52] S. R. Sinha, L. R. Sullivan, D. Sabau, D. S. J. Orta, K. E. Dombrowski, J. J. Halford, A. J. Hani, F. W. Drislane, and M. M. Stecker, "American Clinical Neurophysiology Society Guideline 1: Minimum Technical Requirements for Performing Clinical Electroencephalography," *The Neurodiagnostic Journal*, vol. 56, pp. 235–244, Oct. 2016. Publisher: Taylor & Francis _eprint: https://doi.org/10.1080/21646821.2016.1245527.

[53] T. C. Ferree, P. Luu, G. S. Russell, and D. M. Tucker, "Scalp electrode impedance, infection risk, and EEG data quality," *Clinical Neurophysiology*, vol. 112, pp. 536–544, Mar. 2001.

[54] E. S. Kappenman and S. J. Luck, "The Effects of Electrode Impedance on Data Quality and Statistical Significance in ERP Recordings," *Psychophysiology*, vol. 47, pp. 888–904, Sept. 2010.

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

[56] C. J. Rennie, P. A. Robinson, and J. J. Wright, "Unified neurophysical model of EEG spectra and evoked potentials," *Biological Cybernetics*, vol. 86, pp. 457–471, June 2002.

[57] C. E. D. Alloway, R. D. Ogilvie, and C. M. Shapiro, "The Alpha Attenuation Test: Assessing Excessive Daytime Sleepiness in Narcolepsy-Cataplexy," *Sleep*, vol. 20, pp. 258–266, Apr. 1997.

[58] B.-K. Min, N. A. Busch, S. Debener, C. Kranczioch, S. Hanslmayr, A. K. Engel, and C. S. Herrmann, "The best of both worlds: Phase-reset of human EEG alpha activity *and* additive power contribute to ERP generation," *International Journal of Psychophysiology*, vol. 65, pp. 58–68, July 2007.

[59] P. Shen, Y. Liu, W. Xiong, A. He, and M. Zhang, "A Real-Time Impedance Measurement System for EEG Based on Embedded System," *2020 13th International Congress on Image and Signal Processing, BioMedical Engineering and Informatics (CISP-BMEI)*, pp. 681–685, Oct. 2020. Conference Name: 2020 13th International Congress on Image and Signal Processing, BioMedical Engineering and Informatics (CISP-BMEI) ISBN: 9780738105451 Place: Chengdu, China Publisher: IEEE.

[60] E. K. Webb, J. A. Etter, and J. A. Kwasa, "Addressing racial and phenotypic bias in human neuroscience methods," *Nature Neuroscience*, vol. 25, pp. 410–414, Apr. 2022. Publisher: Nature Publishing Group.

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Supplementary results

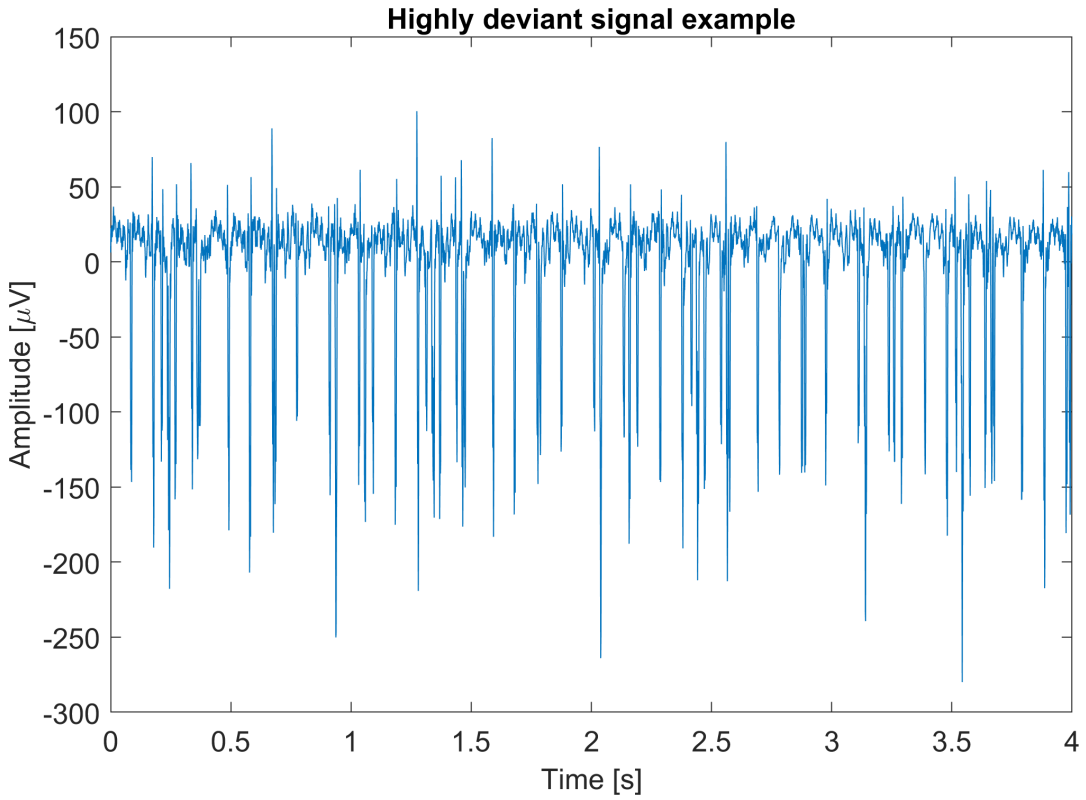

Figure A.1: Deviant artefacted signal for one participant.

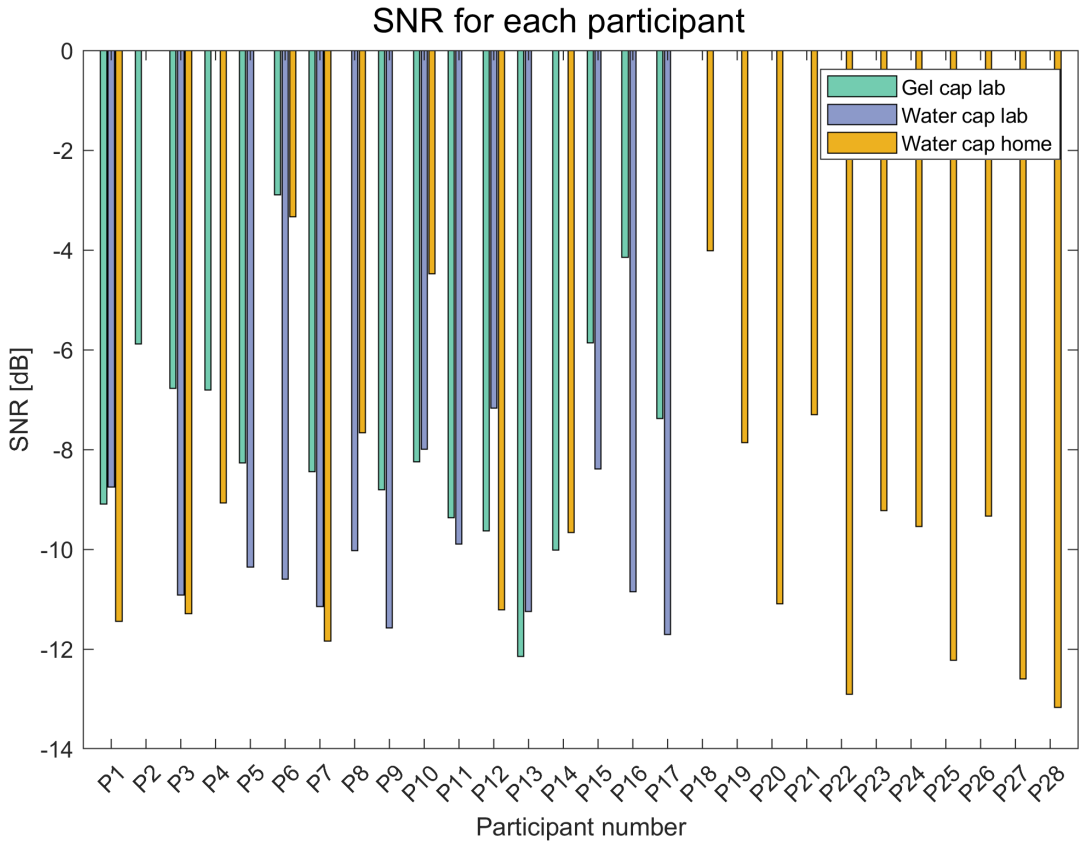

Figure A.2: SNR for every participant.

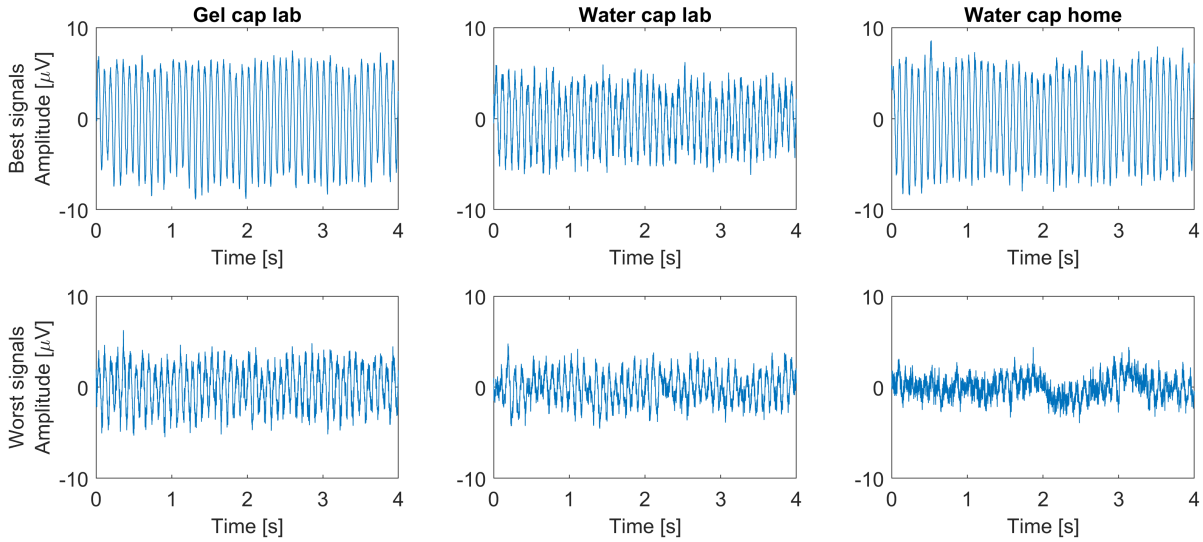

Figure A.3: The four signals with the highest (top row) and lowest (bottom row) SNR, averaged.

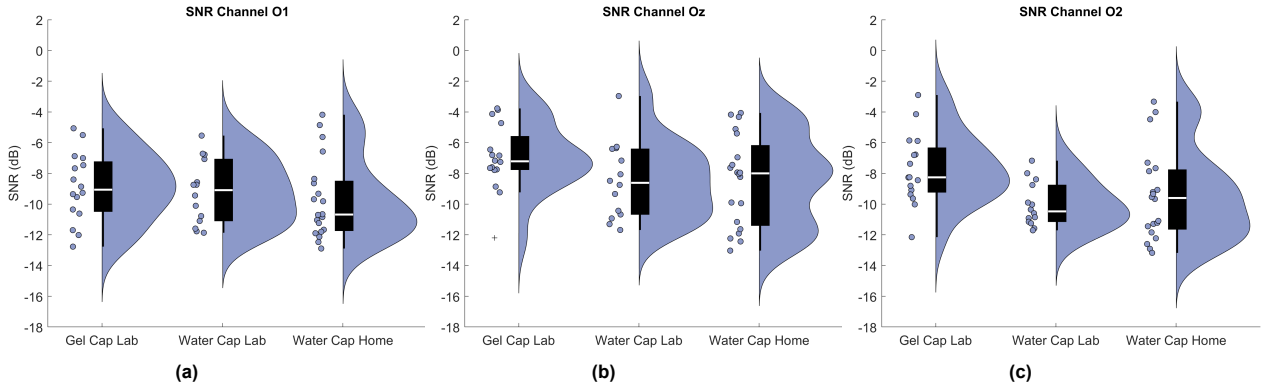

Figure A.4: SNR for all three occipital channels. (a) O1; (b) Oz; (c) O2.

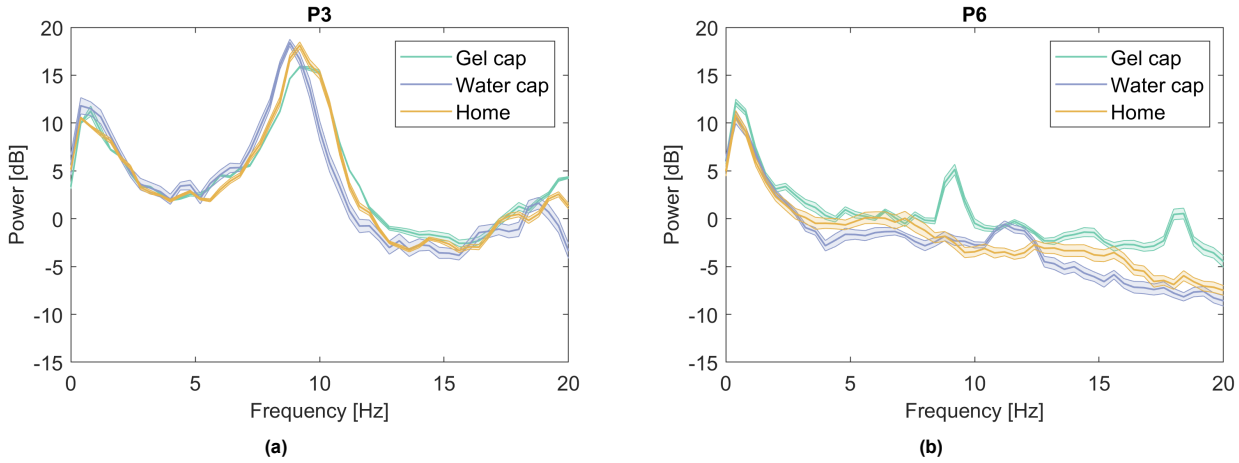

Figure A.5: RS PSD plots for participants corresponding to SVEP plots. (a) participant 3; (b) participant 6.