DEVELOPMENT AND VALIDATION OF DRY ELECTRODES FOR MOBILE EEG MEASUREMENTS

by

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Abstract

Electroencephalography (EEG) is a common neuroimaging technique used in clinical, research, and consumer technology due to its non-invasiveness, high time resolution, and sensitivity. Traditional EEG systems consist of wet electrodes and a desktop amplifier for measuring brain activity. While this system provides high signal quality in lab settings, it limits real-world EEG use, such as for ambulatory EEG monitoring, or EEG measurements during exercise. This is due to the training and set up time required for wet electrode application and the bulky, wired traditional amplifier. While some wearable EEG devices exist on the market, EEG measurements during motion requires further signal quality validation to confirm clinical or research utility due to motion artifacts. The objectives of this thesis are to develop custom dry flexible electrodes with improved usability for wearable systems and to compare signal quality of the custom dry electrodes with traditional wet electrodes in mobile measurements. We have developed comb-shaped, dry electrodes for measuring scalp EEG. This electrode incorporates embedded silver thread for a novel yet simple to fabricate design, which allows for a flexible electrode providing impedances similar or lower to those in literature. Participants were instrumented with an EEG cap containing gold cup, gel, and dry electrodes for brain activity measurement as well as an inertial measurement unit (IMU) mouthguard for head kinematics measurements. Data were collected from 8 participants, and during nearly all trials, dry electrodes showed worse signal correlation with gold cup electrodes compared with gel electrodes. For motion artifact characterization, EEG-IMU coherence was found to be higher for dry electrodes in low motion scenarios (e.g., walking), but similar between the 3 electrode types for high motion scenarios (e.g., jumping). This indicates that dry electrodes may be more prone to motion artifacts at lower levels of activity, but with large enough activity all 3 types of electrodes could be equally
affected. In summary, we developed custom flexible dry electrodes with improved usability for wearable systems. However, we showed that dry electrode signal quality may be substantially affected by motion artifacts, and wet electrodes may be more suitable for measurements during motion.
Lay Summary

Electroencephalography (EEG) is a technique for measuring brain activity, and is commonly used in clinical, research, and consumer technology. The goal of this research is to extend the use of EEG outside of the lab and into the real world, by developing and validating custom flexible dry sensors used for measuring signals from the brain. In this study, participants were instrumented with a wearable EEG system containing two types of wet sensors and novel dry sensors, in addition to a mouthguard that measures head movements. During testing, brain activity recordings for the two types of wet sensors were more similar compared to the dry sensors. During motion, dry sensor recordings had signals more similar to the head movement, indicating that dry sensors are more susceptible to changes in signal during motion, and further work is required to ensure dry sensors provide accurate signal measurements in motion scenarios.
Preface

The work presented in this thesis was supervised by Dr. Lyndia Wu and Dr. Mike Van der Loos, and the main contributors include me and biomedical engineering undergraduate student Han Nguyen.

For the work in Chapter 2, I collaborated with Han to design and fabricate the electrodes, as well as conducted testing to ensure basic functionality of the electrodes. Dr. Saeid Soltanian, the lab manager of the shared facility where electrodes were fabricated, provided some input on our design. Further into electrode development, Kaylee McGeough, another biomedical engineering undergraduate student, helped with electrode fabrication and testing as well. I developed code to analyze the data and results comparing signal quality of the different electrode designs, as well as conducted the electrochemical impedance spectroscopy experiments and data analysis.

For the work in Chapter 3, Han, Kaylee and I ran the experiments validating the wearable EEG system. Han and I iterated through the wearable EEG system prototypes and printing the caps for each participant. Kaylee fabricated new sets of electrodes for the participants. We then ran the experiments with participants together. The human participant experiments were approved by the UBC research ethics board (UBC Clinical Research Ethics Board H20-02313). After all the data were collected, I conducted the EEG data processing and analysis.

I presented the results in Chapter 3 as a poster at the 2021 BMES Annual Meeting, and I am currently preparing manuscript of Chapter 3 for submission to a journal.
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List of Abbreviations

AAC – alpha attenuation coefficient
BCI – brain-computer interface
CAD – computer aided design
CMRR – common-mode rejection ratio
CT – computed tomography
EEG – electroencephalography
EIS – electrochemical impedance spectroscopy
ID – indoor
IMU – inertial measurement unit
MRI – magnetic resonance imaging
OD – outdoor
PLA – polylactic acid
Pw – wait period before each EIS frequency measurement
SNR – signal-to-noise ratio
Te – wait time for current stabilization before EIS measurement
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I would like to thank everyone who has been a part of my journey as I finish up my master’s program. First and foremost, I would like to thank my supervisor Dr. Lyndia Wu, for her consistent help and guidance throughout the ups and downs of my research journey and especially her ability to provide a sense of security and direction even as we navigated the uncertainty and “unprecedented times” brought on by the pandemic. I would also like to thank Dr. Saeid Soltanian for training me in the process of electrode fabrication and validation, and providing suggestions and guidance to help me further develop and improve my electrodes. I would also like to thank my co-supervisor Dr. Mike Van der Loos for helping provide a different viewpoint and perspective on my research, and leading me to think more broadly about my research topic. I would also like to express my gratitude to Professor Peyman Servati and Professor Naznin Virji-Babul for being a part of my supervisory committee during my master’s program.

Thank you to everyone in the SimPL and CARIS lab who helped make my research project a reality, whether that be helping with the experimental setup, or participating in my research as a subject, or both. Special thanks to Han Nguyen, who has been a part of my research journey from day one, and has been by my side every step of the way; I truly could not have accomplished this without you! I would also like to thank Kaylee McGeough, who was a massive help throughout my research journey as well, from initial electrode testing up to the end with human participant testing and data collection.

I also would like to thank UBC and NSERC for the financial support I received for my research project.
Dedication

To my friends and family, who have been a massive support for me throughout my education, as well as my journey through life.
Chapter 1: Introduction

Electroencephalography (EEG) refers to the measurement of electrical brain activity as a voltage, or electric potential difference, typically through the scalp (surface EEG) [1]. To measure this brain activity, the subject typically wears a skull cap with EEG electrodes, which make electrical contact with the subject’s scalp. Traditionally, wet electrodes are used to allow for a good electrical connection, where a conductive gel or paste is applied between the electrode and scalp. The system measures the voltage signal from different areas of the brain, depending on where the electrode is placed. A commonly used system for electrode placement is the 10-20 system, which is shown in Figure 1.1. This voltage signal from the brain to the electrodes on the cap then goes through the wired connection to an amplifier, which amplifies and records the EEG signal. EEG has relatively high temporal resolution when compared with other neuroimaging technologies such as magnetic resonance imaging (MRI) or computed tomography (CT) scans, which have lower temporal resolution but high spatial resolution [2].

![Diagram of 10-20 system](image)

**Figure 1.1 Diagram of 10-20 system**

EEG is a mature neuroimaging modality that has become increasingly applied in clinical, research, and even consumer technology given its high sensitivity, non-invasiveness, and high
Some clinical applications of EEG include diagnosis of epilepsy and sleep disorders [3]. Imbalances in the brain’s electrical activity due to epilepsy can be detected as spikes and sharp waves in the EEG signal [4]; this can be monitored to detect the onset of a seizure. In terms of sleep disorders, EEG can be used to track the different sleep stages, and identify changes or abnormalities in the order of sleep stages [5]. EEG has also been used in the development of brain computer interfaces (BCI), which are systems that can interpret a subject’s brain signal to perform a task and assist patients with physical disabilities [6]. EEG is particularly promising to be developed into wearable form factors for real-world brain sensing applications. Having the ability to record long-term ambulatory EEG data could enable continuous patient monitoring and provide an alternative to the current periodic clinical measurements [7]. Overall, EEG is a safe and non-invasive way to monitor brain activity of both health subjects and patients.

However, technological limitations of current EEG systems are hindering the application of EEG in real-world, continuous mobile measurements. Clinical EEG systems are expensive, require long set up time and they do not allow the subject to move around during measurements due the wiring and hardware system [8]. Most importantly, motion can create artifacts that could introduce substantial noise and mask the EEG signal [9], so these systems cannot provide high quality data in scenarios with motion. This has led to traditional EEG being measured in resting state at the laboratory or clinic, and real-world ambulatory EEG monitoring, which includes more motion, requires further signal quality validation to confirm clinical or research utility. While there are some clinical/research grade mobile EEG systems, most are expensive, costing on the order of $5,000 to $50,000USD [10]. In addition, while these systems are wireless and
reduced in size compared with traditional desktop systems, there is limited evidence of their ability to mitigate motion artifacts during real-world measurements [11].

1.1 Thesis Overview

In this research, our long-term goal is to develop a wearable EEG system that can mitigate motion artifacts and provide high-quality measurements during mobile applications. To contribute to this goal, my thesis research has two main objectives: 1) develop flexible dry electrodes with low electrode-skin impedance to improve usability and signal quality in wearable systems, 2) compare the performance of the custom dry electrodes with wet electrodes in mobile measurements. With these objectives, I focused on dry electrode design, fabrication, and testing. After initial validation of the electrodes, I integrated the electrodes into a wearable EEG system and conducted further testing, where the entire system was tested on human participants in motion scenarios, allowing us to not only test the system, but compare different types of EEG electrodes during motion.

In the following introduction sections, we will provide motivation and background on the current state of wearable EEG systems, EEG electrodes, motion testing and artifacts during EEG measurement, as well as the use of electrochemical impedance spectroscopy (EIS) for electrode characterization. In Chapter 2, we will look at design requirements for dry electrode development, the process as we iterated through different designs and materials, and the final design and validation testing. In Chapter 3, we will compare gold cup, gel, and dry electrodes using a wearable, lab-developed EEG system during mild, moderate, and high motion scenarios. We will also correlate motion data measured using Inertial Measurement Unit (IMU) sensors to the EEG electrode data to measure the level of noise in the EEG data due to motion artifacts.
Chapter 4 summarizes the findings from the chapters, providing the conclusion of this thesis and future directions for continuing this work.

1.2 Motivation for Mobile EEG Measurements

With some of the limitations of traditional EEG systems highlighted previously, there has been a push to develop mobile/wearable EEG systems. The use of mobile EEG has great potential in improving current clinical EEG measurements for monitoring of brain diseases such as epilepsy as well as sleep disorders. The traditional setup for epilepsy monitoring requires patient hospitalization for one or multiple days, where multiple EEG sessions are recorded to increase the chance of epilepsy diagnosis; this is costly and inconvenient for the patient [12]. The development of a wearable EEG device allows for ambulatory recordings at home which improve the convenience of these measurements. It also provides more insight into the patient’s condition, by allowing for long, continuous recordings in an environment where they may be exposed to epilepsy provoking factors [12]. In diagnosing sleep disorders, traditionally polysomnography is used, which monitors parameters such as EEG, heart rhythms, eye and muscle movements, and breathing [13]. Sleep tests are typically performed in a clinical setting, which is expensive and not ideal for long-term monitoring [14]. In addition, the instrumentation and foreign environment can hinder the reliability of the sleep patterns measured [14]. Wearable EEG could be an affordable method for providing long-term and reliable sleep monitoring [14]. In addition to the measurement of EEG in lower motion scenarios, there is potential in using wearable EEG during exercise and sports. EEG can be used to identify neural markers for improving sports performance, which is of great interest to elite athletes [15]. Many studies in sports EEG look at measurements before and after physical activity [16], with physical exercises that allow the participants to remain in place such as biking [17], or exercises that do not require
vigorous motion, such as golfing [18]. In addition to this, in high impact sports, EEG can be used to look at changes in brain activity due to head impacts [19]. Mobile EEG could allow for brain activity measurements outside of the lab [15] and even during physical activity, which could provide more information to improve sports performance as well as monitor brain health in sports where athletes are susceptible to head impacts [19]. While these are some key applications of wearable EEG, many more exist, including for emotion recognition, stress, and meditation [20].

1.3 State of EEG and Wearable EEG systems

Traditional clinical and research grade EEG systems have limitations that prevent the use of these systems outside of lab settings. In terms of the EEG system components, the cap and EEG electrodes are inherently portable; wearability issues arise due to the bulky amplifier and data acquisition system and long wired connections. Both factors restrain the movement of the subject, and lead to periodic in-clinic measurements as opposed to continuous real-world measurements [21]. In terms of the amplifier and data acquisition system, large, bulkier plug-in amplifiers can provide higher channel count, sampling rate, and common-mode rejection ratio (CMRR), which lead to higher spatial resolution, temporal resolution, and better EEG signal quality compared to wearable electronics during stationary measurements [22]. CMRR is a commonly used metric to measure the performance of an amplifier to amplify the signals of interest, while rejecting signals that we do not want in the final output. However, the long, wired connections of desktop systems not only restrain subject movement, but can also lead to noisier data through the injection of current due to the movement of wires through ambient environment electromagnetic fields, which in turn lead to changes in voltage that do not reflect the brain activity [21].
Table 1.1: Traditional Desktop EEG Amplifiers

<table>
<thead>
<tr>
<th>Product</th>
<th>Channels</th>
<th>Sampling Rate (Hz)</th>
<th>CMRR (dB)</th>
<th>References</th>
</tr>
</thead>
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<tr>
<td>TMSi Refa8</td>
<td>40</td>
<td>10000</td>
<td>&gt; 90</td>
<td>[23]</td>
</tr>
<tr>
<td>Geodesic EEG System 400</td>
<td>32, 64, 128, or 256</td>
<td>1000</td>
<td>≥ 90</td>
<td>[24]</td>
</tr>
<tr>
<td>SynAmps RT</td>
<td>64</td>
<td>20000</td>
<td>&gt;110</td>
<td>[25]</td>
</tr>
</tbody>
</table>

In the current market there are mobile research grade systems that allow for robust EEG measurement during some motion scenarios, providing similar signal quality to desktop systems; in comparing Table 1.1 and Table 1.2, most of the mobile research grade systems have a similar CMRR to the traditional desktop amplifiers. However, there is the tradeoff of lower sampling rates, lower channel counts, and the constraint of battery life (Table 1.2). Another factor is the cost; high quality research grade EEG systems can have similar costs to traditional EEG amplifiers. For example, the 32-channel Geodesic EEG System 400 is around 30000USD [24], which is similar to the cost of many systems in Table 1.2.
Table 1.2: Higher Quality Research-grade EEG Systems

<table>
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<tr>
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<th>Battery Life (h)</th>
<th>CMRR (dB)</th>
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<td>100 [26]</td>
<td>[27]</td>
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<tr>
<td>ANT Neuro eego rt sports</td>
<td>64</td>
<td>2048</td>
<td>&gt;25000</td>
<td>6</td>
<td>&gt;100 [28]</td>
<td>[27]</td>
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<td>24995</td>
<td>8</td>
<td>-115 [30]</td>
<td>[27]</td>
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<tr>
<td>LiveAmp 32</td>
<td>32</td>
<td>500, up to 1000</td>
<td>&lt;25000</td>
<td>6</td>
<td>&gt; 80 [31]</td>
<td>Company quote</td>
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<td>ABM B-Alert X24</td>
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<td>256</td>
<td>19950</td>
<td>8</td>
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</tbody>
</table>

There are other relatively low-cost wearable commercial systems available (Table 1.3) which overcome the mobility issues mentioned with the traditional amplifiers and the higher cost issues of the research and clinical grade mobile systems. Through the development of lightweight, wearable electronics that are attached near the head, these low-cost systems provide opportunities for measurement in motion scenarios; since the amplifier is near the head, this
helps reduce the length of wires and therefore the noise induced by the wires as well [21]. These wearable commercial systems come with their own issues, however, such as limited battery life [21], lower channel count, and reduced reliability compared to research and clinical grade systems; for example, the Muse and Mindwave devices are more prone to eye blink and muscle artifacts compared to the medical grade B-Alert and Enobio EEG devices [33]. Recently, many commercial EEG companies have been expanding the use of their devices to a variety of applications, including detecting changes in emotion, attention, meditation, and mental workload [22]. Notably, the OpenBCI system is open source and modular, and has been customized for research studies in with a variety of applications, including in BCI [34], emotion recognition [35], and quantifying pain perception [36].
Table 1.3: Low-cost Commercial EEG Systems

<table>
<thead>
<tr>
<th>Product</th>
<th>Channels</th>
<th>Sampling Rate (Hz)</th>
<th>Price (USD)</th>
<th>Battery Life (h)</th>
<th>CMRR (dB)</th>
<th>References</th>
</tr>
</thead>
<tbody>
<tr>
<td>EMOTIV EPOC Flex Gel</td>
<td>32</td>
<td>128</td>
<td>2099</td>
<td>9</td>
<td>Not available</td>
<td>[37]</td>
</tr>
<tr>
<td>OpenBCI Cyton and Daisy</td>
<td>16</td>
<td>125/250</td>
<td>1999</td>
<td>26 [27]</td>
<td>-110 [38]</td>
<td>[37]</td>
</tr>
<tr>
<td>Muse</td>
<td>4</td>
<td>256</td>
<td>249.99</td>
<td>5</td>
<td>Not available</td>
<td>[39]</td>
</tr>
<tr>
<td>Neurosky Mindwave Mobile 2</td>
<td>1</td>
<td>512</td>
<td>109.99</td>
<td>8</td>
<td>Not available</td>
<td>[40]</td>
</tr>
</tbody>
</table>

Many lower cost wearable EEG systems focus on stationary applications; from (Table 1.3) the EMOTIV EEG system focuses on BCI applications, the Muse device focuses on sleep and meditation [39], and Neurosky devices focus on attention, meditation, and mood [41]. While some higher quality mobile EEG systems have some focus on sports and mobile applications, there is a focus on activities that provide less vigorous motion. For example, the ANT Neuro eego rt sports device has been used in studies with participants on cycling machines [17] as well as soccer kicking [42]. The focus on stationary applications for low-cost wearable EEG systems, and the focus on lower motion sports scenarios for higher cost, higher quality wearable systems
suggests that signal quality during mobile measurements is limited for current devices on the market. With both commercial and research-grade wearable EEG systems, while the hardware has been developed to allow subjects to move around during recording, this motion can lead to motion artifacts in the signal, which current commercial and research-grade EEG systems have not been designed to handle.

1.4 EEG electrodes

The main components of an EEG system are the electrodes, cap, and amplifier/data acquisition device. Many types of electrodes have been developed for measuring EEG signals; the electrode used can have a significant impact on the EEG signal quality. In electrode design, parameters such as mass and surface area of the electrode can have significant effects on EEG signal [43].

The current gold standard for non-invasive EEG are wet electrodes, because they provide a good signal and low electrode-skin impedance, typically less than 5 kΩ [1]. Electrode-skin impedance refers to the impedance measurement at the connection between the electrode and the outer layer of the scalp. An increase in electrode impedance leads to lower signal quality by decreasing the CMRR of the system, meaning that the output signal will contain more noise [44]. This is seen in [44], where high-impedance sites on the head led to an increase in low frequency noise.

Traditional wet electrodes consist of a gold cup or Ag/AgCl electrode, and a conductive gel that forms an electrochemical connection between the electrode and the scalp [45]. However, there are known disadvantages to wet electrodes, such as long setup and clean up time as well as shortened recording times due to gel drying. Some other disadvantages include potential bridging issues between gel electrodes if they are too close together, and irritation to the subject’s scalp due to the skin abrasion and gel [1]. To address these issues, semi-dry and dry electrodes have been developed.
Semi-dry electrodes provide a small amount of electrolyte liquid to the scalp for local skin hydration [1], this liquid also forms the electrochemical connection between the electrode and scalp. The semi-dry electrode is a middle ground between the wet and dry electrode, where less electrolyte is required compared to the wet electrode, but more electrolyte is required compared to dry electrodes. For dry electrodes, no electrolyte is provided, so the electrical connection consists mainly of electrode contact to the scalp; however, there may be trace amounts of sweat or moisture on the scalp which provide some conductive properties [1]. Development of semi-dry and dry electrodes have become more popular and improved with time, but there are still issues associated with these types of electrodes. While there are many advantages to semi-dry electrodes, such as having impedances similar to wet electrodes (typically below 30 kΩ [46]–[49]), reduced setup and clean up time compared to wet electrodes, and reliable signal quality [1], there can still be comfort issues [45], and having to refill the electrolyte or self-apply the electrodes can be a complicated/tedious task or even lead to errors [1], [45]. As well, many semi-dry electrodes have a bulkier design compared to the wet and dry electrodes, due to requiring a mechanism for electrolyte release.

With dry electrodes, there are many advantages, such as reduced preparation time, easier electrode and cap self-application, and reduced clean up time after measurements, since no electrolyte is required [45]. Many different types of dry electrodes have been developed, such as metal tip electrodes [50], [51], spring-loaded electrodes [52], [53], bristle electrodes [54], claw-type electrodes [55], [56], and flexible comb shaped elastomer electrodes [57]–[59]. However, most electrodes have impedances in the range of 100s of kΩ [60]; for example, even with recent developments in electrodes, electrodes in [57] have impedances of 601.5 ± 400.8 kΩ, and in [61] it was mentioned that electrodes under a force of 2N have less than 250 kΩ impedance. While
the 3D printed electrodes in [62] are promising, reporting impedances comparable to wet electrodes at 1-3 kΩ, they have only been tested on one human participant, and most of the other tests are on a head phantom. It is likely more human testing is required, as the impedance can vary depending on hair length and type. In papers comparing performance between gel and dry electrodes, the gel electrode performance is typically superior, such as in [57], where there was significant decrease in signal-to-noise-ratio (SNR) for dry compared to gel electrodes: dry SNR was 5.50±3.30, compared to 6.97±4.75 for gel electrodes. As well, in [63] it was noted that the number of faulty electrodes during recording was higher for dry electrodes, and this was similar in [57], where the dry EEG system had 10.5% ± 5.7% data rejection for every subject compared to 3.9% ± 2.7% for the gel system. In addition, dry electrodes are more sensitive to noise and motion artifacts when compared to the wet and semi-dry electrodes; this can significantly impact signal quality [64]. As such, there is a need to further evaluate the effect of dry electrodes on the recording of EEG signals. Overall, there are currently still many limitations with dry electrodes that hinder their widespread use in EEG systems, namely, the higher impedance, difficulty getting measurements through the hair, and electrode discomfort.

1.5 Electrode Testing and Characterization Methods

1.5.1 Motion Testing

The main challenge in wearable or mobile EEG sensing is that motion artifacts can introduce substantial noise that masks the underlying EEG signal [65], in addition to power line noise (at 50Hz or 60Hz), muscle, eye movement, and heart artifacts [66]. Motion artifacts during EEG can be substantial even during low intensity ambulatory activities and are difficult to remove because they appear in a similar frequency band as EEG but with higher amplitude [67]. These motion artifacts are thought to arise from electrode or cable movements that occur while the subject is
moving [68]. Accurately characterizing motion artifacts will help understand signal quality and ensure EEG data are not misinterpreted [68]. While some papers have conducted EEG measurements during low to medium intensity movement (e.g. walking, stationary cycling) and applied artifact removal techniques [69]–[71], fewer studies include other common and more intense movements (e.g. running and jumping) [72].

There is also limited characterization and comparison of motion artifacts across electrodes, especially for dry electrodes. In a study by di Fronso et al., a dry electrode system developed by ANT Neuro b.v., measured EEG during cycling on exercise bikes [73], and in a study by Oliveira et al., different gel (Biosemi Active Two and Cognionics Mobile) and dry (Cognionics Dry) systems were compared during treadmill walking [74]. While the study by di Fronso et al. found that dry and gel electrode caps were similar in terms of signal characteristics, information content, and system comfort [73], Oliveira et al. concluded that dry EEG systems will need to be improved to meet the current wet electrode standards. One major difference between the studies is the physical activity during EEG measurement; while [73] looks at cycling, [74] looks at treadmill walking. Treadmill walking involves impacts that are not present during cycling, which could lead to more motion artifacts. In [65], head accelerations during different walking speeds were measured, and at speeds of 3.0km/h and 4.5km/h, prominent spikes were seen in the power spectra at the stepping frequency, which could induce motion artifacts in the EEG signal.

Something to note is that both treadmill and cycling studies used research-grade systems, which, again would be risky in more vigorous motion scenarios.

While there are some previous studies looking at quantifying motion artifacts in EEG using IMU motion data [65], [75], [76], they do not look at high intensity motion scenarios, focusing on ambulatory movement. The IMU sensor is instrumented on the participant as a mouthguard.
IMU, which has been shown to accurately measure head impacts and motion [77]. Mouthguard IMU sensors have also been shown to detect head motions more accurately during intense movement compared to patch sensors on the head or sensors on a helmet [78]. Through our research, we would like to look at how using different types of EEG electrodes (gold cup, gel, and dry) affect the EEG in states of rest and various levels of motion. Our hypothesis is that during resting state, all 3 electrode types will have similar signals, but that the signal from the dry electrode will be most affected during movement and will have a higher correlation with the acceleration data when compared to the gold cup and gel electrodes.

1.5.2 Electrochemical Impedance Spectroscopy (EIS)

EIS is a method used to measure the impedance of a given sample through a range of frequencies to see the change in impedance over frequency. This allows for non-destructive characterization of the sample, and in the case of EEG, can help characterize the electrode-skin interface. In comparison to simple impedance measurements using an EEG system, EIS can characterize the electrode impedance through a range of frequencies as opposed at just one fixed frequency, and the EIS setup also allows for repeated measurements in a controlled environment, allowing for a better comparison and repeatability studies. For example, in [79], EIS is used to look at the performance of different EEG electrodes with repeated use. While the electrode impedance at a single frequency using the EEG system was consistent after many uses, EIS found that there were significant changes in the electrode-electrolyte impedance across frequencies after 20 uses with the Ag/AgCl electrodes, and 10 uses with the gold cup electrodes. EIS is typically performed by applying a voltage at a specific frequency to the system, and a current at the same frequency but different amplitude and phase is measured. Figure 1.2 shows a schematic of the
EIS process; there is voltage sine wave input, and current sine wave output with same frequency, but amplitude change and phase shift. Equation 1 shows the impedance calculation.

\[
Z = \frac{E_t}{I_t} = \frac{E_o \sin(\omega t)}{i_o \sin(\omega t + \phi)} = Z_o \frac{\sin(\omega t)}{\sin(\omega t + \phi)} \\
\text{Eq. 1 [81]}
\]

### 1.5.2.1 Main Components of EIS

The description of the main components in an EIS setup will focus on the common 3 electrode setup, which consists of the reference electrode, the working electrode, and counter electrode. The reference electrode is always at constant potential, and the working electrode is the electrode or sample of interest that will be measured. The voltage is applied between the reference and working electrode. Since the reference electrode stays constant, the change in potential is all through the working electrode. The counter electrode is the current carrying electrode, it helps close the electrical circuit, and keeps the reference electrode at constant potential by ensuring there is not significant current passing through the reference. The electrodes are all immersed in an electrolyte solution, which helps connect them electrically [58].

### 1.5.2.2 Equivalent Circuit Model

The sources of impedance in an EIS system are important for data fitting in the impedance determination process. The EIS curves in the Nyquist plot are fit to an equivalent circuit model to describe the specific system designed. This includes the elements mentioned in the previous
section being modelled as resistors and capacitors, and Figure 1.3 shows some examples of models and the data that would fit those models. Fitting is important because it helps extract the values of specific components in the system. More introductory material on EIS is provided in Appendix A.

Figure 1.3 Various equivalent circuit models and their Nyquist plots: permissions to use figures from [82] obtained from author

1.6 Summary of Gaps

Traditional wet EEG electrodes have issues such as long setup and clean up time, gel drying for long recordings, electrode bridging, and skin irritation due to the skin abrasion required for low impedance. To counter these issues, semi-dry and dry electrodes have been developed. Semi-dry electrodes address many of the wet electrode issues but have problems including discomfort and
bulky design due to requiring an electrolyte reservoir. Moreover, refilling the reservoir can be tedious. Dry electrodes address many of the wet and semi-dry electrode issues, but typically have lower SNR, higher impedance, and higher sensitivity to motion artifacts. In Chapter 2 of this thesis, we will go through the dry electrode design, development, and validation process with the aim of developing dry electrodes that address the concerns mentioned previously.

Current traditional clinical EEG systems are expensive, and require a bulky amplifier, which restricts a subject’s movement. The development of clinical and research grade wearable EEG allows for patient mobility, with the tradeoff of lower sampling rates, lower channel counts, and limited battery life. In addition, research grade systems can be costly as well. Low-cost commercial wearable systems are available which provide light-weight electronics, but have lower channel count, sampling rate, and reliability compared to research grade mobile systems. While there are currently mobile EEG systems with wet and dry electrodes on the market, there have not been many studies on how these systems and the EEG signal quality are affected during high motion scenarios, especially in systems with dry electrodes. To address this gap, in Chapter 3 we will compare gold cup, gel, and dry electrodes in a wearable EEG system during various motion scenarios.
Chapter 2: Dry Electrode Design and Fabrication

This chapter will focus on the design and fabrication process of the EEG dry electrodes, from the factors and considerations that led to the final design, to the testing required to validate the electrodes for further testing.

2.1 Dry Electrode Design

2.1.1 Design Requirements

Through EEG dry electrode literature reviews [45], [59], [83], [84], and some preliminary electrode design and testing, we found that some important design considerations for dry electrode usability in a wearable system are impedance, signal quality, size/geometry, and comfort (see Table 2.1). With some of these requirements, there is a trade off in mind for the electrode design. For the size of the electrode, the larger the electrode (with more tips), the lower the electrode skin impedance, since there is more contact between the electrode and scalp. However, if the electrode is too large, it may reduce the resolution of the EEG signal, so that EEG from multiple locations on the head is collected into one channel. This hinders the opportunity to measure in more locations on the head. In terms of comfort, typically dry electrodes will have lower impedance with increased pressure applied on the electrode to the head [61], but there needs to be a balance between the impedance and the comfort of the electrode for the participant. Through adjusting the size and comfort of the electrodes within specifications by using different electrode materials and shapes, we aim to achieve electrode impedance and signal quality that meets the requirements.
Table 2.1 Dry Electrode Design Requirements

<table>
<thead>
<tr>
<th>Property</th>
<th>Specification/Requirement</th>
<th>Justification</th>
</tr>
</thead>
<tbody>
<tr>
<td>Impedance</td>
<td>&lt; 100 kΩ for both measurements on the skin and through the hair without any skin preparation or skin abrasion</td>
<td>Most dry electrodes are on the order of 100s of kΩs [57], [60], [61]</td>
</tr>
<tr>
<td>Signal Quality</td>
<td>&gt; 0.8 correlation with gold standard wet electrodes when subject is stationary</td>
<td>The dry electrode should provide similar signal quality to wet electrodes, from previous papers [50], [85], dry electrodes with correlation greater than 0.78 during resting provide similar signals</td>
</tr>
<tr>
<td>Alpha Attenuation Coefficient</td>
<td>Have an alpha attenuation coefficient larger than 1 at the occipital locations on the head</td>
<td>It has been found that in the occipital region of the head, the alpha power of the EEG signal during eyes closed measurement is higher than the alpha power during eyes open measurement [86]</td>
</tr>
<tr>
<td>Property</td>
<td>Specification/Requirement</td>
<td>Justification</td>
</tr>
<tr>
<td>----------------</td>
<td>---------------------------------------------------</td>
<td>--------------------------------------------------------------------------------</td>
</tr>
<tr>
<td>Electrode Size</td>
<td>&lt; 15mm diameter, &lt;10mm height</td>
<td>The electrode should be within the range of sizes of current electrodes in development [45], [57], [61], [62]</td>
</tr>
<tr>
<td>Comfort</td>
<td>Is less than 5 for comfort score on a scale of 0 to 10, where 0 feels like there is nothing on the head and 10 is too uncomfortable to wear for more than 10 min</td>
<td>The electrode should be comfortable to wear</td>
</tr>
</tbody>
</table>

Two different types of electrodes were developed, one for mounting on the forehead, and one for mounting in the hair. The one on the forehead was more straightforward to develop, therefore the development process of the electrode in the hair will be discussed in more detail.

### 2.1.2 Geometry

We tested multiple sizes and shapes of dry electrodes. We selected the comb-shape form factor for the through-hair electrodes because we discovered that narrower tips were better able to penetrate through the hair, and having many tips allowed for an increase in surface area contact between the electrode and scalp, providing lower impedances and better signal. In terms of size, the goal was to create a small enough electrode to easily fit in a cap, while still providing enough skin-electrode contact area to provide a good signal. Typically, the larger the contact area, the better the signal will be. Initially, the electrodes had a circular base diameter of 1.5cm, and a
height of 0.7cm. Some different electrode shape design iterations are shown in Figure 2.1A. The electrode shape that provided the best impedance and signal is highlighted in the red box, and a computer aided design (CAD) representation is shown in Figure 2.1B. We found that there needed to be a compromise between having the radius of the tips be large enough to provide reasonable electrode-scalp contact surface area, and having the tips small enough so that they could easily penetrate through the hair layers to the scalp. Many previous designs had tips that were too large and did not allow the electrode to get to the scalp, leading to higher impedance measurements and lower signal quality. For this electrode, we were able to reach an average impedance of $153.96 \pm 82.87$ kΩ for hair measurements.

![Figure 2.1](image)

**Figure 2.1**: (A) Various electrode shapes tested with red square highlighting chosen electrode (B) CAD drawing of chosen dry electrode

### 2.1.3 Material

Initially, the dry electrode was made using the polyurethane rubber PMC-780 mixed with 12% carbon black conductive powder. PMC-780 was chosen because it has a hardness of 80A, which is within the range of polyurethane hardness mentioned in other dry electrode design papers [57], [61], [87]. PMC-780 requires mixing a ratio of 2:1 of part A liquid to part B liquid of the urethane and then allowing it to cure and solidify. This 2:1 ratio was another reason for choosing
this polyurethane, since the carbon black powder increases the viscosity of part A when it is mixed in. With a larger ratio of part A, the mixture was less viscous and easier to mix. However, even with the larger ratio of part A to carbon powder, the mixture was quite difficult to mix at 12% carbon. The electrodes also became stiffer and brittle with the addition of carbon powder, which hindered their ability to weave through hair to the scalp. We also found that even with 12% carbon, the electrodes had impedances above what could be measured by our system, and there was poor signal quality.

After these issues, we decided to change the mechanism of signal conduction from conductive powder to a conductive silver thread that would be embedded into the electrode. This led to lower impedances and improved signal quality but increased the fabrication time of the electrodes. The initial process of 3D printing the electrode shapes, and then developing the mold is time consuming in the beginning, but after having a set electrode shape, the molds could be used repeatedly for electrode fabrication. The main bottleneck in the electrode fabrication process is the sewing of the conductive thread through the mold; this had to be repeated with every electrode fabricated and took approximately 10 minutes per electrode. The rest of the process, which consists of pouring the polyurethane mixture into the molds to form the body of the electrode and coating the electrode tips with conductive silver was scalable as well. Initially a small amount of carbon black (1%) was still added to the electrodes for aesthetics, but since the signal conduction method had pivoted to the conductive thread, the carbon black was phased out of the fabrication process.

2.1.4 Connector

Initially, when the electrodes were made with PMC-780 and 12% carbon, a snap button connection was used, and even after switching to the conductive thread to measure the signal, the
same button connection was used. An image of the snap button connection is shown in Figure 2.2A-C. However, the button had a bulky design, and the materials for the button were difficult to consistently source. We then switched to a Dupont connection (Figure 2.2D), which provided a sleeker design and was more accessible.

![Image of electrode connections]

Figure 2.2: (A) View of electrode tips (B) View of back of electrode (C) View of electrode with snap button connection (D) View of electrode with Dupont connection

2.1.5 Conductive coating

For the first iteration, no coating was used, as the conductive carbon black in the electrode was supposed to conduct the EEG signal. After we switched over to using the conductive thread, we began coating the electrode tips to increase the contact surface area between the electrode and the scalp, as well as electrically connect the electrode-skin interface to the thread. Through review of previous literature, we found that Ag/AgCl electrodes provide low and reliable resistance, drift, and noise level when compared to gold plated silver, platinum, stainless steel, and tin electrodes [88]. However, the Ag/AgCl electrode coating process requires specialized equipment and training, and Ag/AgCl flexible ink is expensive [62]. Silver coated electrodes
have reduced performance compared to Ag/AgCl electrodes, but perform better than the gold-plated silver, platinum, stainless steel, and tin electrodes. The silver paint is lower cost and simplifies the fabrication process [62]. We first tried coating with the Fast-Drying Silver Paint from Ted Pella, and while it provided decent signal and impedance measurements in the hair of 129 ± 33.77 kΩ, the silver paint rubbed off after 1-2 uses. We then tried a silver epoxy (8331D - Silver Conductive Epoxy Adhesive from MG Chemicals), which improved the durability of the coating, but was not as conductive as the silver paint. We then tried using both, where we first placed a layer of silver epoxy and then the fast-drying paint; while this improved conductivity, the epoxy was a hard and uncomfortable on the scalp. We then tried a silver acrylic paint (842AR - Super Shield Silver Conductive Paint from MG Chemicals) and a silver polyurethane paint (842UR - Package Level Shielding from MG Chemicals). We found that the silver acrylic paint was not very conductive or durable, but that the polyurethane paint was conductive, providing an average impedance of 133.62 ± 67.17 for the larger Dupont electrodes, and could last at least 4-5 uses before starting to wear out. Note that similar conductive paints and coatings have been tried in [62], although the silver paint they used was durable enough to be incorporated into their final 3D printed electrode design.

2.1.6 Final Design

Two electrodes were developed, one for measuring signals on the forehead, and one for measurements in the hair. The forehead electrode is flat, and 3D-printed using a conductive filament (Eel™ TPU 3D printing filament from NinjaTek). This filament consists of carbon black in polyurethane, which provides its conductive properties. This electrode provides enough skin-electrode contact area to provide low impedances and good signal. The entire fabrication process for the flat electrode took 15 minutes for the 3D printer to print the electrode shape, and
a few minutes for a Dupont connection to be crimped onto the end of the thin cylindrical extension of the flat electrode. Dimensions and images of the electrode are shown in Figure 2.3A-B.

The hair electrode has 9 pins in a grid pattern, with a small circular tip to increase the contact area between the electrode and scalp, while also providing more comfort compared to a sharp tip. The pins are spaced 3mm apart, center to center, and the base of each electrode tip is 2mm diameter, narrowing to a 1mm diameter rounded tip. The narrowing from the base to tip helps the electrode tips maneuver through the hair and to the scalp. The rounded tips of the electrodes are coated with a conductive polyurethane silver paint, which allows for an increase in conductive contact area between the electrode and scalp. The electrode is made with PMC-780 polyurethane rubber from Smooth-On, which has a Shore hardness of 80A [89]. Inside the electrode, conductive silver threads run through the electrode from the tip to the Dupont crimp connection on the other end, which can then be connected to a wire for measuring brain activity. While the initial fabrication process takes a week, once the electrode shape is set, the bottleneck of the process is the sewing required for the conductive thread in the electrodes; even with this bottleneck, approximately 30 electrodes could be fabricated within a week. Through optimizing the different electrode design parameters, we were able to develop electrodes with an impedance of $118.4 \pm 50.70 \text{ k}\Omega$. A more detailed fabrication process is provided in Appendix B. Dimensions and images of the electrode are shown in Figure 2.3C-D.
2.2 Electrode Validation and Measurements

2.2.1 Validation Testing and Comparison with Design Requirements

To test the dry electrodes, we measured EEG with a gold standard paste electrode and the dry electrode at similar locations on the head, looking at the impedance and signal quality of the electrodes. We were able to develop electrodes with impedances in the 5-30 kΩ range using the desktop EEG system (ANT Neuro Refa8), which is comparable to other dry electrodes currently in development [57], [60], [61]. In terms of signal quality, we initially looked at artifacts that could be measured very clearly, such as eye blinks and jaw clenches. For Figure 2.4, it is shown...
that just by inspection, the wet electrode (top) and dry electrode (bottom) provide similar signals. After the electrodes passed this test, we moved on to further testing.

For the next set of measurements, we compared the wet and dry electrode during resting state (eyes closed) EEG. We calculated the correlation between the dry and wet electrode signals, as well as the power spectrum. Figure 2.5A shows a comparison between the signal for the paste (gold cup) and dry electrode, and Figure 2.5B shows a comparison of the power spectrum between the paste and dry electrode. For the button connection type dry electrodes, the signal correlation was over 80% which meets the design requirements. Differences in the amplitude of the peak in Figure 2.5B could be due to slight differences in the location of measurement for the two signals; while the electrodes were placed within 1.5cm of each other, the location and direction of the two electrodes could lead to the reduced amplitude in the dry electrode. After this, we changed to the Dupont electrode connection, which provided similar results, and further testing with the Dupont electrode is mentioned below.

Figure 2.4: Comparison of wet and dry electrodes using eye blinks and jaw clenches with same scale
After the electrode showed promising results from above, we did a comparison with gold cup, gel, and dry electrodes. We measured resting-state EEG in a human participant (UBC Ethics Protocol H20-02313) using gold cup electrodes with Ten20 conductive paste, flexible wet electrodes with conductive Signa gel, and flexible comb-shaped dry electrodes (Figure 2.6A). The electrodes were mounted at the Cz, O1, and O2 positions and measurements were conducted during eyes-open and eyes-closed conditions (ten trials each). The ANT Neuro Refa8 desktop EEG amplifier was used to record EEG signals from the electrodes and measure electrode-skin impedance. The data were bandpass filtered from 1Hz to 50Hz. Pearson’s correlation coefficient was calculated between the gold cup and flexible dry/wet electrodes. We computed the ratio of alpha power in the eyes-closed condition to the eyes-open condition (Figure 2.6B), or Alpha Attenuation Coefficient (AAC), as a common EEG biomarker to compare across electrodes. The flexible gel electrodes and gold cup electrodes showed comparable electrode-skin impedances, while electrode-skin impedances for the dry electrodes were an order of magnitude higher (Figure 2.6D). Comparing EEG signals, the gel electrodes had higher correlation with gold cup measurements ($r = 0.89$) compared to the dry electrodes ($r = 0.80$). For all three types of
electrodes, the AAC was greater than 1, although the ratio at the O1 position was greater for the gold cup electrodes compared to the wet/dry electrodes.

In addition, we observed that dry electrodes commonly showed signal artifacts related to variations in applied pressure (Figure 2.6C). Note that the impedances found in this set of testing are higher than the test mentioned above for Figure 2.4, mainly due to a slight difference in setup. For the test above, the electrodes were pressed against the scalp using hands, to allow for the subject to vary the pressure and impedance, but for the testing and results in Figure 2.6, a band was used to apply a more consistent pressure on the electrode to the head, leading to less pressure on the electrode in comparison to the testing in Figure 2.4, and higher impedance in the tests in Figure 2.6.

Figure 2.6: (A) Images of gold cup (top), gel (middle), and dry (bottom) electrodes, (B) EEG power spectrum comparison of these electrodes at O1 position, (C) EEG signal comparison of these electrodes, where inconsistency in dry electrode pressure led to changes in signal (D) Impedance, correlation and alpha power values for gold cup, gel, and dry electrodes across all trials.

In terms of design requirements, the dry electrodes do not meet the requirement that the electrode impedance is below 100 kΩ, but the dry electrodes do meet the signal correlation requirement, showing that the dry electrodes have good signal correlation with gold cup electrodes and could
effectively detect EEG biomarkers even with higher impedances. The dry electrode also meets the requirements for the AAC marker and electrode size. While the comfort metric is met during these experiments, comfort before and after testing for longer durations is further explored in Chapter 3. A summary of electrode testing results is provided in Table 2.2.

From this test, it is shown that both wet and dry electrodes can provide good signal quality in comparison to the gold cup electrodes based on signal correlation. However, pressure-related signal artifacts are common with flexible comb-shaped dry electrodes, which may be a major artifact source for mobile applications.

**Table 2.2: Design Requirements and Final Design Specifications Comparison**

<table>
<thead>
<tr>
<th>Property</th>
<th>Requirement</th>
<th>Final Design Specifications</th>
<th>Meets Requirements?</th>
</tr>
</thead>
<tbody>
<tr>
<td>Impedance</td>
<td>&lt; 100 kΩ</td>
<td>118.4 ± 50.70 kΩ</td>
<td>No</td>
</tr>
<tr>
<td>Signal Quality</td>
<td>&gt; 0.8</td>
<td>0.8 ± 0.1</td>
<td>Yes</td>
</tr>
<tr>
<td>AAC</td>
<td>&gt; 1</td>
<td>6.9 ± 2.4</td>
<td>Yes</td>
</tr>
<tr>
<td>Electrode Size</td>
<td>&lt; than 15mm diameter, 10mm height</td>
<td>13mm diameter, 6.5mm height</td>
<td>Yes</td>
</tr>
<tr>
<td>Comfort</td>
<td>&lt; 5</td>
<td>4.63 ± 2.07</td>
<td>Yes, for some subjects</td>
</tr>
</tbody>
</table>

Table 2.3 shows a comparison of our dry electrodes with similar electrodes in literature, which are also applied to the scalp without skin prep. From this table we can see that our dry electrodes are among those with the lowest impedance.
Table 2.3: Comparison of Current Design with Similar Electrodes in Literature

<table>
<thead>
<tr>
<th>Dry Electrode</th>
<th>Impedance (kΩ)</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Multipin flexible electrodes</td>
<td>&lt;150</td>
<td>[90]</td>
</tr>
<tr>
<td>Active Dry electrodes</td>
<td>&gt;300</td>
<td>[51]</td>
</tr>
<tr>
<td>Electrode with carbon pin bristles</td>
<td>133</td>
<td>[54]</td>
</tr>
<tr>
<td>Multipin flexible electrodes</td>
<td>601.5 +/- 400.8</td>
<td>[57]</td>
</tr>
<tr>
<td><strong>Current Design</strong></td>
<td>118.4 ± 50.70</td>
<td></td>
</tr>
</tbody>
</table>

2.2.2 EIS Measurements

In addition to impedance, signal correlation, and AAC metrics, EIS is another method of validating electrodes. EIS provides a controlled environment for testing, allowing for more robust repeat testing. This technique also allows for the characterization of electrode impedance through a range of frequencies as opposed to at a single frequency provided by most EEG amplifiers, providing more information on the changes of electrode impedance with changes in signal frequency.

2.2.2.1 Materials and Methods

Figure 2.7 shows the current setup for EIS measurements. The working electrode is the polyurethane dry electrode. For the reference electrode, there is an Ag/AgCl electrode in potassium chloride solution, and for the counter electrode there is a platinum wire. Electrically connecting the three electrodes is a saline solution in the glass container. The third image in Figure 2.7 shows a top view of the system, and the gold prongs are where cables with clips are
attached, which lead into the potentiostat (VMP-300, Biologic Science Instruments), the machine used to acquire measurements.

![Diagram](image.jpg)

**Figure 2.7:** The EIS setup currently used for collecting measurements

### 2.2.2.2 Experimental Parameters

Common experimental parameters in an EIS setup can be divided into physical and potentiostat parameters. Some physical parameters include the distance of the electrodes from each other, and the concentration/type of electrolyte solution. For the electrode distance, the electrodes should be close enough that the solution resistance is not very high, but not so close that the current path is blocked. For electrolyte concentration/type, the solution used should be similar to the environment being simulated. For these experiments 0.9% saline was used, which is similar to the salt concentration in human sweat, and used in previous studies for EIS measurements [48], [91], [92]. Some potentiostat parameters include the frequency range and sine wave amplitude, which are values typically found in the literature, and can vary depending on the application. The DC offset refers to the DC voltage with which the sine wave is superimposed. Other parameters include the wait time for current stabilization before measurement (Te) and wait period before
each frequency measurement ($P_w$); these last two parameters were first varied to note the effect on the EIS measurement.

### 2.2.2.3 Results

Figure 2.8 compares the differences between cycles for the Nyquist plots when $Te$, the wait time before stabilization, is varied. The test was run for 3 cycles at different $Te$, and it seems that setting the wait time to 60s provided the most consistent results for the Nyquist plot, although it is more difficult to conclude that based on the bode plots (Figure 2.9).

Figure 2.10 looks at cycle variation when $Pw$, the wait period before each frequency measurement, is changed. The results are comparable between 1 and 10s, so 1s will be chosen since it reduces the measurement time. With the bode plot (Figure 2.11), again it is more difficult to see the differences between the cycles.

![Figure 2.8: Comparing Data for Different Te, Nyquist Plot, DC Voltage = 1V, Pw = 1s](image)

**Figure 2.8: Comparing Data for Different Te, Nyquist Plot, DC Voltage = 1V, Pw = 1s**
Figure 2.9: Comparing Data for Different $T_e$, Bode Plot, DC Voltage = 1V, $P_w = 1s$

Figure 2.10: Comparing Data for Different $P_w$, Nyquist Plot, DC Voltage = 1V, $T_e = 60s$
2.3 Conclusion and Next Steps

In this chapter, we focused on the design, development, and validation of EEG dry electrodes. For the design, we discovered key electrode specifications such as impedance, signal quality, electrode size, and comfort through testing and reviewing the literature. In terms of development, we summarized the changes in electrode geometry, material, connector, and coating as we iterated through designs. Concurrently with electrode development, electrode testing was performed to examine whether the electrodes met specifications. The final design of the electrode is the one that most closely matched the requirements.

Through the electrode design and testing, we were able to develop dry electrodes with a novel design and fabrication process; currently no other previous electrode designs in literature have embedded silver thread into their dry EEG electrodes. This embedded silver thread allows for a softer (Shore hardness 80A), flexible electrode that still provides comparably low impedances. For comparison, the soft multi-pin electrodes developed in [57] have Shore hardness ranging from 40A on the forehead to 80A on the back of the head with an average impedance of $601.5 \pm$
400.8 kΩ. In addition, using polyurethane silver coating allows for a soft coating which can deform with the electrode, reducing wear of the coating and providing more comfort compared to a harder coating. The electrode is also designed with a universal Dupont connector, making it easy to incorporate the dry electrode using jumper wires into any EEG measurement system. For electrode fabrication, the process is simple and requires few specialized procedures or equipment. The materials for the electrode are easy to source, and the fabrication steps are straightforward, allowing other research teams to replicate our manufacturing process. This manufacturing process is also scalable, with the main bottleneck being sewing silver thread through the mold. Besides the sewing, all other steps in the manufacturing process allow for large quantities for electrodes to be fabricated in a short time frame.

Currently, dry EEG electrodes in literature still provide worse performance and signal quality compared to wet electrodes. Through our novel electrode design and fabrication process, we provided new techniques for improving current dry electrodes, as a step towards dry electrodes that will provide comparable or better performance to current wet electrodes.

Continuing on from this, we will look at other methods to compare and characterize the dry electrodes. In Chapter 3 we will be comparing gold cup, gel, and dry electrodes in a lab-developed wearable system that will compare signal of the different electrode types during motion. Note that electrode comfort will be investigated in Chapter 3 as well.
Chapter 3: Wearable EEG System Electrode Validation Testing

This chapter focuses on the human participant experiments to compare the gold cup, gel and dry electrodes in a wearable EEG system during motion.

3.1 Methods and Materials

3.1.1 Participants

Eight healthy volunteers participated in the study (5 females and 3 males, with average age of 23 ± 2.1 years). Volunteers were excluded if they had any neurological or spinal illnesses. This study has been approved by the Research Ethics Board at the University of British Columbia (H20-02313). All participants provided written informed consent for the study.

3.1.2 Instrumentation

Three types of electrodes were compared in this study: gold cup, gel, and dry electrodes (Figure 3.1C). We used standard gold cup electrodes with Ten20 conductive paste as the gold standard with which the custom-made gel and dry electrodes were compared. Design variations were created for the gel and dry electrodes— one type for the forehead, and one for the hair. The gel electrodes were 3D -printed using NinjaFlex TPU by NinjaTek; we used the filament with a width of 1.75mm. This filament was used to create a circular gel reservoir that had an opening to allow a gold cup to be inserted. Super Visc conductive gel from BrainVision was used as the electrolyte. The difference between the gel electrodes on the forehead and in the hair were that the forehead electrodes were flatter, to accommodate the cap, which was typically tighter at the circumference; note that Figure 3.1C shows only the forehead gel electrode, since both gel electrodes look similar. The forehead dry electrodes have a flat circular shape, with a 14mm diameter and 3mm height. The dry forehead electrode was created by 3D printing EEL Conductive Flexible Filament by NinjaTek. We used the midnight black colour filament, with a
width of 1.75mm. The flat shape allowed for enhanced comfort during testing. The dry EEG hair electrodes were developed by casting liquid polyurethane rubber (PMC-780 Dry from Smooth-On) and allowing it to cure in silicone molds (developed using Moldstar-30 from Smooth-On). The final electrode shape consists of a circular base and conical prongs with small spheres attached to the end. The dry hair electrode has a 13mm diameter circular base, and a height of 6.5mm. The narrowing of the tips helped the electrodes weave through hair, and the small sphere at the end helped increase contact surface area on the scalp and allow the silver conductive coating (842UR Silver Conductive Coating from MG Chemicals) at the tip of the electrode to have a stronger adhesion to the electrode. The two types of dry electrode design were selected because they are commonly applied for on-skin and through-hair applications [57], [59], [83], [84], [87], [93]. A custom-fit EEG cap (Figure 3.1B) was 3D printed for each participant using NinjaFlex TPU by NinjaTek, the flexible filament also used for the gel electrode reservoirs. For EEG signal amplification and data collection (Figure 3.1G), we used the OpenBCI Cyton board (Figure 3.1D) with the Daisy module (16 channels, 250Hz sample rate). We found that the dry electrodes required a separate reference from the gold cup and gel electrodes due to impedance differences. The gel and gold cup reference electrode was placed at FCz, and the dry reference electrode was placed nearby at Cz. Electrode positions used for testing are shown in Figure 3.1A. Each participant wore a boil-and-bite mouthguard with an IMU attachment (Figure 3.1E) at the incisor location (Invensense ICM20649). Linear acceleration data (Figure 3.1H) were sampled from the IMU at 1000Hz using an Adafruit feather M0 board. Another Adafruit feather M0 board was used to sync the data between the EEG and accelerometer signals by sending square wave pulses (Figure 3.1F). Start (2 pulses) and stop (3 pulses) buttons were used to denote the beginning and end of trials (Figure 3.1I).
3.1.3 Equipment Setup

Each participant was fitted with a custom cap with gel and dry electrodes in place. Some hair spreading was required for the dry electrodes, to ensure the electrode tips contact the scalp, however no other skin prep was used in electrode application. Gold cup electrodes were applied directly to the scalp after skin preparation. Gel electrodes were applied concurrently with the gold cup electrodes. When checking the electrode-skin impedances before measurement, all gold cup measurements were below 15 kΩ, gel measurements were below 20 kΩ, and the dry electrodes were below 250 kΩ. A modified GoPro strap was used to secure the Cyton/Daisy board, the Adafruit Feather M0 board, and the syncing board to the back of the participant with housings 3D printed using PLA filament.
Figure 3.1: (A) Gold cup, gel, and dry positions in experiment, (B) Experimental setup on participant, (C) Electrodes used in testing, (D) OpenBCI board used for data acquisition, (E) IMU mouthguard used to collect head accelerations, (F) Board used for syncing EEG and IMU signals, (G) Gold cup, gel, and dry electrode signals and accelerometer data, (I) Square wave syncing signal

3.1.4 Experimental Protocol

In the experimental session, we collected resting state measurements before and after motion testing. The resting state measurements included 3 minutes of eyes closed, then 3 minutes of eyes open measurements for the participants. During the eyes open condition, participants would fixate on an image of a cross on the wall. The motion testing included common mobile activities and exercises, including indoor activities (walking, jogging, jumping) and outdoor activities (walking, jogging). EEG and IMU data were collected throughout the experiment.
These trials tested the system during vigorous motion; a larger range of similar exercises is found in [94]. First the participant exercised on a treadmill at 4 tiers of speed: fast walk (4km/h-5.6km/h), fast jog (7.2km/h-8.9km/h), slow walk (2.4km/h-4km/h), and slow jog (2.4km/h-4km/h). The speeds for this experiment are within the range of speeds used in other EEG treadmill studies [95]–[97]. Participants were asked to choose a comfortable speed within each tier to accommodate differences in height and fitness level. The walk and jog trials were 1 min each. The jump trials each contained a set of 15 jumps. Each trial was repeated twice; with vigorous motion there are higher chances for issues with the wearable system which could impact data collection. Repeating the measurement allowed us to ensure we had a good trial and compare different trials for repeatability in high motion scenarios. For the outdoor activities, the participants walked/jogged on the sidewalk between two points with a known distance (57m). The walking trial was repeated once, and the running trial was repeated twice. The running trial had more repetitions to ensure we had similar length data compared to the walking trial.

3.1.5 Data Pre-processing

First, the syncing signal was used to section off and match up the EEG and IMU data. However, it was found that after matching the start syncing signal, the end syncing signal sent for the EEG and IMU had slight time differences; to address this issue, we scaled the IMU time vector to match the EEG signal before resampling from 1000Hz to 250Hz, so the EEG and IMU data would have matching sampling rates. This processing was performed in MATLAB. After syncing, both the EEG and IMU data were bandpass filtered from 0.1Hz to 50Hz; 0.1Hz was used to ensure the inclusion of the delta band of EEG while still removing the DC offset in the EEG signal. Some channels of data were removed if it was clear that the EEG had not been
recorded. EEG channels with an amplitude greater than 10000µV were rejected, or when possible, the section in the signal with larger amplitude was removed.

### 3.1.6 Data Analysis Methods

Average impedances were calculated across participants for the different electrode types and positions, as one metric for comparing the electrodes. We also calculated the average speeds and peak vertical accelerations across participants for the treadmill and outdoor walking and running trials to quantify the motion for different trials. Channel rejection metrics were also calculated, to provide insight into performance of different electrodes in varying motion scenarios. We also investigated electrode comfort by asking participants to rate the comfort of the dry electrodes, and then averaging these values across participants before and after motion testing to reveal changes in comfort with time. The alpha attenuation coefficient (AAC) was then calculated for the resting states before and after motion, and the percent difference between the gold cup-gel and gold cup-dry electrodes were calculated. This provided another comparison metric between electrode types. Pearson’s correlation coefficient was used to compare the gold cup electrode with the lab-developed gel and dry electrodes at the same position. The correlation coefficient was also used to compare the EEG and IMU signal as a method to quantify EEG motion artifacts. The vertical acceleration component was used for analysis, by first subtracting the gravity vector from the acceleration components, and then projecting the acceleration onto the gravity vector. Then, magnitude-squared coherence between the EEG and IMU signals was calculated, using the method of disjoint sections [98], [99] to calculate coherence across participants, for each trial and electrode position. The EEG IMU wavelet coherence was then calculated to see how coherence changes over time for each channel and trial.
For statistical analysis, z-scores were calculated for individual correlations to conclude whether the correlation values were significantly different from 0. First, the Fisher Transform was applied to normalize the correlation data. These values were divided by the variance to obtain a z-score. These values were then compared to the z-score for an alpha value of 0.05 of a two tailed test with Bonferroni correction for multiple comparisons, where the value 0.05 was divided by the number of comparisons performed. Correlations were deemed significant if their z score was outside the range of the alpha value z score; more detailed methods can be found in [100]. In comparing correlations, the $\chi^2$ test of homogeneity was used on both the EEG-EEG correlations and EEG-IMU correlations to determine whether the values for different electrode types were significantly different at an alpha value of 0.05 with Bonferroni correction; detailed description of the procedure can also be found in [100]. To investigate whether coherence values were significantly different from 0, coherence values were compared with an upper 95% confidence limit; values greater than this limit were considered significant; further details can be found in [98]. Gold-cup EEG-IMU coherences were compared with gel EEG-IMU and dry EEG-IMU coherences for significant differences, by converting the coherence values to a $\chi^2$ statistic, and applying the $\chi^2$ test of homogeneity; details of the procedure can be found in [98].

3.2 Results

3.2.1 Electrode Impedance, Participant Kinematics, Channel Rejection, and Comfort Scores

Electrode-skin impedance for the different electrode types was measured at the beginning of each experiment, and the average impedances and standard deviations are shown in Table 3.1 below. Note that while the forehead and hair had different types of dry electrodes, the results for both types of dry electrodes were grouped because the EEG signal patterns were comparable. A
closer look at the variation of electrode impedance with different hair types is provided in Appendix C.

Table 3.1: Average Impedances Across Participants for Different Electrode Types and Channels

<table>
<thead>
<tr>
<th>Channels</th>
<th>Fp1</th>
<th>C4</th>
<th>P3</th>
<th>O2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gold Cup (kΩ)</td>
<td>5.5 ± 4.53</td>
<td>4.25 ± 2.44</td>
<td>3.875 ± 1.62</td>
<td>4.75 ± 1.48</td>
</tr>
<tr>
<td>Gel (kΩ)</td>
<td>3.5 ± 0.87</td>
<td>5.25 ± 2.63</td>
<td>6.625 ± 2.12</td>
<td>8.25 ± 3.15</td>
</tr>
<tr>
<td>Dry (kΩ)</td>
<td>58.38 ± 69.42</td>
<td>123.13 ± 61.74</td>
<td>148.38 ± 43.68</td>
<td>131.13 ± 71.41</td>
</tr>
</tbody>
</table>

The average speeds for indoor treadmill and outdoor motions are shown in Table 3.2. The average peak acceleration for each activity is provided as well.

Table 3.2: Participants Average Speed and Average Peak Vertical Acceleration

<table>
<thead>
<tr>
<th>ID Fast Walk</th>
<th>ID Fast Run</th>
<th>ID Slow Walk</th>
<th>ID Slow Run</th>
<th>OD Walk</th>
<th>OD Run</th>
</tr>
</thead>
<tbody>
<tr>
<td>Average Speed (km/h)</td>
<td>5.25 ± 0.53</td>
<td>8.49 ± 0.55</td>
<td>3.66 ± 0.55</td>
<td>6.98 ± 0.47</td>
<td>5.27 ± 0.43</td>
</tr>
<tr>
<td>Average peak acceleration (m/s²)</td>
<td>7.88 ± 1.82</td>
<td>19.65 ± 4.51</td>
<td>5.52 ± 1.31</td>
<td>18.49 ± 4.79</td>
<td>8.91 ± 1.93</td>
</tr>
</tbody>
</table>

While no channels were rejected for the gold cup electrodes across all participants and trials, 1% of gel electrode channels were rejected, and 6% of the dry electrode channels were rejected,
based on the 10000µV threshold. Of the rejected channels, 48.3% of reject trials occurred during tuck jumps, 27.6% occurred during jumping jacks, 17.2% occurred during outdoor running, 3.4% occurred during the treadmill fast run, and 3.4% occurred during outdoor walking. During motion trials, the dry electrodes were prone to higher amplitude motion artifacts compared to the gold cup and gel electrodes. In terms of comfort scores, overall participants rated the dry electrodes to be 4.63 ± 2.07 before testing, and 5.25 ± 2.12 after testing, where a score of 0 feels like there is nothing on the head, and a score of 10 is too uncomfortable to wear for more than 10 min. The question asked to assess comfort is in Appendix D.

3.2.2 Resting State Alpha Attenuation Coefficient

For the AAC, the average percent difference for the gel electrodes was 0.80 ± 27.67%, whereas for the dry electrodes it was 21.12 ± 50.98% (Table 3.3).
Table 3.3: Alpha Attenuation Coefficient (AAC) Values During Resting State at Channel O2 for Participants

<table>
<thead>
<tr>
<th>Participant</th>
<th>Motion</th>
<th>Gold cup O2 AAC</th>
<th>Gel O2 AAC</th>
<th>Dry O2 AAC</th>
<th>Gold cup-Gel Difference (%)</th>
<th>Gold cup-Dry Difference (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>U1</td>
<td>Before</td>
<td>20.17</td>
<td>17.26</td>
<td>27.87</td>
<td>-15.51</td>
<td>32.08</td>
</tr>
<tr>
<td></td>
<td>After</td>
<td>7.51</td>
<td>6.31</td>
<td>11.99</td>
<td>-17.31</td>
<td>45.94</td>
</tr>
<tr>
<td>U2</td>
<td>Before</td>
<td>13.36</td>
<td>12.53</td>
<td>19.09</td>
<td>-6.43</td>
<td>35.34</td>
</tr>
<tr>
<td></td>
<td>After</td>
<td>7.27</td>
<td>8.70</td>
<td>12.99</td>
<td>17.89</td>
<td>56.43</td>
</tr>
<tr>
<td>U3</td>
<td>Before</td>
<td>4.77</td>
<td>5.25</td>
<td>4.57</td>
<td>9.72</td>
<td>-4.28</td>
</tr>
<tr>
<td></td>
<td>After</td>
<td>3.78</td>
<td>4.64</td>
<td>4.08</td>
<td>20.33</td>
<td>7.57</td>
</tr>
<tr>
<td>U4</td>
<td>Before</td>
<td>3.48</td>
<td>3.91</td>
<td>4.29</td>
<td>11.54</td>
<td>20.74</td>
</tr>
<tr>
<td></td>
<td>After</td>
<td>2.78</td>
<td>3.09</td>
<td>3.16</td>
<td>10.57</td>
<td>12.80</td>
</tr>
<tr>
<td>U5</td>
<td>Before</td>
<td>2.01</td>
<td>1.79</td>
<td>6.09</td>
<td>-15.45</td>
<td>195.71</td>
</tr>
<tr>
<td></td>
<td>After</td>
<td>2.67</td>
<td>2.50</td>
<td>3.34</td>
<td>-6.49</td>
<td>22.54</td>
</tr>
<tr>
<td>U6</td>
<td>Before</td>
<td>2.55</td>
<td>1.46</td>
<td>1.87</td>
<td>-54.43</td>
<td>-30.68</td>
</tr>
<tr>
<td></td>
<td>After</td>
<td>2.24</td>
<td>2.52</td>
<td>1.88</td>
<td>11.49</td>
<td>-17.38</td>
</tr>
<tr>
<td>U7</td>
<td>Before</td>
<td>2.54</td>
<td>2.25</td>
<td>2.36</td>
<td>-12.20</td>
<td>-7.21</td>
</tr>
<tr>
<td></td>
<td>After</td>
<td>2.25</td>
<td>2.34</td>
<td>2.18</td>
<td>4.18</td>
<td>-2.79</td>
</tr>
<tr>
<td>U8</td>
<td>Before</td>
<td>1.69</td>
<td>2.23</td>
<td>2.51</td>
<td>27.59</td>
<td>38.95</td>
</tr>
<tr>
<td></td>
<td>After</td>
<td>1.69</td>
<td>2.22</td>
<td>2.45</td>
<td>27.30</td>
<td>37.05</td>
</tr>
</tbody>
</table>

Mean Error 0.80 27.67
Std Error 21.12 50.98

3.2.3 Electrode Comparison During Motion Testing

We show example time domain plots of the signals in Figure 3.2. There are higher amplitude motion artifacts with the dry electrodes during motion scenarios, which seem to correspond with the IMU measurements in certain trials, such as the jumping jacks and tuck jumps.
Figure 3.2 Time domain EEG plots of gold cup, gel, and dry electrodes during select trials for participant 1
Figure 3.3A shows an electrode correlation heat map; correlations were calculated by concatenated signals from each participant together. Note that in the following figures, ID refers to indoor testing, and OD refers to outdoor testing. Figure 3.3B shows a heatmap with the results from the hypothesis testing. A “1” in the heatmap represents that the correlation value in Figure 3.3A is significantly different from 0, whereas a “0” depicts that the correlation is not significantly different from 0. This figure shows that the gold cup and gel electrodes had correlation coefficients ranging from 0.04-0.98, compared to the gold cup and dry electrodes with correlation coefficients ranging from 0-0.39, where most of these correlations were statistically significant.

![Figure 3.3A](image1)

**A**

<table>
<thead>
<tr>
<th></th>
<th>Eyes Closed</th>
<th>Eyes Open</th>
<th>ID Fast Walk</th>
<th>ID Fast Run</th>
<th>ID Slow Walk</th>
<th>ID Slow Run</th>
<th>Jumping Jacks</th>
<th>Tuck Jumps</th>
<th>OD Walk</th>
<th>OD Run</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fp1 Gold Cup-Gel</td>
<td>0.83</td>
<td>0.81</td>
<td>0.85</td>
<td>0.78</td>
<td>0.30</td>
<td>0.17</td>
<td>0.16</td>
<td>0.11</td>
<td></td>
<td></td>
</tr>
<tr>
<td>C4 Gold Cup-Gel</td>
<td>0.64</td>
<td>0.85</td>
<td>0.77</td>
<td>0.33</td>
<td>0.06</td>
<td>0.39</td>
<td>0.03</td>
<td>0.02</td>
<td></td>
<td></td>
</tr>
<tr>
<td>P3 Gold Cup-Gel</td>
<td>0.93</td>
<td>0.84</td>
<td>0.86</td>
<td>0.69</td>
<td>0.36</td>
<td>0.29</td>
<td>0.07</td>
<td>0.22</td>
<td></td>
<td></td>
</tr>
<tr>
<td>O2 Gold Cup-Dry</td>
<td>0.69</td>
<td>0.38</td>
<td>0.52</td>
<td>0.56</td>
<td>0.02</td>
<td>0.16</td>
<td>0.11</td>
<td>0.03</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Fp1 Gold Cup-Dry</td>
<td>0.33</td>
<td>0.45</td>
<td>0.73</td>
<td>0.64</td>
<td>0.11</td>
<td>0.13</td>
<td>0.00</td>
<td>0.30</td>
<td></td>
<td></td>
</tr>
<tr>
<td>C4 Gold Cup-Dry</td>
<td>0.70</td>
<td>0.50</td>
<td>0.88</td>
<td>0.26</td>
<td>0.01</td>
<td>0.01</td>
<td>0.06</td>
<td>0.00</td>
<td></td>
<td></td>
</tr>
<tr>
<td>P3 Gold Cup-Dry</td>
<td>0.66</td>
<td>0.28</td>
<td>0.88</td>
<td>0.50</td>
<td>0.07</td>
<td>0.00</td>
<td>0.01</td>
<td>0.10</td>
<td></td>
<td></td>
</tr>
<tr>
<td>O2 Gold Cup-Dry</td>
<td>0.64</td>
<td>0.50</td>
<td>0.89</td>
<td>0.60</td>
<td>0.10</td>
<td>0.05</td>
<td>0.10</td>
<td>0.07</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

![Figure 3.3B](image2)

**B**

In addition to calculating individual correlation statistical significance, the $\chi^2$ statistic was calculated to investigate whether the gold cup-gel correlation values at specific trials and locations were significantly different from the gold cup-dry correlation values. We found the
threshold using the $\chi^2$ distribution with 1 degree of freedom, at an alpha value of 0.05 with Bonferroni correction. From the $\chi^2$ testing, the correlations between the gold cup-gel and gold cup-dry electrodes are significantly different in all cases.

### 3.3 Electrode-IMU Comparison During Motion Testing

For the comparison between electrodes and acceleration movements, another heat map (Figure 3.4A) with the correlation values was created. In correlating the electrode EEG signals to the IMU signals, the gold cup and gel electrodes in general have lower values than the dry electrodes. Similar to the electrode correlations, Figure 3.4B shows a heatmap with the results from hypothesis testing. A “1” in the heatmap represents that the EEG-IMU correlation value in Figure 3.4A is significantly different from 0, whereas a “0” depicts that the correlation is not significantly different from 0. This figure shows that for the gold cup and gel electrodes, correlation coefficients were below 0.29 for all trials, whereas the dry electrodes had correlation coefficients up to 0.77. Correlations less than 0.01 were not significant, whereas values greater than 0.01 were significant.
Figure 3.4: (A) Heatmap of the EEG-IMU correlations, with x-axis depicting position and electrode type, and y-axis showing the activity (B) Heatmap showing which trials and electrodes had significant EEG-IMU correlation values.

For the EEG-IMU correlation values, another $\chi^2$ test of homogeneity was performed, to investigate if the gold cup-IMU correlation values were significantly different from the gel-IMU
and dry-IMU correlation values. Similar to the electrode correlations, we found the threshold using the $\chi^2$ distribution with 1 degree of freedom, at an alpha value of 0.05 with Bonferroni correction. From the $\chi^2$ testing, the correlations were significantly different in most cases, although there were some cases where the gold cup and gel electrodes were not significantly different, mainly in resting state scenarios. For the gold cup and dry electrodes, there is only one resting state trial, one jumping and one running trial where the correlation is not significantly different.

### 3.3.1 Magnitude-Squared Coherence

Through coherence analysis, some trends were seen between the electrodes for different types of motion. Below are select plots looking at the coherence values, which demonstrate the trends described. The remainder of the plots have been placed in Appendix E. For resting state trials (Figure 3.5), the coherence for gold cup, gel and dry electrodes with IMU is low, and from the 95% confidence limit calculated using methods found in [98], not significantly greater than 0 at most frequencies. For the walking trials (Figure 3.6), dry electrodes had significantly greater coherence at most frequencies below 20Hz compared to the gold cup and gel electrodes. The gold cup and gel electrodes did not have significantly different values at most frequencies for walking trials; one exception is at the higher frequencies (>30Hz) for the outdoor walking trials at electrode position C4 (in Appendix E, Figure E4.8). For the slow run trial (in Appendix E, Figure E4.6), the dry electrodes had higher coherence than the gold cup and gel electrodes at frequencies below 15Hz, but little difference above 15Hz. The gold cup and gel electrodes did not have significantly different coherence values at most frequencies for different positions on the head. For the faster running trials (Figure 3.7), some electrode positions showed significantly higher coherence in the dry electrode, while others showed the opposite. For most electrode
positions, the gold cup and gel electrodes did not have significantly different coherence values. For the jumping trials (Figure 3.8), at most frequencies there were not significant differences between the gold cup, gel, and dry electrode coherences.
Figure 3.5 (A) Eyes Closed Coherence values with 95% confidence limit of significant coherence (B) Chi-square values comparing homogeneity of gel and dry electrodes with gold cup electrodes, along with 0.05 significance level
Figure 3.6: (A) Fast Walk Coherence values with 95% confidence limit of significant coherence (B) Chi-square values comparing homogeneity of gel and dry electrodes with gold cup electrodes, along with 0.05 significance level.
Figure 3.7: (A) Fast Run Coherence values with 95% confidence limit of significant coherence (B) Chi-square values comparing homogeneity of gel and dry electrodes with gold cup electrodes, along with 0.05 significance level
Figure 3.8: (A) Jumping Jacks Coherence values with 95% confidence limit of significant coherence (B) Chi-square values comparing homogeneity of gel and dry electrodes with gold cup electrodes, along with 0.05 significance level
3.3.2 Wavelet Coherence

![Wavelet Coherence Plots]

Figure 3.9: Wavelet coherence plots for select trials and position; shown are the gold cup, gel, and dry C4 channels for participant U1

The wavelet coherence for each trial and participant were calculated and plotted. Select trials are shown in Figure 3.9. While there are some trends seen in terms of higher impact activities leading to higher coherence, looking at the shorter duration wavelet coherence plots in Figure 3.9 there is an inconsistency in the coupling between acceleration and coherence. For the fast walk trial, there appears to be higher coherence during peaks in acceleration in the dry electrode, but
this is not the case for the fast run and jumping jacks trial. There is also variability in the coherence within same participant trials, as well as between participants.

3.4 Discussion

3.4.1 Comparison of Electrodes

In terms of electrode-skin impedance, the gold cup and gel electrodes had similar impedance values, whereas the impedances for the dry electrodes were an order of magnitude higher; this is expected as dry electrodes typically have higher impedances than gold cup and gel electrodes, as seen in previous studies [57], [61].

From comparison of the alpha attenuation coefficient between gel and dry electrodes, the dry electrodes have larger difference and variation when compared to the gel electrodes. This suggests that even during resting state, the dry electrodes are more prone to noise and artifacts. The significantly higher correlation between the gold cup and gel electrodes compared to the gold cup and dry electrodes was expected as dry electrodes can be more prone to noise and motion artifacts compared to gold cup and gel electrodes. Electrode noise is inversely proportional to the area of electrode-skin contact, and smaller skin-electrode contact surface area for dry electrodes leads to more signal noise [21]. This is in addition to potential sources of noise from wires, and movement of wires leading to injection of voltage into the system, which will also affect the EEG reading [21]. Note that in looking at both the significance of the gold cup-gel electrode correlations and the gold cup-dry correlations, values above 0 are considered significant, even if they are close to 0. This is because of the large sample size inherent in collecting time domain signals with even a moderate sampling rate, which leads to weak correlations being detected with significance.
In terms of electrode comfort, the comfort score provided by participants for the dry electrodes before testing is below the requirement threshold of 5, but the score after testing is higher than 5. Further materials development may be required to improve the comfort of the electrodes for participants.

### 3.4.2 Electrode Motion Artifacts

In comparing the average walking and running speeds to the average peak vertical accelerations, the values are as expected, since with higher speeds, higher peak vertical accelerations were observed. In terms of channel rejection for the gold cup, gel, and dry electrodes, similar trends have been seen in [63], [73], [101], where dry electrodes have lower channel reliability, more faulty electrodes, and more motion artifacts in the EEG compared to wet electrodes. In looking at the trials with faulty dry and gel electrodes, the majority of occurrences were during higher intensity motion such as the tuck jumps and jumping jacks. The larger forces and higher impact motion on the electrodes and wearable system could lead to more disconnections. These disconnections could include between the electrode-skin interface, as well as the wired connections of the electrodes to the OpenBCI board for data acquisition.

The higher amplitude motion artifacts for the dry electrodes in comparison to the gold cup and gel electrodes shown in Figure 3.2 is likely due to the change in contact at the electrode-skin interface, which would occur more drastically in dry electrodes due to micromotions at the interface compared with the conductive paste and gel used for the gold cup and gel electrodes. While during the higher impact trials (such as jumping jacks and tuck jumps) the trends between the acceleration and EEG electrodes match for the dry electrodes, this does not seem to be the case for the walking and running trials, where in some instances there seems to be a phase shift or difference in directionality between the motion artifacts of the EEG signal and the
accelerometer. However, even with the phase shift, from Figure 3.2 both the dry EEG and vertical acceleration seem to have cyclic motion with similar periods, which could indicate some association between the signals.

Correlations between the EEG and IMU signals were low for most trials, and it was difficult to detect any trends, even though through applying statistical methods from [100] most results were considered significant. As mentioned in the previous section, this is likely due to the large sample size. Similar low correlation results are noted in [102]. This likely means that linear correlation may not be a suitable metric in characterizing motion artifacts. In [103], it is noted that the relationship between head movement and EEG motion artifacts is nonlinear, due to the interactions between cable bundles and cable sway. This means that the linear correlation metric may provide a limited and incomplete description of the relation between head acceleration and motion artifacts in EEG. We then looked at coherence analysis to see if this would provide more insight into the dataset; coherence is also mentioned in [103] as a method to find trends between EEG and IMU signals for different walking speeds.

At resting state, most coherences were not significantly different from 0, which is as expected since we expect there to be little motion artifacts when participants are stationary. For the walking trials, the dry electrodes have significantly higher coherence, but this is not the case for the running and jumping trials. The higher coherence in the dry electrodes for lower intensity movements compared to the gel and gold cup electrodes implies the gold cup and gel electrodes may provide a more stable electrode/electrolyte-skin interface that has a higher threshold to motion before micromotions occur, meaning the dry electrodes are more prone to motion artifacts. This higher susceptibility of dry electrodes to motion is noted in [74], where it is mentioned that the noisier signal may be due to changes in dry electrode contact impedance from
head motion and walking. The similarity in coherence for all types of electrodes for jumping scenarios could mean with motion that is vigorous enough, micromotions are induced in not only the dry electrodes, but the gel and paste electrodes as well. Typically, in studies focusing on more ambulatory walking and less vigorous motion such as in [65], there is a focus on the lower frequency bands such as the delta band; however, from our data, especially for dry electrodes, looking at frequencies up to 20Hz may provide a better model for motion artifacts in EEG.

The wavelet coherence plots help to provide a time-frequency visualization of how well the IMU and EEG data correlated over time. While some trials had high coherence corresponding to the peak accelerations measured by the IMU mouthguard, such as the fast walk for participant U1, other trials, such as the fast run and jumping jacks for participant U1 did not follow this trend. This leads us to believe that the real-time coupling between the IMU mouthguard and EEG electrode measurements are not always consistent, due to the varying coherence values at peak acceleration.

### 3.4.3 Implications for Wearable EEG Measurements

In this chapter, we looked at comparing gold cup, gel, and dry electrodes during motion scenarios using correlation, and characterizing motion artifacts in EEG during motion, through EEG-IMU correlation and coherence metrics. Through these metrics, we were able to develop an idea of the extent of motion artifact contamination in EEG signals during low and high intensity movement. The knowledge and dataset created through this study could be used for further analysis through the application of different motion artifact removal algorithms, now that there has been an improved understanding of the level of motion artifacts in EEG, as well as its relationship to accelerometer IMU data, especially in high impact scenarios. There are few papers that combine the use of EEG and IMU data for the purpose of motion artifact removal.
[65], [75]; this could be a key component and provide the information required to further improve existing motion artifact removal algorithms.

3.4.4 Limitations

While this study has led to the development of a dataset and motion artifact characterization techniques for low and high intensity motion, there are some limitations as well. One issue is that due to the restrictions of the hardware setup, there is a shared reference between the gold cup and gel electrode, whereas the dry electrode has a separate reference. Having a separate reference for the gold cup and gel electrode would provide a better comparison between the three electrode types, since the claim that the gold cup and gel electrode having higher correlation compared to the gold cup and dry electrode could be skewed. Electrodes sharing a reference may intrinsically have higher correlation. In this study, steps were taken to minimize this effect, by having the dry electrode reference as close to the gold cup and gel reference as possible. As well, currently the dataset only covers 4 electrode positions; increasing the number of positions in future studies could provide a more detailed picture of the difference in motion artifacts at different locations of the head. In addition, for the study, the treadmill speeds were tiered to accommodate the wide range of heights and different strides of individuals, and during the outdoor walking and running, participants were told to go at their natural pace. This allowed for more natural movement for the individual, but makes it more difficult to compare wavelet coherence plots, since for similar trials different participants were going at different speeds, leading to differences in the frequencies where high coherence values would occur. For future studies, increasing the sample size and age range of participants would provide a dataset that is more generalizable, since currently 8 subjects in the 18-35 year age range participated in the study. There may be differences in gait and motion for people outside of this age range [104], [105].
3.5 Conclusion

In this chapter, we compared gold cup, gel and dry electrodes in motion scenarios by instrumenting participants with a wearable EEG system, and having them perform exercises such as walking, running and jumping. We also instrumented participants with an IMU mouthguard to characterize motion artifacts in EEG signals for different types of electrodes. We found that in resting and motion scenarios, the gold cup and gel electrodes have a higher correlation compared to the gold cup and dry electrode. During motion, it was noted that the dry electrodes had larger amplitude motion artifacts and lower channel reliability. Through coherence analysis, it was also found that the dry electrodes in lower intensity motion scenarios have a higher correlation with head motions measured by the IMU compared to gold cup and gel electrodes; this suggests that dry electrodes are more prone to motion artifacts. The data and findings from this study have improved the understanding of motion artifacts in EEG, and in the future could be used to evaluate motion artifact removal algorithms.
Chapter 4: Conclusion and Future Work

4.1 Dry Electrode Design and Fabrication

In terms of dry electrode development, we were able to develop a dry electrode that met the electrode specifications of having a correlation between the dry and gold cup electrodes of greater than or equal to 0.8, and an electrode size less than 1.5cm in diameter and 1cm in height. However, our dry electrode did not meet the requirement of having an impedance less than 100 kΩ. The developed electrode had an average impedance of 114.9 ± 43.8 kΩ, a dry-wet electrode correlation of 0.8 ± 0.1, a diameter of 13mm, and height of 6.5mm. In terms of comfort, from the wearable EEG validation testing in Chapter 3, there was an average comfort score of 4.63 ± 2.1 at the beginning of the experiment, and an average comfort score of 5.25 ± 2.1 at the end of the experiment. While the comfort score at the beginning of the experiment meets the requirements, the comfort score at the end of the experiment is higher than 5. For future electrode development, there are a few directions that could be explored. To lower the electrode impedance further, the electrode geometry could be analyzed in more depth; for example, the electrode tips could be varied in shape or quantity to try and increase the electrode-skin contact area. The impedance may also change depending on the area of the head the electrode is on; variations in the electrode parameters such as pressure could be investigated at different positions on the head to provide optimal impedance. Another factor varying impedance is hair type; differences in impedance and signal due to hair type could be further investigated as well. To increase the comfort of the electrodes, the flexible material used to fabricate the electrode could be explored further to ensure better comfort for participants. This could include trying other polyurethane rubbers of different hardness, or developing electrodes with another type of flexible polymer such as silicone.
4.2 Electrochemical Impedance Spectroscopy

In terms of the current EIS results, while some parameters have been explored, such as Te and Pw, more experiments will need to be performed to investigate other parameters, such as varying the distance between electrodes and looking at the effect on measurements. After finding the optimal parameters, the data will be fit to different circuit systems, and further EIS experiments could include testing on a skin surrogate as opposed to just saline solution for a more accurate model. Some skin surrogates could include conductive gelatin, or even porcine skin. Based on the final shape of the Nyquist plot, a circuit model will be chosen to fit the system, to determine the impedance of the electrode through frequencies. With the gelatin or porcine skin models, there could also be investigations into the effect of pressure on impedance of the dry electrodes.

4.3 wearable EEG System Electrode Validation Testing

In this chapter, we compared gold cup, gel, and dry electrodes in motion scenarios, as well as characterized motion artifacts in the system through correlating EEG signals with IMU signals. Gold cup and gel electrodes have a higher correlation in comparison to gold cup and dry electrodes. As well, in comparison to the gold cup and gel electrodes, the dry electrodes had higher impedances, higher channel rejection rates, and are more susceptible to motion artifacts, especially in low motion scenarios; with large enough motion, all types of electrodes become susceptible to similar degrees of motion artifacts. Future directions of this work could explore different avenues. For wearable system development, we could improve the system based on current device limitations by redesigning or sourcing an EEG amplifier that can facilitate more EEG channels and references for the different types of electrodes. Some of the electronics and hardware such as the syncing board and the housings could be further reduced in size as well. In terms of experimental design, an IMU could be attached to the leg and synced up to the head
IMU and EEG signal to better understand the motion artifacts throughout the gait cycle. Increasing the sample size and age range of participants could provide better generalizability of the study as well as provide more dimensions for comparison across age groups. In terms of data analysis, different motion artifact removal methods could be applied to the current dataset, to investigate which techniques would allow for a cleaner signal of EEG during low and high intensity motion scenarios.

Overall, this thesis described the development and validation of a lab-developed wearable EEG system, with a focus on dry electrode development, as well as motion and electrochemical validation testing. In terms of current wearable system development, wet electrodes provide a better option for reducing motion artifacts in low to medium motion scenarios. In high motion scenarios, development is required to further minimize motion artifacts, either through further wet and dry electrode research, or the use of motion artifact removal algorithms after data collection. This thesis provides a step towards the aim of making it feasible to record high-quality EEG in motion scenarios outside of the lab. This development will expand the applications of EEG to scenarios that were difficult to measure with traditional systems, from ambulatory recordings in one’s home, to the use of EEG during exercise and sports.
References


cognition: EEG and sports performance,” *Neurosci Biobehav Rev*, vol. 52, pp. 117–130,

estimation of professional female soccer players,” *Health Inf Sci Syst*, vol. 9, no. 1, p. 14,

EEG in Sports Sciences: A Fast and Reliable Tool to Assess Individual Alpha Peak
Frequency Changes Induced by Physical Effort,” *Front Neurosci*, vol. 13, p. NA-NA, Sep.

[18] K. Reinecke et al., “From lab to field conditions: A pilot study on EEG methodology in
applied sports sciences,” *Applied Psychophysiology Biofeedback*, vol. 36, no. 4, pp. 265–

[19] G. Ottoboni et al., “Repeated Sub-Concussive Impacts and the Negative Effects of
Contact Sports on Cognition and Brain Integrity,” 2022, doi: 10.3390/ijerph19127098.


Passive Brain–Computer Interface Applications,” *Sensors (Basel)*, vol. 20, no. 16, pp. 1–16,


[31] “LiveAmp series | Brain Products GmbH > Solutions.”


[76] “Characterization and real-time removal of motion artifacts from EEG signals”.


[80] “EIS Basic Background Theory – Pine Research Instrumentation Store.”


EMBS, Nov. 2015, vol. 2015-November, pp. 7131–7134. doi:
10.1109/EMBC.2015.7320036.


Appendices

Appendix A  More EIS background

A.1  Linearity in EIS Systems

One of the main assumptions of EIS is that the system is linear. While this is typically not the case, with small enough voltage signal amplitude, the system can be approximated as linear. With too large of an amplitude in the voltage signal, the corresponding current signal may not be a pure sine wave, and instead a summation of sine waves of different frequencies, which would lead to the impedance measurement being inaccurate. Typically, a voltage amplitude on the order of 5-10mV is small enough that the region covered is linear, and therefore the current caused by this voltage is a pure sine wave at the same frequency [106].

A.2  Common Plots for EIS

Figure A4.1A and B shows some examples of common plots used to display the EIS data. Figure A4.1A shows a bode plot, which is typically the magnitude of impedance or phase vs. frequency, and Figure A4.1B shows a Nyquist plot, which is usually the negative of the imaginary impedance vs. the real impedance. For these plots, the equivalent circuit that generates the plot is shown in Figure A4.1C.
A.3 Sources of Impedance in EIS System

There are many different sources of impedance in the EIS system. One key source is the solution or electrolyte resistance, and in the case where instead of solution there is skin, it would be the skin resistance. The second is the double layer capacitance; this refers to the double layer of ions that is formed when the electrode is place in the electrolyte and is shown in Figure A4.2, which shows one layer of charges in the electrode, and another, opposing layer of charges in the electrolyte. Between these two layers, there is a layer of solvent molecules attached to the electrode surface, which separates the metal and electrolyte charge layers. This is what forms the double layer, which then provides a source of capacitive impedance in the system. Polarization resistance occurs when voltage is applied to the electrode, polarizing it, and causing current flow through electrochemical reactions at the electrode surface. Warburg impedance is a component that accounts for diffusion of ions in the system.
Figure A4.2: Schematic of double layer capacitance, permissions to use figures from [82] obtained from author.
Appendix B : Dry EEG Electrode Fabrication Protocol

- First, we need to create the electrode mold positive:
  1. In the CAD software of your choice, convert your part to a .STL file.
  2. Import file into Preform, the software used to connect to the Form3 Formlabs printer.
  3. Ensure the correct printer is selected (Form 2 vs Form 3) and the correct resin colour and type are selected, since curing times/temperature may vary depending on the resin type. Set the resolution size to either 25nm or 50nm, depending on the size of your part. Higher resolution (smaller dimensions) means that the print will take longer but will be more accurate. Remember to have the part at a 10°-25° angle and click the button to auto-generate supports for the part.
  4. 3D-print the electrodes using the Form 2 or Form 3 printer, an SLA printer which uses resin and UV light to print and cure.

  - Plug USB cable from printer into laptop
  - Use Preform software, under plugin, choose chiefgnu from list of printers
  - File->open->choose STL file you want, you can select multiple at once
  - Click on orientation signal, select base button, and click on base of part
  - Highlight part, right click, duplicate as needed
  - Click supports-> autogenerate
  - Click layout, then layout all -> can change spacing between the parts
  - Last button, rename file, then upload to printer (Look at buttons in order from top to bottom)
  - Printer: calibrate in settings, use levelling tool -> level
  - Jobs-> pick job->print now->confirm
5. After printing, remember to clean up the printer and parts
   - Check if there is enough IPA
   - Slide top with electrodes off, attach to stand by printer
   - Scrap off using the triangular flat tool
   - Put parts in suitable beaker, 3-4 washes, first wash with dirty IPA (1), then with (2)
   - Put in fume hood, let dry 30 min, or use air hose to dry
   - Wash the stand with IPA and place back into machine
   - Put part in cure box, then press start for 60C, 30min

6. The final result is that you should have 4 parts: the mold positive printed onto a flat base, the wall that will surround the positive in order to make the negative mold, and two clips that will keep the wall in place.

- Procedure to create mold:
  1. Assemble the 3D-printed pieces of the positive mold together.
  2. Give three quick sprays of release spray onto the surface the silicone mold material will be in contact with. (Optional for grey resin parts, but required for white resin parts)
  3. Add equal parts Part A and Part B of Mold star 15A Slow silicone from Smooth-On into a disposable cup and mix until thoroughly combined. It will be about 1.5g of silicone required for each electrode you cast. Once combined, place in vacuum chamber and degas until no bubbles appear very slowly in the mixture.
  4. Then pour the silicon into the mold positive until the tips of the electrode are completely covered and degas for another 3 minutes.
5. Leave the mold for 24 hours, then carefully take out the mold.

- Procedure to make electrodes
  1. Sew through the mold with a thin needle, and conductive thread
  2. Leave about 1-2cm of thread on the back side of the mold when threading back through
  3. Take half the threads in one hand, and the other half in the other hand, and twist the threads into a knot.
  4. Separate the threads into thirds, then braid the length of the thread.
  5. Now, the electrode molds with thread are ready to be cast with the polyurethane, to create the electrode shape.
     Typically, about 1.5g of total material is required per electrode but aim to mix at least 6g to ensure more accurate mixing.
  7. After mixing, place mixture in the vacuum chamber to degas, ensure the container can contain about double the amount of liquid PMC-780 that is currently in it; during vacuuming the bubbling mixture can double in size.
  8. After the bubbling slows, pour the PMC-780 into the molds, and level with a spatula.
  9. Let the electrodes cure for 48 hours, then carefully trim the threads and remove electrodes from the mold.
  10. Clean the surface of electrodes using the plasma machine.
  11. Dip tips of the electrodes in silver polyurethane paint.
  12. Let the electrode tips dry under a heat lamp for 15min, then place electrodes with tips down on a metal plate and place the plate on a hot plate set at 140C for 15min.
13. When the silver paint has cured, the electrodes are ready for testing.
Appendix C: Impedance Comparison for Different Hair Types

Figure C4.3 shows the plots of impedance for different hair parameters, across different electrode types and positions. From the current data, it is difficult to conclude any trends between hair length, type, and thickness and electrode impedance. A larger sample size is required to further investigate the trend between hair and electrode impedance.

Figure C4.3: Comparison of Impedance vs different hair parameters
Appendix D: Participant Comfort Questionnaire

1. On a scale of 0-10, where 0 is nothing on the head, and 10 is too painful to wear for more than 10min, how comfortable are the dry electrodes?
Appendix E: Coherence and Chi Square Significance Plots

Figure E.4.4: (A) Eyes Open Coherence values with 95% confidence limit of significant coherence (B) Chi-square values comparing homogeneity of gel and dry electrodes with gold cup electrodes, along with 0.05 significance level.
Figure E4.5: (A) Slow Walk Coherence values with 95% confidence limit of significant coherence (B) Chi-square values comparing homogeneity of gel and dry electrodes with gold cup electrodes, along with 0.05 significance level
Figure E4.6: (A) Slow Run Coherence values with 95% confidence limit of significant coherence (B) Chi-square values comparing homogeneity of gel and dry electrodes with gold cup electrodes, along with 0.05 significance level
Figure E4.7: (A) Tuck Jumps Coherence values with 95% confidence limit of significant coherence (B) Chi-square values comparing homogeneity of gel and dry electrodes with gold cup electrodes, along with 0.05 significance level.
Figure E4.8: (A) Outdoor Walk Coherence values with 95% confidence limit of significant coherence (B) Chi-square values comparing homogeneity of gel and dry electrodes with gold cup electrodes, along with 0.05 significance level
Figure E4.9: (A) Outdoor Run Coherence values with 95% confidence limit of significant coherence (B) Chi-square values comparing homogeneity of gel and dry electrodes with gold cup electrodes, along with 0.05 significance level