

Development of tDCS skin contact materials and their measurement setup

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Abstract

Transcranial Direct Current Stimulation (tDCS) is a non-invasive method that is used to modulate brain function for therapeutic and cognitive purposes. The distribution of electrical current at the interface of tDCS electrodes and skin depend on the quality of electrode-skin contact. There are opportunities for optimization in the design of "verification methods" of this contact for tDCS devices.

This thesis develops a verification testing model for assessing the electrical properties of tDCS electrodes. A grid-type electrode is developed for in vivo measurements to facilitate systematic analysis. SimNIBS software is used to evaluate the distribution of electrical current at the electrode-skin interface and then these findings are validated experimentally.

The verification testing aimed at developing a framework that could be used in identifying possible problems in electrode-skin contact and gaining an understanding of the influencing factors on current distribution. This model thus forms a basis for studying properties like current uniformity and user comfort, enabling future research to deal with issues like hotspots and discomfort due to poor current distribution.

While this thesis highlights areas where the tDCS electrode-skin contact design can still be improved, its main contribution lies in offering a systematic approach that could be used for electrode research and development in the future. The results form the basis for optimizing the structures and materials of electrodes methodically and will thus contribute to refining therapeutic stimulation technologies, and opening possibilities for more personalized treatments in the future.

The thesis comprises a literature review, methods, designing of the grid electrode, and simulation techniques. In addition, the design and implementation of in vivo experimental setup, which includes participant recruitment and ethical considerations, are discussed. It carries out in vivo measurement data analysis using the MATLAB software.

Results are integrated with prior knowledge and aimed to help in optimizing electrode design in tDCS. Conclusively, future directions will be suggested towards improved research with enhanced therapeutic effectiveness for widened clinical applicability with views to advance technology, tDCS improvements of outcomes, and improvements toward its widespread clinical use.

Keywords tDCS, depression, antidepressant treatment, neuromodulation, electrode material, biophysics, current distribution, focality



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Tiivistelmä

Transkraniaalinen tasavirtastimulaatio (tDCS) on ei-invasiivinen menetelmä aivojen neuronitaktiivisuuden muokkaamiseen terapeuttisissa ja kognitiivisissa hoitotarkeituksissa. Sähkövirran jakautuminen riippuu elektrodien ja ihon kontaktista ja laadusta. tDCS-laitteiden ihokontaktin "verifikaatiomenetelmien" suunnittelussa on parantamismahdollisuuksia.

Tässä diplomityössä kehitetään testausmalli tDCS-elektrodien sähköisten ominaisuuksien arvioimiseksi. In vivo -mittauksia varten kehitetään grid-tyyppinen elektrodi, joka mahdollistaa systemaattisen analyysin. SimNIBS-ohjelmistoa hyödyntäen työssä pyritään arvioimaan sähkövirran jakautumista elektrodi-iho-rajapinnalla ja validoimaan nämä havainnot kokeellisesti.

Verifikaatiotestauksen tavoitteena on kehittää viitekehys, jota voidaan käyttää tunnistamaan mahdollisia ongelmia elektrodi-iho-kontaktissa ja saamaan syvempää ymmärrystä virran jakautumiseen vaikuttavista tekijöistä jatkotutkimuksiin. Tämä malli luo siten perustan virran tasaisen jakautumisen sekä kohdentumisen ja käyttäjämukavuuden, tutkimiselle ja mahdollistaa tulevien tutkimusten keskittymisen entistä yksilöllisempiin hoitoihin.

Vaikka tämä työ tuo esille tDCS-elektrodi-iho-kontaktin suunnittelun parannusalueita, sen merkittävin panos on systemaattisen lähestymistavan tarjoaminen, jota voidaan hyödyntää tulevaisuuden elektrodien tutkimuksessa ja kehityksessä. Tulokset luovat pohjan elektrodien rakenteiden ja materiaalien optimoimiselle, mikä puolestaan edistää terapeuttisen stimulaatioteknologian hienosäätöä ja tarjoaa hoitoa erilaisille käyttäjille.

Opinnäytetyö sisältää kirjallisuuskatsauksen, kuvaukset menetelmistä, grid-elektrodin suunnittelun ja simulointitekniikat. Se sisältää myös in vivo -koeasetelman suunnittelun ja toteutuksen, mukaan lukien koehenkilöiden rekrytoinnin ja eettiset näkökohdat. Mittaustuloksia analysoidaan MATLAB-ohjelmistolla. Tulokset integroidaan olemassa olevaan tietoon tDCS-elektrodien suunnittelun optimoimiseksi ja stimulaatiomenetelmien kehittämiseksi. Työ päättyy ehdotuksiin tulevaisuuden tutkimussuunnista hoidon tehokkuuden parantamiseksi ja kliinisten sovellusten laajentamiseksi, pyrkien parantamaan tDCS-teknologiaa ja tarjoamaan edistysaskeleita laajemmissa sovelluksissa.

Avainsanat tDCS, masennus, masennuksen hoito, neuromodulaatio, elektrodimateriaali, biofysiikka, virran jakautuminen, fokaliteetti

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Symbols and abbreviations

Symbols

E	electric field strength
B	magnetic flux density
σ	conductivity [$\frac{S}{m}$]
ρ	resistivity [$\Omega \cdot m$]
ϵ	permittivity [$\frac{F}{m}$]

Operators

$\nabla \cdot \mathbf{A}$	divergence of the vector A
$\nabla \times \mathbf{A}$	curl of vector A
$\frac{d}{dt}$	derivative with respect to variable t
$\frac{\partial}{\partial t}$	partial derivative with respect to variable t
$\sum_i$	sum over index i
$\mathbf{A} \cdot \mathbf{B}$	dot product of vectors A and B

Abbreviations

tDCS	Transcranial Direct Current Stimulation
EEG	Electroencephalography
MRI	Magnetic Resonance Imaging
CNS	Central Nervous System
NIBS	Non-invasive Brain Stimulation
ROI	Region of Interest
TMS	Transcranial Magnetic Stimulation
DLPFC	Dorsolateral prefrontal cortex
FEM	Finite Element Method
CSF	Cerebrospinal fluid

1 Introduction

Transcranial Direct Current Stimulation (tDCS) is a non-invasive neuromodulation method, used in therapeutic and cognitive enhancement purposes [24]. The distribution of electrical current at the interface of tDCS depend significantly on the quality of skin contact materials that are used in electrodes. These materials has affect on the distribution of electrical current at the electrode-skin interface, which is crucial for achieving optimal stimulation while minimizing potential adverse effects such as skin irritation and burns [20] [50].

The motivation behind the thesis is to improve the quality and reliability of tDCS treatment by focusing primarily on foundational research for future applications. During the project, controlled electrical currents are used to influence neural circuits by addressing challenges in evaluating and improving tDCS electrodes by developing a systematic verification testing model that would offer a framework for electrode electrical property analysis, current distribution patterns, and the pinpointing of problems affecting electrode performance and user comfort [55] [26]. While the ultimate goal is to contribute to the advancement of tDCS technology, this study focuses on laying the groundwork for future electrode development by enabling systematic and repeatable testing.

In this regard, the thesis develops and evaluates a grid-style electrode specifically for verification testing. These electrodes are designed to work with a standard cathode and a segmented anode. Measurements are made with dry electrodes placed on the outer edges of each anode segment. The biggest challenge in this setup is to avoid the tDCS electrodes' hydrogel wetting the measurement dry electrodes. With an emphasis on the design of the grid electrode and how it can reveal the current distribution pattern, this work serves to further our knowledge in the electrode-skin interface and its evaluation, opening avenue for method towards designing and verification of personalized treatment systems. These will need a robust current measurement methodology to enable localized current control based on feedback loops.

Simulations using the SimNIBS software will be the initial phase of this work, which provides insights into the current distribution that will further be validated through experimental measurements. The main goal is not to find the "best" electrode design but rather to develop a robust framework in which electrode performance can be systematically studied and evaluated and potential areas for improvement recognized.

The objectives of this thesis are to make a full analysis of the current electrode materials; develop grid electrodes for verification testing; and design a measurement system. The scope will include simulations of electrode electrical properties, lab-based measurements, and in vivo measurements that validate the developed test system. Clinical trials and patient studies over the long-term are beyond the scope of this project.

The thesis begins by reviewing relevant previous studies to understand the current situation and gaps in knowledge regarding tDCS, its electrode materials and their impact on stimulation optimization as well as known side effects regarding user

discomfort from the perspective of electrodes. This is followed by the methodology employed in the thesis that involve electrode material and design selection, choice of grid electrode to perform in vivo measurements, and also implementation of SimNIBS in simulating the experiments.

Further sections cover the experimental test results of the designed grid electrodes and the measuring system, results on electrical characteristics, distribution pattern of the current, and validation against the simulated and real conditions of the study.

In the discussion section of the thesis, results are analyzed and compared to earlier studies. This will help in showing how they can improve the design of tDCS electrodes, this developed setup and give further insights into the stimulation process. The thesis concludes by summarizing key findings and giving ideas for future studies to make tDCS technology even better for therapeutic use and research in neuroscience.

These problems, once solved, will yield even more advanced and useful tDCS technology; in this way, better treatment results could be achieved with even broader usage in both medical and research fields. There is a very important need to understand the interaction of electrical currents with biological tissues to make tDCS treatments safer and more effective.

2 Theory and Background

This section explores the theoretical background and previous research for the subsequent experimental and analytical parts of the thesis. The section begins with an overview of the basics of brain electrical activity, which is very important for understanding the mechanisms of transcranial direct current stimulation (tDCS) and its effects on brain function. Then basics of tDCS treatment both from the practical and theoretical perspective are introduced and after this tDCS' skin contact materials are discussed.

2.1 Basics of brain electrical activity and properties

The electrical activity of the brain is complex, but essential to the nervous system's functioning because it makes neuronal communication through both electrical impulses and chemical signals possible. Neurons communicate via synapses, where neurotransmitters help to relay messages across networks of interconnected cells (figure 1). This activity forms complex networks. The fundamental electrical mechanism making this form of communication possible is called action potential, which is a rapid electrical signal that enables information transmission between neurons [55].

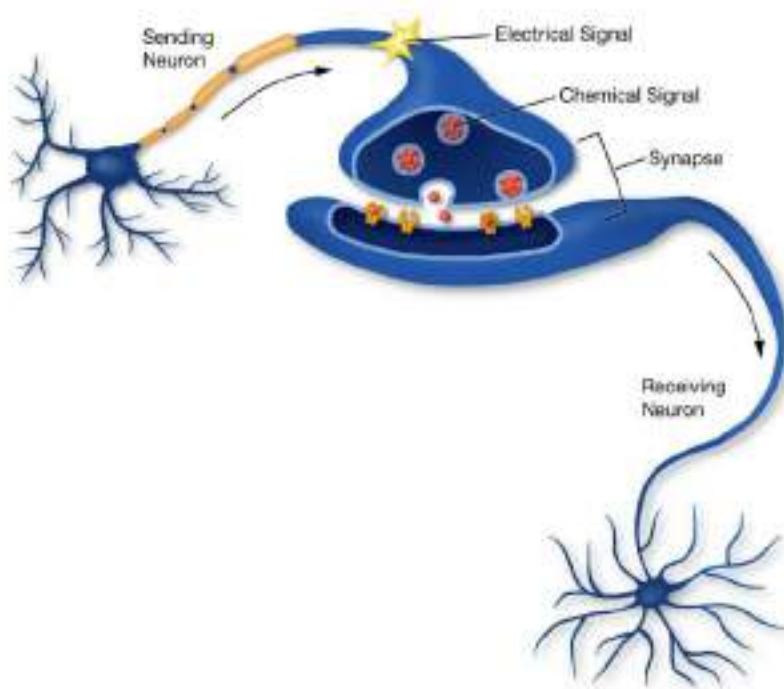


Figure 1: In neurons, signals are travelled as electrical impulses along nerve fibers and are then transmitted to other neurons via synapses using chemical messengers, called neurotransmitters, including for example dopamine, serotonin, and noradrenaline. [23]

Since neural electrical activity is important for brain to work, it is also necessary to understand how external electrical stimulation, such as tDCS, influences brain

function and neuroplasticity, meaning the brain's ability to adapt and change.

2.1.1 Neural signaling and networks

Neural signaling has two parts; electrical impulses traveling along neurons and chemical neurotransmitters like dopamine and serotonin are exchanged across synapses. Synapses are the points at which two neurons meet. Brain networks are the key in multiple functions such as information processing and regulation, enabling the transfer and processing of information between different brain regions. Examples of brain networks include sensorimotor networks which allow coordinated responses and complex cognitive processes such as movement regulation and limbic networks involved in emotional experiences and memory [55]. Understanding these networks is relevant for comprehending neuronal electrical activity and how it contributes to brain function and how it could be modified through electrical stimulation which has potential in therapeutic interventions for neurological and psychiatric disorders [6].

2.1.2 Electrical stimulation's influence on brain function

When applying electrical stimulation, such as tDCS, to the brain, influences brain function and plasticity either increasing or reducing it, depending on the stimulation's type and parameters. Neuroplasticity refers to the brain's capacity to reorganize and modify its structure and function in response for example to experiences, learning, and injury [55]. There are other electrical stimulation methods such as transcranial magnetic stimulation (TMS), and deep brain stimulation (DBS) which modulate brain activity in different ways compared to tDCS. Common factor to these methods is that electric fields that different stimulation methods form, generally become weaker in deeper cortical structures in the brain [70]. Multiple studies have shown that these neuromodulation techniques can have impact in various neurological and psychiatric conditions, like depression, pain, and hyperexcitability [6] [50].

2.2 Electrical properties of biological tissues

One needs to understand the electrical properties of biological tissues to assess how tDCS currents interact with tissue. One important example is that tissue resistance decreases with increasing frequency. This relationship is key to understanding how electrical current in tDCS (or other neuromodulation methods), interact with different tissue types and influence the stimulation effectiveness. In the figure 2 the frequency dependence of the electrical properties of biological tissue and important parameters are presented and also the key frequency regions of permittivity and conductivity are explained. This frequency dependent phenomenon is significant for assessing tissue impedance across different frequencies, which varies with tissue composition, cell structure, and interstitial space [44]. In the late 19th and early 20th centuries, Bernstein did studies that has shown that the cell interiors function as electrolytes, while the cell membrane acts as a resistive barrier.

However, tissue impedance is not only resistive but it also includes a reactive component. This has influence on how electric current spreads within tissues, which

makes it relevant especially at different frequencies, where reactive elements can significantly affect tissue impedance and the overall electrical behavior of tissues.

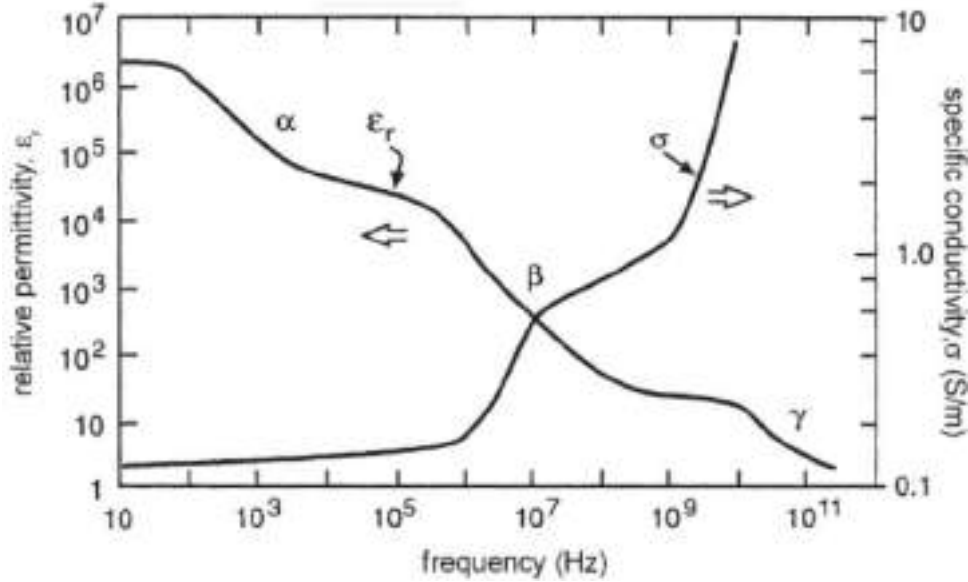


Figure 2: In the figure can be seen how the electrical properties of biological tissue change when frequency changes. It has two main parameters: relative permittivity (ϵ_r) and specific conductivity (σ), which are both shown as functions of frequency (Hz). The left y-axis represents relative permittivity, which tells how much electrical energy a material can store compared to a vacuum. The right y-axis shows specific conductivity, which explains how well the material can conduct electrical current. There are three main frequency regions in the figure. In the α -region (low frequency, below 1 kHz), the relative permittivity is very high because of ion movement and charge polarization in cell membranes. However, the specific conductivity is low since ion mobility is limited. The next is the β -region (1 kHz to 10 MHz), where permittivity drops a lot because water molecules and other polar molecules relax. At the same time, conductivity increases significantly because ions and water molecules move more freely. Finally, in the γ -region (10 MHz to 100 GHz), permittivity becomes very low due to electronic polarization effects, while conductivity keeps increasing, now mainly influenced by electron conductivity [44].

These results are very important for treatments like tDCS and transcranial alternating current stimulation (tACS), because knowing the tissue impedance at different frequencies is necessary for evaluating the efficacy and safety of these types of treatment [42].

There are useful tools for analyzing and understanding the electrical properties of biological tissues at different frequencies. One example is Cole-Cole model based equation which is the complex impedance model:

$$Z = R + \frac{(R_0 - R)}{1 + (i\omega\tau)^{1-\alpha}} \quad (1)$$

This Cole-Cole model is a mathematical approach for helping to understand

how biological tissues and electrochemical systems work. This is very helpful for developing better tDCS applications [44] [42] [40].

To understand how tissue resistance affects tDCS, it is important to look at how it changes the current distribution and how the materials used for skin contact influence this.

2.3 Transcranial direct current stimulation (tDCS)

Transcranial Direct Current Stimulation (tDCS) is a non-invasive brain treatment that uses a low-level electric current delivered through small electrodes placed on the head. It was initially made to treat brain injuries and mental health issues like depression [24] [20].

Sooma's tDCS device fulfills the definition of a medical device: "In clinical context, medical device include any instruments, materials, equipment or software intended to be used alone or in combination with other tools for the prevention, diagnosis, monitoring or treatment of human diseases" [Regulation (EU) 2017/745].

2.3.1 Basic principles of tDCS

Low-level direct current (typically 1-2mA) is delivered to the brain through electrodes on the scalp during tDCS treatment (see figures 3, 7). One treatment session of stimulation lasts typically for 10-30 minutes and can be targeted to specific brain areas depending on the placement of the electrodes [64].

In typical setup there are one pair of electrodes that is used: an anode (positive electrode) and a cathode (negative electrode). The current flows through the anode, through the scalp and skull, into the brain, and returns through the cathode, where the anode stimulates the underlying brain areas, while the cathode inhibits them [9]. This direct current influences the resting membrane potential of neurons so that the positive anode depolarizes the neuronal membranes, which makes them more likely to become active, while the negative cathode hyperpolarizes them, meaning they are less likely to activate. These effects can last for over an hour after the stimulation ends [61].



Figure 3: The figure shows a treatment kit made by Sooma Oy. Sooma Oy is a Finnish medical device company that focuses on tDCS therapy [1]. The Sooma kit includes all the equipment needed for tDCS treatment: electrodes, connecting wires, a direct current device that connects to a smartphone via Bluetooth, gel pads for the electrodes, and 0.9 % NaCl saline solution for wetting the pads before use. There is also a cap to keep the electrodes in place during the treatment session [62].

Physiologically, the effects of tDCS are based on its ability to modify synaptic plasticity with both immediate (during stimulation) and lasting (post-stimulation) effects. From the perspective of biophysics, tDCS's influence depends on the electrical current and how it interacts with biological tissues. Current flow through tissue follows Ohm's law ($I = \frac{V}{R}$), where I is current, V is voltage, and R is tissue resistance. Skin and skull resistance affect current distribution (fig. 5). The size and how long the effect lasts depends on stimulation duration and current intensity, which also affects to how big electrical field it creates (fig. 9) [52] [43]. tDCS can increase or decrease the ability of neurons to produce action potentials. Anode that depolarizes neurons, makes them more sensitive to synaptic activity, while hyperpolarizing cathode reduces this sensitivity. tDCS can influence long-term synaptic plasticity (LTP and LTD), altering the strength of synaptic connections long after stimulation, which is especially significant for learning and memory. Moreover, tDCS can affect the release and activity of neurotransmitters such as glutamate and the primary inhibitory neurotransmitter in the brain (GABA), influencing also broader brain function [29].

Electrodes also induce chemical reactions at the electrode-electrolyte interface, affecting the efficiency and safety of stimulation. This is why, electrode material and fabrication are critical factors too [21].

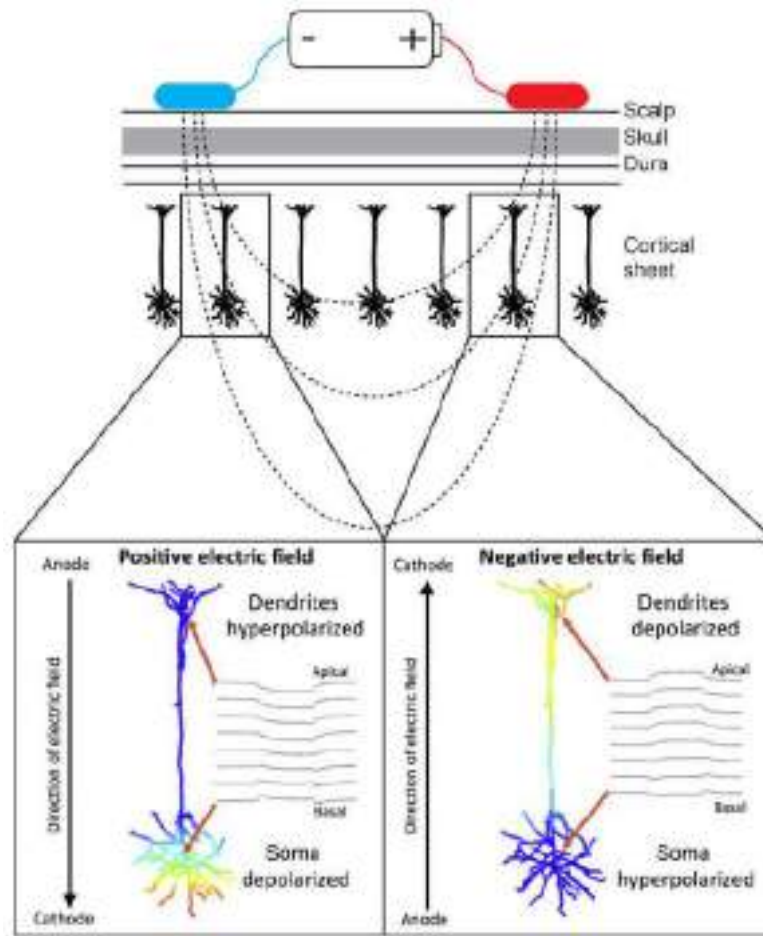


Figure 4: This figure illustrates the physiological effect of neuroplasticity of tDCS-treatment. It either increases or decreases the ability of neurons to produce action potentials depending on the electrode. Anode depolarizes and cathode hyperpolarizes neurons [57].

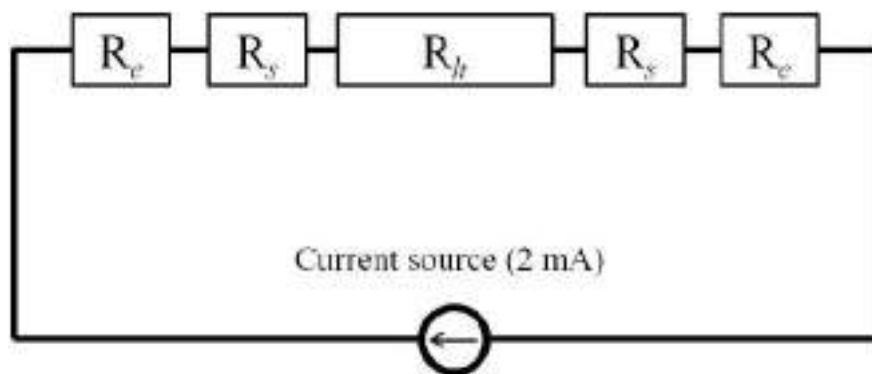


Figure 5: Circuit model of the tDCS system. R_e , R_s , and R_h are the resistance of the rubber electrode with a sponge, the skin layer (2mm thickness), and the head model excluding the skin, respectively [25].

It's important to note that the electric field is not completely uniform on the

skin surface; it can vary from one location to another. This variation is mostly caused by skin irregularities, the placement of electrodes, and their contact with the skin. The distribution of current density depends on the head's anatomy, including its size, skull thickness, cerebrospinal fluid (CSF), and the structure of the cortex. Diseases and injuries can affect brain tissue such that it changes the tissue so that it affects how the current flows. If electrodes and their placement do not take these factors in consideration, it can lead to high concentrations of current in certain brain areas or on the skin, causing burns or other negative side effects. Other considerable factors that affect current density are for example the distance between electrodes, the thickness of the sponge holding the electrolyte, and the electrolyte's concentration [61] [33]. In addition to these, one of the problems with using DC is polarization effect, which could potentially work as a negative current source near anode location. In this context, polarization means the buildup of electrical charge and voltage where the electrode touches the skin during continuous DC stimulation. This can change how good is the skin-electrode contact during the stimulation, which then affects how the current spreads at this point. If the DC stimulation continues for a long time, polarization potentials can form between the electrode and the skin. This can make the electrode less effective over time and cause side effects [8].

How much polarization happens at the electrode-skin interface depends on a few things; like how good is the skin-electrode contact, the size and shape of the electrodes, and how carefully the electric current is directed to the ROI. These details are important because, as mentioned, the success and safety of tDCS depend on how the electric field spreads and how it interacts with neurons in the brain. That's why it is so important to place the electrodes in the proper positions and try to control the polarization effect at the electrode-skin interface [52] [10].

Meanwhile, it's also necessary to look at the clinical and research areas where tDCS has shown good results as a possible treatment method.

2.3.2 Clinical and research applications

According to current knowledge, modulating brain networks with tDCS, can offer therapeutic benefits without big side effects in different clinical conditions like depression, chronic pain, and neurodegenerative disorders. There might be other advantages too, for example with learning and memory [13]. Its non-invasive nature, combined with the ability for home use, makes it a feasible alternative or supplement to traditional treatments like drugs [4] [7]. For example, tDCS has gotten interest because of its benefits especially for patients that have suffered from chronic depression and/or have been unresponsive to antidepressants. This section gives an overview of those medicine and research areas, where tDCS has been studied and shown potential.

One of the most studied applications of tDCS is in treating depression; it has shown good results. It offers several advantages, such as fewer side effects compared to traditional treatments, and it can also be used at home [62]. By focusing on specific brain areas linked to mood regulation, tDCS might help reduce symptoms with very few side effects [52] [71].

Other areas where tDCS has been studied include managing chronic pain and

treating neurodegenerative diseases like Parkinson's, stroke, and Alzheimer's. Research has shown that these patients could also benefit from tDCS, since it can reduce pain and improve quality of life in various chronic pain conditions, like fibromyalgia, neuropathy, and back pain. The effect is based on improving motor function, cognitive abilities, and daily functioning in these before mentioned neurological disorder conditions [69] [43] [21]. As mentioned, tDCS research with healthy individuals indicates also possible improvements with memory, learning, attention, and other cognitive functions [13]. It has also been investigated as a potential intervention in early stages of neurodegenerative diseases [4].

2.3.3 Mechanisms and actions of tDCS

tDCS has treapeutical effects through various neurobiological mechanisms; it influences synaptic plasticity, neuronal excitability, and brain activity [63]. This section presents these key mechanisms, while also pointing some details why does the placement and shape of the electrodes matter and how.

As mentioned earlier, tDCS affects so that it modulates synaptic plasticity. This means the synapses' ability to strenghten or weaken over time in response to activity. This change can cause a long-term potentiation (LTP) or long-term depression (LTD) of synaptic connections, which then lead to changes in neural networks and how they function. This modulation of synaptic plasticity is considered as the reason in tDCS-induced improvements in cognitive functions and motor learning.

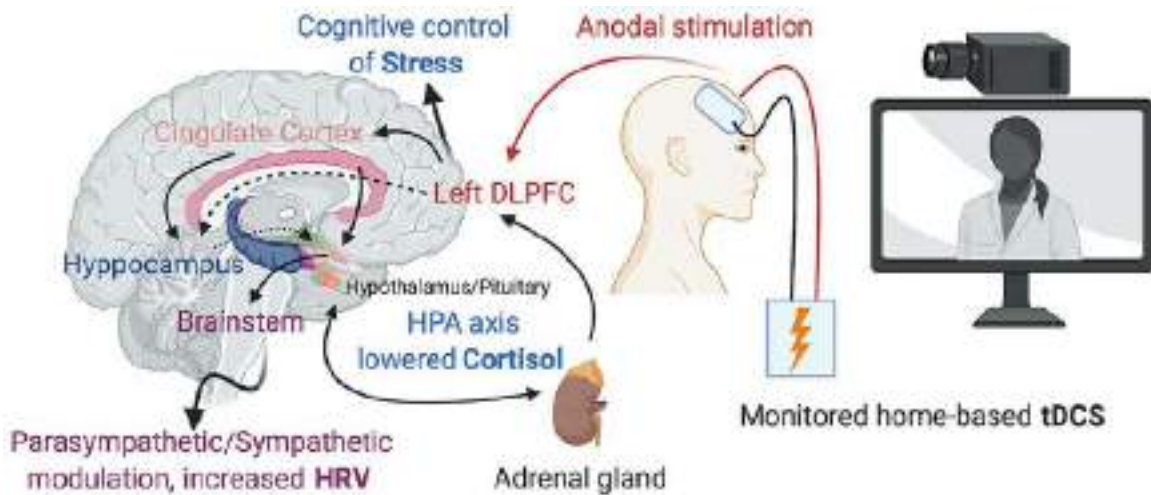


Figure 6: This figure shows how the home-based tDCS is applied to the left dorsolateral prefrontal cortex (DLPFC). This is just one of the applications of tDCS and it shows its effects on how it enhances cognitive control of stress, alters the hypothalamic-pituitary-adrenal (HPA) axis, and also modulates the parasympathetic and sympathetic nervous systems [51].

One of the effects of tDCS-stimulation is also changes in neuronal excitability. This happens because tDCS affects the resting membrane potential of neurons (explained in more detail in section 2.3.1). These changes in neuron's sensitivity can

alter how neurons communicate and affect brain network activity, which then in turn influences various cognitive and motor functions [52].

Stimulation can affect blood flow and metabolism in targeted regions, which facilitates neuroplastic changes that propagate through interconnected networks, resulting in even broader functional and behavioral effects [63] [8].

Electrode placement in tDCS therapy is a crucial aspect of its effectiveness and purpose (the intended therapeutic goal), since with different conditions there is different ROI to target on [29]. For example, in the figure 6 the illustrated home-based transcranial direct current stimulation (tDCS) targeting the left dorsolateral prefrontal cortex (DLPFC). The DLPFC is known to play a role in regulating cognitive control and managing stress which in many conditions such as depression is somehow disturbed [51]. The aim by stimulating this certain area with tDCS, is to increase the ability of cognitive control and stress management. That is why correct electrode placement can be the key to achieve effective and targeted outcomes in tDCS therapy [15] [14] [5].

Understanding these neurobiological mechanisms is important for using tDCS safely. Further research is still needed for characterizing these mechanisms and their implications for clinical and research applications of tDCS.

2.4 Skin contact materials

The materials used for skin contact in tDCS are very important for the treatment effectiveness and safety. These materials help control how the current spreads at the electrode-skin interface [52] [8] [47]. Studies show that skin moisture levels and electrode material type have big effect on contact impedance. If there is poor-quality contact with the skin and the electrode, the current can distribute unevenly, which creates hotspots. These hotspots cause one of the most common side effects of tDCS, which is discomfort that may even lead to skin irritation during tDCS [32]. Increasing the skin moisture reduces the impedance and improves the quality of the contact. At the same time, different materials have different effect on impedance, which shows why it is so important to choose correct materials for effective tDCS [26] [28].

It is very important to choose materials which have good conductivity, are safe for skin (biocompatibility), and are strong enough for long-term use. These materials must also be reliable and safe for the patient. For example, in tDCS treatment by Sooma Oy (see Fig. 7), the electrodes use a special gel pad; superporous hydrogel pads. These gel pad are first moistened with 0.9% NaCl saline solution, and then they expand to cover the whole surface of the electrode. This makes the contact between electrode and skin low and evenly distributed. This system uses electrode cups and one-time use stimulation pads (called ComfoTodes and ComfoPads) to make the stimulation safer and more comfortable [62] [32] [61].



Figure 7: Electrode used in tDCS therapy by Sooma Oy. The figure has three parts; On the left is electrode cup and a frame (ComfoTrode), in the middle is single-use stimulation pad (ComfoPad) in the electrode cup and on the right is electrode and pad after moistening with 0.9% NaCl saline solution and swelling of the hydrogel until it covers the whole electrode area. [62]

One problem at the electrode-skin interface is air gaps between the electrode and the skin. Such gaps can make tDCS less effective or increase the risk of hotspots, because they may increase resistance on some areas on the electrode and make the current flow uneven. If there are air gaps, the resistance and impedance change in different places, and the current does not spread correctly. This can make treatment less effective and more difficult to understand how current flows through tissues.

There can also be inter-individual differences. For example, hair differences like how thick they are if there are any, and what is hair density etc. in addition to how moistured the skin is, or how thick it is can all change how well the treatment works. Problems with skin contact materials can also cause side effects like skin irritation or even burns, which are likely because of uneven current distribution [26]. Some patients report discomfort or sensitivity where the electrodes are placed, often caused by stimulation of peripheral nerves or minor electrochemical reactions [47]. Because of these challenges, one key focus in tDCS is managing the current distribution at the electrode-skin interface. This means understanding how the current spreads between the electrode and the skin.

First, understanding the current distribution helps to optimize stimulation parameters like intensity and duration [46]. This is important because different people have different skin properties and electrode placements, which can change how the current flows [31]. By controlling and monitoring the current distribution, it is likely possible to ensure that the right brain area gets enough stimulation while avoiding

both over- and under-stimulation [15].

Secondly, to see the local effects of the treatment, one needs to understand how the current spreads. tDCS is often used to target specific brain areas, like for pain relief or helping to improve motor skills [67]. By studying how the current actually distributes on the electrode surface, it gives more accurate position to conclude how it changes the activity in these targeted brain areas and what kind of side effects it possibly has on the skin surface. This paves the way to make the treatment work better and more exact, and it also helps to avoid side effects.

When thinking about the safety of tDCS, it is also very important to understand how the current spreads between the electrode and the skin. Checking how the current moves on the skin and deeper into the tissues, whether there is a pattern. Between different persons, areas with higher current density could be found that may cause skin irritation or other side effects [7].

So in conclusion, investigating electrode-skin interface current distribution during tDCS treatment not only gives possibility to create more personalized treatments based on individual properties and variability but also deepens the understanding of how electrical stimulation affects brain function and how it can be best utilized in clinical practice.

2.4.1 Electrical properties of materials

Materials that make tDCS electrodes are very significant in the functionality of the device. The materials used have to have low impedance for easy flow of current to the brain. Other key factors include conductivity, resistance, and impedance also must be adjusted so that the present flow is even to avoid discomfort or skin problems.

Materials also change the way the current spreads on the electrode skin interface. This influences the way of the electrical current into different tissue layers. This is very important to make sure the brain areas that are being targeted get the right amount of stimulation, while also lowering the chance of skin issues [26].

Resistance ($R = \frac{V}{I}$) indicates a material's ability to resist the flow of electric current. Materials with lower resistance allow for better electrical conductivity. The resistance of skin contact materials is important because too high resistance can lead to uneven stimulation and discomfort or ineffective current levels.

Capacitance ($C = \frac{Q}{V}$) describes a material's ability to store electrical charge. The capacitance of tDCS electrode materials can affect the smoothness of the stimulation and its long-term effects. Good capacitance helps to smooth and stabilize fluctuations in the electrical current.

Conductivity ($\sigma = \frac{1}{\rho}$) means the ability of a material to conduct electrical current. High conductivity materials, such as doped carbon electrodes or silver/silver chloride electrodes are commonly used in tDCS since they ensure that the current flows smoothly and effectively through the skin [56].

Materials that have a different conductivity compared to each other, a bending of current density field lines, J , occurs at the interface between these two materials as shown in the figure 8. This bending of current density field lines at the interface between materials happen because of the changes in the direction and density of the

current flow due the different conductivities of the materials, which could possibly explain negative voltage values near anode location. The bending of current density field lines follows eq. (2):

$$\sigma_2 \tan \theta_1 = \sigma_1 \tan \theta_2, \quad (2)$$

describes how the current changes direction at the interface between two materials with different conductivities. Here, θ is the angle between the current direction (J) and the line perpendicular to the interface. This happens when the electric current will be passing through various tissues during the treatment of tDCS, like the skin, skull, and brain. All these present different electrical properties, because of which the behavior of the current density and distribution-something that is important in order to provide an assurance of efficacy and safety changes [39].

The conductivities of these tissues relevant to tDCS treatment are given in figure 9, and the simplified scheme of an electric circuit including these tissues and the direct current shown in figure 5.

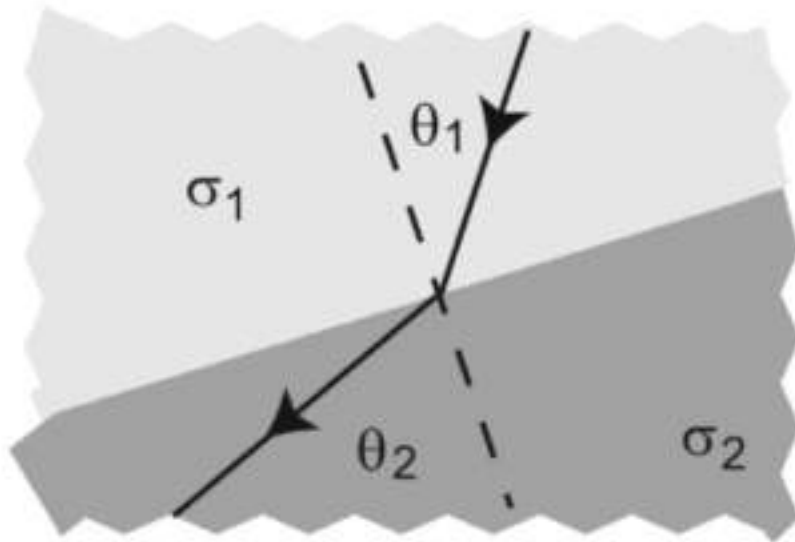


Figure 8: The figure shows how current density field lines change direction at the boundary between two materials with different conductivities (σ_1 and σ_2). The angles of the field lines relative to the perpendicular to the interface are shown as θ_1 in the material with conductivity σ_1 and θ_2 in the material with conductivity σ_2 . This same phenomenon also happens during tDCS treatment when the current passes through different types of tissues. The bending of current density field lines at the interface between materials is due to changes in the direction and density of the current flow because of the differing conductivities of the materials [39].

The electrical interfaces between skin contact materials and the skin influence both the comfort and effectiveness of tDCS stimulation. The materials should be selected to minimize the contact resistance between the skin and the electrode [39]. To improve contact, electrolytes and gels are often used [71].

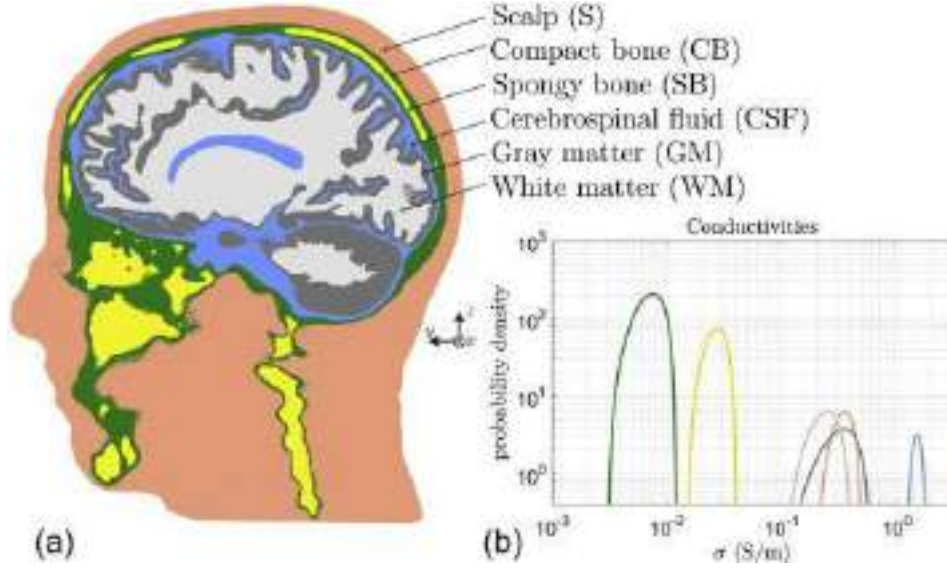


Figure 9: a) The isotropic head model showing six different tissue types (each represented by a unique color), divided into tetrahedra and vertices, with each element having an average size of about 1 mm³. (b) Symmetric beta distributions showing the probability density functions for tissue conductivities on a double logarithmic scale. The figure also shows the probability density functions for different layers of the treatment area, where one can see how these conductivities vary across different areas with some uncertainty involved. [59].

2.4.2 Electrode-skin-interface

This preparation of the tDCS treatment will be important mainly for understanding what happens at the interface of the electrode and skin. As already said, factors such as electric field and current density play a big role in safety and comfort. The theoretical background for these phenomena are explained in this subsection, together with what their role in tDCS will be.

The electric field, E , means a force produced by the electric charges in the space surrounding them. It can be framed as a function of two key variables: charge density, ρ , and permittivity, ϵ . During tDCS, this electric field is induced by the voltage between the two electrodes placed on the patient scalp. In conclusion, according to Gauss's law, the magnitude of the electric field can be related to the electric potential:

$$E = -\nabla V$$

It means that with greater voltage difference between electrodes electric field is stronger and vice versa. Depending on its direction, strength and duration of the field it can either stimulate or inhibit neural activity [48] [27]. Moreover, most critical components of electric field was identified as CSF thickness with the variance between persons estimated to be around 50%. Results show that the more CSF surrounds the cortical surface, the weaker the electric field. Moreover, from literature, with increasing age, the field strength decreases slightly [34].

Current density (J) describes the quantity of electric charge passing through the tissues. It is defined by both Ohm's law and Faraday's law:

$$J = \sigma E + \epsilon_r \epsilon_0 \frac{\partial E}{\partial t}$$

In the equation above, the first term, σE means the conducted current while the second term $\epsilon_r \epsilon_0 \frac{\partial E}{\partial t}$ means the displacement current. The displacement current varied as the variation in electric field.

Here, σ denotes the tissue conductivity. The current density gives an idea of the ease with which electric current flows in the tissues, as well as the manner in which it spreads at the interfaces of electrodes and skin. It is a significant feature in tDCS as it outlines how faithfully the stimulation crosses target areas of the brain [58]. Bioimpedance: Impedance in biological contexts; let us denote it as Z . It tells the amount of opposition the tissue conducts the flow of current. It is noted as complex impedance, where resistance and reactance. Bioimpedance investigation can give us a better understanding of how the electrode and skin interact-that is, things such as skin resistance and how the current spreads across different tissues. This is of value when considering optimal electrode placement and choosing appropriate electrical parameters for the treatment [68].

Biot-Savart's law describes how magnetic fields are produced by electric currents. It may have some relevance for the tDCS because the currents in the electrodes will produce small magnetic fields which could affect the activity of cells.

$$B(r) = \frac{\mu_0}{4\pi} \int \frac{J(r') \times (r - r')}{|r - r'|^3} dr \quad (3)$$

The magnetic field produced by the currents flowing into the skin will also reach the cells and interfere with the biological process, including ion channel activity in the cell membranes. It can affect how current is distributed in tissues. Using Biot-Savart's law, the way the magnetic field's intensity and direction are spread out in a different layer of skin can be estimated. Understanding this is important to analyze how the geometry of the electrode-skin interface and electrical stimulation affect the treatment outcomes [58].

Biot-Savart's law gives a mathematical conceptualization that helps in modeling and, therefore, understanding how the electric current within the electrode generates the magnetic field, traverses the skin and other tissues, and interjects with neurons. This would perhaps enable understanding of the current distribution, strength, and direction, coupled with its effects on the targeted area when using tDCS treatment.

In summary, the electric field and current density form the basis for the physiological effects of tDCS therapy. These equations deepen understanding of how electrical phenomena and physical laws relate to the electrode-skin interface during tDCS treatment. Each formula offers a different perspective on how electric and magnetic fields interact the surrounding environment, particularly in the interface between the skin and the electrode [58] [27].

2.4.3 Biocompatibility of materials

Biocompatibility is another important factor in tDCS electrodes, and materials should be compatible with skin and other tissues to avoid various types of problems. They also need to ensure safety for a long period [73]. The materials have to be hypoallergic and not cause any skin irritation or allergic reactions. Materials such as hydrogels, silicones and salines are normally used because of their good compatibility [2].

They should be bioinert, especially for long-term use, in that they should not cause a reaction with other tissues. This negates interference with normal tissues and ensures inflammation does not take place. Materials should not result in allergic reactions or skin irritation. For this, dermatologically tested materials are used, found to be safe under a wide range of users.

Materials should be resistant to long-term use, that is, not losing properties and not causing side effects. That is, materials are durable, do not degrade, and do not wear out the skin over time [72].

So the biocompatibility is an important requirement for tDCS materials, but in this thesis we mainly focus specifically on how the electrical current distributes on the electrode-skin interface during tDCS treatment. Our goal is to avoid hotspots, which may cause discomfort or harm and hence improve the biocompatibility.

2.4.4 Conventional and novel materials in tDCS

Traditional materials include saline-soaked sponges and conductive rubber; these widely used electrode materials have been favored in practice because of their availability and relatively good performance [2]. However, further research is being conducted to establish new materials which could provide better conductivity, higher biocompatibility, and more comfort for patients. Innovations in material science have the potential to enable further optimization of tDCS treatments so that they are more effective and accessible. Innovative new materials are under heavy research.

Most common conventional tDCS electrodes are prepared with carbon, silver, and their compounds, such as silver/silver chloride. These materials are widely used due to their good conductivity and biocompatibility [41].

Hydrogels offer good electrical contact and are comfortable on the skin. Silicone-based materials are flexible and durable, which makes them suitable for long-term use [2] [72].

Innovative materials include graphene, conductive textile and conductive polymers. Graphene is an extremely thin and strong material with excellent electrical conductivity. Its use in tDCS electrodes can improve the effectiveness and uniformity of the stimulation [54] [37]. These materials usually have silver or other conductive materials either printed into textile surfaces or as in with Sooma, the electrode conductive plate is silicone which is doped with silver [1].

3 Materials and methods

In this section, materials and methods used in measurement setup are introduced. The section describes the measurement methods, the factors that affect measurement accuracy, the equipment used for data processing as well as the validation process for this setup.

3.1 Selection and preparation of materials

Selection and preparation of materials for tDCS electrodes are crucial to ensure effective, reliable, and safe stimulation. The selected materials and preparation processes were chosen not only for their immediate performance in tDCS in vivo-testing but also to support the overarching goal of establishing a systematic and repeatable verification testing model that is intended to identify critical factors influencing electrode performance and provide a basis for future electrode development and optimization. The following subsection outlines the criteria on the choice of grid electrodes and measurement electrodes, describing the processes of preparation.

In the selection of skin contact materials for tDCS experiments, there were a number of factors to consider. Biocompatibility was also among the important factors. Materials needed to be hypoallergenic and cause minimal irritation, even if used for long periods [73]. Flexibility and durability were also critical so that the materials could adapt well to the skin's and scalp's shape and survive multiple usage cycles without breaking down. The materials had to conduct electrical current effectively through the skin without creating much resistance. Low resistance is necessary in order to provide stable distribution of current and avoid so-called "hotspots", which could concentrate currents too much in one place and cause the user to feel discomfort [14] [61].

Other particular prerequisites of the tDCS setup also had to be put into consideration. For safety reasons, all the materials had to ensure the safety of the participants and the apparatus. Materials had to be electrically and chemically compatible with both the tDCS device and skin to ensure effective and safe treatment [39].

Further to the physical and chemical properties listed above, mathematical simulations were used to model the distribution of current across various electrode configurations and select the most efficient designs for testing.

Using these as yardsticks, materials that satisfied all the safety and efficiency requirements for tDCS treatments were selected and prepared. Preparation involves several further steps aimed at optimizing their output during the tDCS trials.

3.1.1 Criteria for selecting grid electrodes

Selection criteria of grid electrodes used in measurements are given in this subsection.

In this thesis, the grid electrodes are designed in a way to achieve precise control of the current distribution across the segmented anode. Therefore, it is made by choosing materials and structure regarding their electrical, mechanical, and biocompatibility properties.

For electrical applications, the most important properties involve the range of conductivity and resistance. Materials should conduct electricity at a proper level to ensure that the tDCS device works efficiently and safely. For example, electrodes conducting electrical current through the scalp and to the brain should be made with good conductivity materials [40] [49].

This can be a very critical consideration when the materials come into direct contact with the skin. The requirements of being compatible with skin and the risk of allergies will be very critical. So, the material should be hypoallergic and biologically compatible to refrain the skin from irritation or allergic reactions. In this way, it offers comfort and safety during long-term use [73].

Other necessary characteristics include flexibility and durability: flexible enough to bend without fracture while maintaining good skin contact, yet strong enough to withstand multiple use. This is especially true for electrodes that go through repeated use cycles.

In finding the optimal grid-style electrode for measurements, mathematical simulations were done alongside the above criterion. Simulations tested electric fields produced by electrodes with various sizes, thicknesses, and segmented designs. This helped to find the best electrode size and shape for reliable and safe tDCS measurements [32]. Further, by predicting the current distribution in tissues, simulations support the choice of materials and electrode shapes to maximize therapeutic benefits while minimizing risks.

3.1.2 Criteria for selecting measurement electrodes

In addition to stimulation electrodes, also measurement electrodes were needed. Measurement electrodes were required to reliably record current distribution at the edges of the anode segments without interference from the tDCS hydrogel. To achieve this, Dätwyler's Swiss CE marked dry-comb electrodes with brush-like ends were selected. These electrodes are small and provide excellent skin contact despite hair and other surface irregularities.

Another essential criterion was that the measurement electrodes had to avoid contamination from hydrogel. Direct contact with the hydrogel could create a short circuit or compromise the accuracy of the readings. Therefore, careful positioning and insulation were necessary to maintain clean measurements.

3.1.3 Material preparation processes

Material preparation for the tDCS electrodes included a number of vital processes that ensured quality and safety. Shaping of the various materials was done through compression molding or casting. Following the shaping, each of the materials was disinfected to ensure cleanliness, preventing contaminations.

Quality was ensured by thoroughly checking all the prepared materials. That included cleaning, disinfection, and thorough quality control measures to ensure each piece was to the required standard for tDCS.

3.2 Design of measurement system

This section describes the devices, specifications, and electrode designs used in the system. The aim was to propose a scalable framework for the systematic testing and validation of electrodes. The measurement system is designed with a focus on its simplicity, reliability and repeatability. It should require minimal additional equipment and provide consistent and reliable contact between the electrodes and the skin [3]. The setup should be as close to normal tDCS treatment protocol as possible so that the accuracy of the results are comparable to real usage of the treatment. The setup also addressed participant comfort and safety-essential components of practical usability and ethical considerations in future studies involving electrodes.

The design of the measurement setup of the tDCS electrode contact quality has to consider all the factors carefully, especially in terms of current distribution through the tDCS-skin contact material [56] [71]. For this purpose, measurement techniques should be employed which can measure and analyze the spatial distribution of the current with good fidelity [14].

Together with measurement techniques, calibration and validation procedures are significant for the accuracy and reliability of the measurement setup. The measurement protocol should also be repeatable [15]. The calibration and validation procedures will be described in the following sections.

3.2.1 Selection and justification of measurement devices

When developing measurement methodology, the selection of suitable passive measurement techniques is included. As it was highlighted, some of the main requirements of the measurement devices are precision and reliability but also real-time data acquisition and separation of AC and DC signals is needed [65]. This subsection justifies why these selected devices and components are relevant and needed in this measurement setup.

Passive methods have some advantages in simplicity, low cost, and lower complications in the setup and operation. It also reduces the level of noise [3]. However, they might also bring some limitations in sensitivity and resolution, especially when trying to observe small changes in current distribution. Therefore, in most cases a compromise between simplicity and measurement accuracy needs to be found with great care to establish valid and useful results.

Moreover, the measurement frequencies should also be chosen. The use of different frequencies during measurement can make this setup more complete with respect to the quality evaluation of electrode contact in order to make the approach more effective for the performance assessment of tDCS.

The measurement process could also be performed in a different way: using known alternating voltage and current measurements, it could calculate impedance-a comprehensive assessment of the contact quality at different frequencies and decrease the polarization effect on the skin. Because the current distribution changes with frequency, using a range of frequencies provides a fuller assessment of electrode contact quality. Above, the alternating current setup allows for the determination of how current flows through various layers and depths of skin. This is crucial because

it assures homogeneity in the current distribution inside the tDCS [36] [39]. Besides, calibration and validation ensure the measuring system offers correct and reliable measurements under any circumstances.

That is, calibration establishes a standard reference from which measurements are compared with, while validation ensures results consistent and repeatable over changing conditions. The measurement setup can reach confidence in estimate accuracy and, by that virtue, support tDCS treatment outcomes with more credibility with a calibration and validation protocol [3] [28]. Thus, the developed measurement setup shall provide a straightforward, effective, and reliable method of assessing the skin contact quality with gel electrodes in tDCS applications, enabling the recording of valid and repeatable measurements to allow for effective and safe treatment with tDCS [28]. The devices selected to constitute the setup shall measure the low-amplitude electrical signals with high accuracy and in real-time, with data acquisition and analysis while filtering environmental noise in order immediately to observe and perform adjustments.

Since tDCS treatment covers both DC and AC, the measurement devices should be able to differentiate between them. The devices should also reduce electromagnetic interference and other noise sources affecting the signal [39]. That is why devices with advanced filtering options were in focus, as they help to capture high-quality signals and prevent noise from interfering in measurement. Also, real-time data collection and the possibility to handle high sampling rates were important factors because such a feature enables immediate feedback and adjustment when measuring and analyzing current distribution across the electrode sections. Devices selected for this system include the Digitimer D440 Amplifier, which provides accurate amplification and filtering options. It is not designed to measure DC signals, but that possibility is also offered. The ADC selected for data capture was the Micro 1401. The D440 was selected because it is designed for electrophysiological studies and incorporates adjustable filters that support clear separation of AC and DC signals, essential for this study. The Micro 1401 ADC allows for real-time data acquisition and storage; hence, it interfaces directly with software to immediately display and record data. The combination of equipment provides for the accuracy and stability that can give consistency and reliability to measurement across sessions [17] [12].

The measurement system consists of a Digitimer D440 Amplifier/Filter device that has superior high-cut and low-cut filtering options.

These filtering options will help in capturing the low amplitude signals with the best quality, which should minimize environmental noise and movement artefacts, adding to the accuracy of data.

Designed specifically for electrophysiological measurements, the device tackles signals in both microampere-range AC and milliamperere-range DC components. Filtering and amplification in hardware are important steps, since they enhance the signal-to-noise ratio prior to digitization and, hence, diminish the need for extensive post-processing together with increasing the quality of measurement. By filtering, the D440 will be efficient in reducing motion artifacts as well as environmental electromagnetic noise, especially in measurements with significant interference [17]. The D175 Electrode Impedance Meter was also chosen for being able to check

electrode impedance prior to measurements [16]. The stimulation can be weakened and the measurement accuracy can be affected if electrode impedance is high. [39] [71] This equipment ensures the quality of electrode contact both before and after the measurement, aiding in ascertaining whether all electrodes are correctly attached and whether the results for the measurement are reliable. These are selected in such a way that they can give correct and reliable measurement results, handle AC as well as DC signals, and reduce the noise of several types.

This suite of equipment ensures both accuracy and stability to form the foundation for consistent, reliable measurement across sessions; Matlab provides powerful tools for post-processing and analysis [65].

The devices used within the measurement setup and in the overall system are further explained within section 3.3.1. As a whole, these aforementioned devices ensure that the results of measurement gained during the tDCS treatment are as reliable and reproducible as possible.

3.2.2 System requirements and specifications

The measurement system needed to meet a few particular technical requirements: it needed to measure the current distribution across the segments of the anode with high accuracy, variable settings of the frequency were allowed. In addition, high resolution and sufficient signal-to-noise ratio were needed for the dependability of the measurement [28]. On the other hand, user-friendliness encourages the use of the measurement system and makes its operation fast and effortless [26].

Moreover, the system had to be calibrated and validated for its accuracy across different sets of sessions. The validation confirms the reproducibility of the results in different conditions.

Technical specification of the measurement system: the two most important features are accuracy and resolution. Accuracy is responsible for the capability of the system to observe minute changes in the signal and present valid results in various measurement conditions. On the other hand, resolution affects the system's ability to make a distinction and analyze details of the signal, which is very important in skin contact for detecting minor changes. Therefore, an optimal balance of these characteristics is of great importance when designing the measurement system to obtain its best performance and user experience [15] [14] [5].

Consequently, this minimizes the level of disturbances and simplifies the analysis of data. Therefore, it gives the rationale for passive measurement methods being attractive for assessing tDCS electrode skin contact [3] [28]. In a nutshell, to ensure that the measurement system works effectively and is reliable, finding an appropriate balance between requirements and technical specifications is vital.

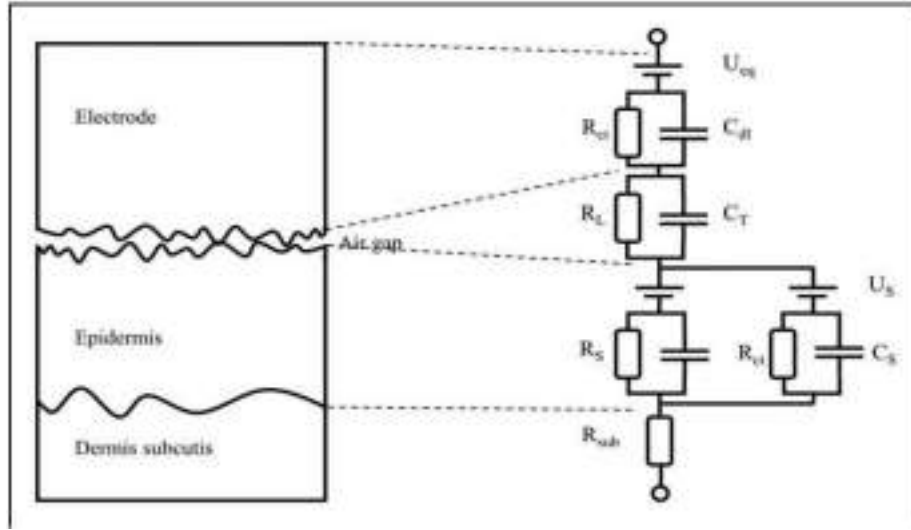


Figure 10: Schematic of a simplified theoretical circuit model corresponding to the measurement setup including multiple interfaces. This model represents the concept of the three-lead in vivo measurement method to determine the skin-electrode impedance of textile electrodes, underlining the complexity of the interfaces that are involved in the measurement process. Adapted from three-lead in vivo measurement method for determining the skin-electrode impedance of textile electrodes [28].

Electrode size and stimulation parameters should also take into consideration the threshold for safe current densities that could cause skin or neural tissue damage. For example, burns can happen at current densities around 1.3 mA/cm^2 [35], whereas brain tissue damage might be possible at more than 14.29 mA/cm^2 [38] [61].

3.2.3 Sensors and electrodes

This section introduces sensors, electrodes and cables that are used for the in vivo measurements and goes through their basic properties.

The Sooma ComfoCap (fig. 3) is used in this thesis' in vivo measurements for precise electrode placement in tDCS session. These caps are designed to achieve the accuracy and reproducibility of electrode position, ensuring the maximum therapeutic efficacy of each stimulation session. The cap is made of hydrophobic fabric, thus preventing electrode bridging, thereby allowing for consistent targeting of stimulation areas throughout the duration of each session.

In addition to ComfoCaps, ComfoTrobe (with a surface area of 25 cm^2 cathode, fig. 7) is utilized as high-quality stimulation electrode in these measurements. For the anode electrode we use a traditional circular design, divided into four insulated sections. Each anode block receives current independently from the same Sooma tDCS device, allowing controlled current flow through each section. Measurement electrodes located at the outer edge of each anode segment assist in analyzing the inter-block current distribution and detect the unwanted "leakage" of current outside the boundary of the anode. The design maintains a focused, stable stimulation area without excess spread of current by the electrolyte. The electrodes are designed such that accurate

stimulation at the site should be delivered without dispersing the current due to excess electrolyte and maintain focus on the stimulation site. ComfoTodes are used along with ComfoPads, which expand upon contact with electrolyte to create a uniform hydrogel stimulation surface.

To measure current distribution, we use Dätwyler Swiss CE-marked dry comb electrodes at four points along the outer edge of each anode segment, set in their own isolated areas attached in the ComfoTode as shown in figure 14. These dry electrodes, with their brush-like ends, ensure good contact with the skin despite the presence of hair. An important challenge in this setup is ensuring that the hydrogel from the tDCS electrodes does not wet the dry measurement electrodes. These dry electrodes avoid interference from the hydrogel’s moisture, providing clear data on current flow between segments without risking short circuits or invalid data.

The electrodes are connected with the Sooma ComfoLeads, touch-protected and color-coded 1.5 mm cables, to the tDCS device. These cables also connect the reference electrode to the measurement system for secure and consistent connections throughout each session.

Electrolyte solution intended for use with the ComfoPads: a sterile 0.9% sodium chloride (NaCl) solution in medical grade, manufactured by B. Braun, maintains constant ion balance on the electrode interface essential for consistent stimulation delivery, and enables good electrode contact during tDCS sessions [1].

3.3 Measurement system

In this section, measurement system used in this thesis is introduced. It explains measurement methods and devices, factors affecting the measurement accuracy and data processing steps. The setup was implemented with the focus on replicating real-world tDCS conditions while addressing challenges such as hydrogel contamination. It was tailored to meet the verification testing model requirements. The dry measurement electrodes were set in positions that would allow minimum interference from hydrogel, enabling clear data gathering of current distribution. Further, this arrangement supports the intent of the thesis to build a strong framework for electrode performance analysis under controlled and repeatable conditions.

3.3.1 Measurement methods

In this thesis, specific measurement methods and equipment were used to analyze current distribution during tDCS in real-world conditions which are introduced in this section. The methods utilized in this work were chosen to validate the developed grid electrodes and measurement system as a repeatable verification testing framework. Emphasizing the current behavior at the electrode-to-skin interface, the aim of this thesis is to bring forth factors affecting current distribution for the design of future tDCS electrodes and limiting user discomfort.

All measurements were performed in vivo with healthy volunteers to replicate typical tDCS real-world conditions (Section 3.6). Each subject underwent two 15-minute tDCS on the left DLPFC, with monitoring of current distribution and skin-

electrode impedance for the entire course of the session. The goal is to measure and analyze, for the first time, how current behaves and distributes at the skinelectrode interface in the course of tDCS treatment, and to study relevant individual differences of head geometry, skin thickness, or hair density influencing the contact resistance between skin and electrode.

The protocol uses the Sooma tDCS device [62], which gives a DC of 2mA, using Ag-based electrodes, with diameter of 6 cm, with NaCl saline-soaked gel pads (figures 7 and 3). Prior the treatment, the skin is cleaned with an alcohol wipe to reduce dirt from the electrode area. The tDCS device delivers a 2 mA DC, replicating standard treatment conditions to ensure that the findings of this study are comparable to real treatment scenarios.

Apart from the tDCS electrodes, dry electrodes are placed at the outer edges of the anode segments to record current distribution. See section 3.4 for details. By doing this, the interference of hydrogel-which might compromise the accuracy of the measurements-was avoided. The setup was supposed to determine the current distribution across multiple points at the electrode interface, identifying potential hotspots and poor skin contact areas. High-frequency, low-amplitude AC is also applied from the tDCS device for impedance measurement without causing any harm or giving significant disruption to the tDCS treatment. AC current, which has been shown to be effective in various studies [36] [19] [66], helps in understanding the electrical properties of the interface. For instance, EIS with AC at various frequencies has been demonstrated to determine the quality of, and changes over time in, skin electrode contact [36].

The AC frequency used in the measurement arrangement is about 1 kHz, which is sufficiently high not to interfere with DC stimulation, yet low enough not to induce significant capacitive effects in the skin. The AC amplitude is kept at a low value of about 10-100 μ A so that there is no significant heating or discomfort to the participant's skin. Differentiation of the signals at analysis would require AC/DC filters. The reference electrode is placed on the shoulder of the participant for purposes of measurement. Measurement data is collected on a computer through Micro 1401 port, which handles voltage and current measurement.

3.3.2 Factors affecting measurement accuracy

In this section the factors that can affect measurement accuracy are discussed. There are several factors that can impact the accuracy of measurements in this type of setup. The design of the measurement electrodes was designed to ensure that they did not touch the hydrogel, trying to prevent short circuits. However, it was difficult to keep the electrodes connected to the skin with the same intensity and when there was hair.

Natural variation of skin resistance is very high between subjects and will have a big influence on the accuracy of the measurement (see figure 11). The hydration level of skin, thickness, oils, and dirt may cause variations in the resistance caused by the contact of skin and electrode [5]. One of the problems with these inter-individual properties is that the typical signal processing methods cannot reliably remove

physiological artifacts [22]. Also, electrode quality, size, shape, and placement are all critical to optimization [18]. Large electrodes spread out the current, whereas hotspots are generated using small electrodes. Even the shape of the electrode affects current distribution; rounded electrodes tend to disperse current on a more uniform basis compared to electrodes with irregular shapes [15] [14].

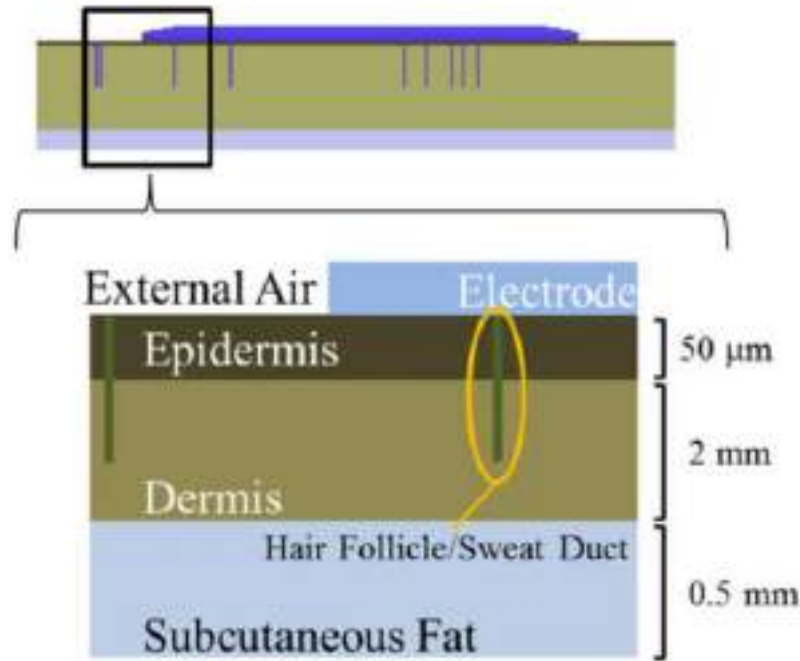


Figure 11: This figure illustrates the different layers and their mean thicknesses of the skin. These numbers varies inter individually and may thus cause current distribution differences between individuals [25]. Also the density of hair follicles and sweat ducts cause differences among population.

The quality and type of contact gel between the electrode and skin can affect the impedance measurement. High-quality gels are needed that remain steadily conductive and do not dry out during the measurement period for reliable results [14]. Also variations in skin condition, such as dryness, damage, or the presence of hair, can alter the electrical properties at the electrode interface. Such effects are minimized by a proper preparation of the skin, e.g., cleaning with alcohol wipes.

To guarantee the repeatability and stability of the electrical properties of electrodes and skin electrode interface for the whole measurement period, one has to maintain consistent pressure and contact quality during the same period of time [5] [15].

On the other hand, there are a few environmental aspects that may affect the accuracy of the measurements, like temperature, humidity, and movement. during the treatment with tDCS, also movement artifacts are generated, which are important to remove through filtering and common mode rejection. Both of them are sources of interference in DC, which becomes non-idealized into different frequencies due to movement and the induction of other interferences [22]. Controlling the measurement

environment as much as possible helps to minimize these influences.

Taking these factors into account and controlling for them can greatly enhance the accuracy and reliability of the measurements, thereby determining current distribution and impedance characteristics with greater precision at the skin-electrode interface in tDCS treatments. The AC measurement approach minimized polarization effects that may happen under DC stimulation. This gave a more stable assessment of the skin-electrode interface.

3.3.3 Data processing

Data collected during measurements were analyzed in real time with the use of **Spike2** and **MATLAB** software for post-analyzing and visualisations. These tools allowed for signal processing, reducing noise, and visualization of the current distribution. Real-time processing allowed for on-the-spot adjustments during the sessions, which gave the guarantee that the data were both accurate and comprehensive.

Filtering occurs during measurement of the raw data. The selected devices D440 and Micro 1401 have these properties, along with the Spike2 software. This eliminates almost all of the noise and artifacts introduced at this stage of measurement.

Data normalization is performed to accommodate variations across participants and measurement sessions. Such standardization provides an opportunity for data comparison between different datasets while reducing participant-specific anomalies.

After that, an analysis of the data was performed; it is explained in more detail in section 3.5. In such a way, this organized approach to the treatment of data makes sure that the measurements, carried out by using the device Micro 1401, are precise, reliable, and useful for the purposes of application.

The following sections will present an overview of the measurement results and compare them to the theoretical prediction.

3.4 Validation of measurement methods

The measurement system was validated by simulations, laboratory tests, and in vivo experiments. Such multi-step validation ensures reliability of the system to be used in this verification study. Operating environment considerations include electrode durability, flexibility, and performance maintenance over repeated measurement sessions.

The validation phase focused on testing the repeatability and reliability of the developed verification test model. In this context, the thesis tried to simulate data versus real-world measurements to ensure the model would reliably identify inconsistencies in electrode performance and guide future improvements. The optimization of the current grid electrode design was not the focus; rather, the aim was to devise a robust and scalable framework for ongoing electrode research.

Simulations on current distribution with different electrode designs were done. Scenarios were performed by the SimNIBS software. These identified that the chosen configuration minimizes hotspots while providing uniform current flow.

Laboratory tests in controlled environments measured impedance and studied current flow without biological variables. These tests validated the accuracy of the measuring system under ideal conditions.

Finally, **in-vivo tests** were conducted on healthy volunteers to replicate real-world conditions. These tests showed that it was possible for the system to overcome the challenges posed by tDCS treatment, such as hydrogel interference and head geometry variability.

The results of the validation proved that the system met the thesis requirements. for accuracy, reliability, and repeatability-are met, thus being suitable for current distribution analysis during tDCS.

3.4.1 Simulation test

This subsection introduces the tools used for simulation tests. The open source SimNIBS software will be used in all tDCS simulations for the implementation of a simulation of the action of electrical current on the brain. The software uses FEM in order to model tissue distribution of electrical current [60]. These were simulated using pre-set configurations of electrodes, which were modified according to parameters like electrode size, spacing, and gel layer thickness. This was done to optimize the placement of electrodes to produce a homogenous distribution of current.

This software is used to create 3D anatomy from the features of a patient's head and brain, taken from MRI images. These MRI images are then converted into a triangular mesh compatible with FEM simulations. The whole process involves the following steps: first, the creation of the head model from MRI data; second, modeling electrodes; third, the representation of anisotropic white matter conductivity; fourth, numerical simulations; and lastly, visualization of results. Through semi-automated segmentation, various tissues such as skin, skull, air pockets, cerebrospinal fluid, white matter, and gray matter are separated from T1-weighted MR images. The conductivity values for each element in sub-compartments of the mesh are calculated using the following formula:

$$\sigma_{Tissue} = \sigma_{Tissue} \cdot \begin{bmatrix} 1 & 0 & 0 \\ 0 & 1 & 0 \\ 0 & 0 & 1 \end{bmatrix} \quad (4)$$

Incorporating the anisotropic conductivity information of white matter involves a volume-constraint method, assuming fixed anisotropy ratios between principal and auxiliary directions. The calculation of eigenvalues is reliant on the white matter conductivity value σ_{WM} , an anisotropy ratio, and must meet the following conditions:

$$1)\sigma_{WM}^2 = \sigma_{main} \cdot \sigma_{aux}, 2)\sigma_{aux} = \frac{\sigma_{main}}{10} \quad (5)$$

The electrode models can be of any shape and number. Meshing of the head model is done to maintain electrode features and is not restricted by the topology of the head's structures [30].

These include several positionings of grid electrodes, mesh densities, and studying their effects on the distribution of electrical current (example in fig. 12). Also, different electrode thicknesses are being tried with a view to assessing their impact on contact resistance and current distribution.

Furthermore, SimNIBS software is also employed to simulate different current intensities, for example, 1 mA or 2 mA, and to explore their various effects on the distribution of currents. Moreover, electrodes are divided into different segments to analyze current distribution in different regions.

The simulation test is aimed at optimally choosing the grid electrodes for a measurement setup in order to cover as large an area of the tDCS electrode surface as possible. In this way, in vivo, a precise and reliable measurement of current distribution within the electrode area can be achieved. However, the aim was to maintain the measurement setup as simple as possible because an excess of active components may disrupt the measurement.

SimNIBS does have limitations, including not being able to present current distribution on the skin directly. However, the simulation does consider skin, skull, and other tissues' resistances, resulting in a rather realistic current reaching the brain. Thus, it is an excellent source of information regarding electrode selection, even though no direct information on current distribution on the skin is obtained. The goal of this simulation is to determine the optimum density of measurement points for grid electrodes, the number of points needed, and their optimum placing for optimum results. Besides, the thickness of grid electrode measurement surface is assessed for effective and precise current distribution analysis.

The image 12 shows the results of four different simulations of electrodes arranged in various ways. All electrodes are defined with the same thickness and material, and they all include a gel pad. The left color map depicts the magnitude of the electric field in different areas, while the right one indicates the magnitude of the direct current applied to the electrode. It is observed that increasing the direct current and segmenting the electrodes increases the magnitude of the electric field. In segmented electrodes, an effort has been made to maintain a total current of 2mA by dividing it by the number of electrodes. Moving the cathode and segmenting the electrodes affect the location of the highest electric field. Maximum values for the electric fields are (a) 0,439mA, (b) 1,38mA, (c) 0,496mA and (d) 0,468mA.

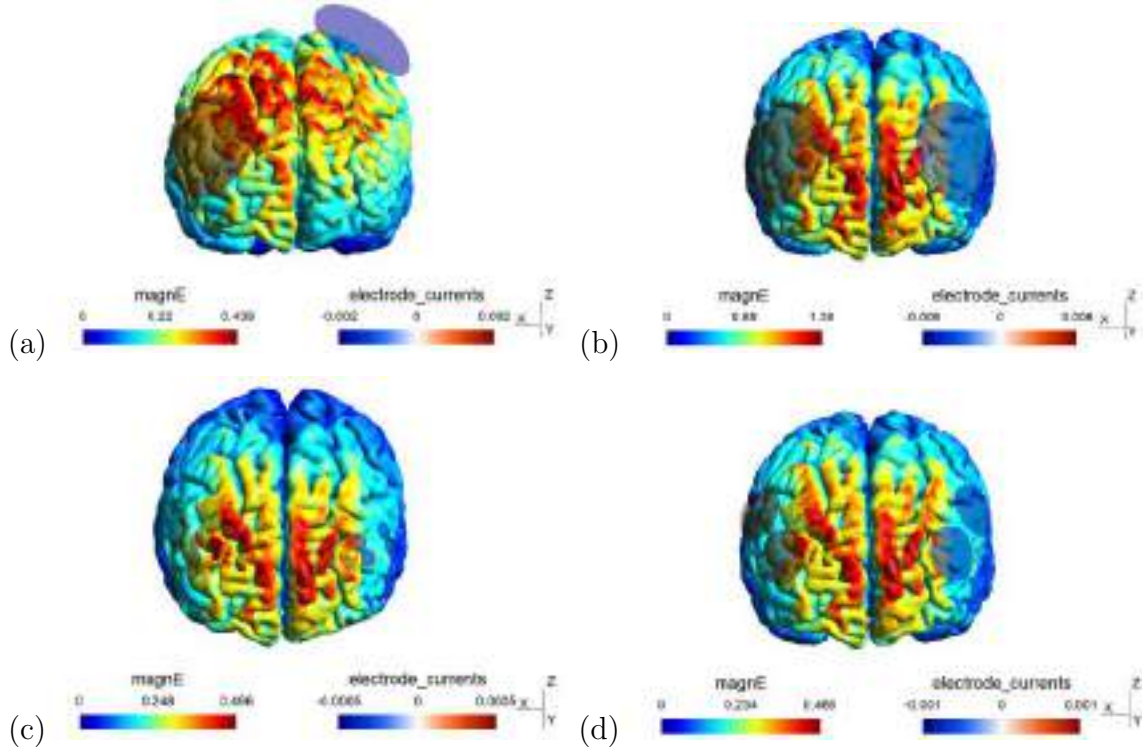


Figure 12: Electric field ($\frac{V}{m}$) distributions of the simulation results with different electrode configurations and current values (mA). (a) $\pm 2mA$ current, electrode diameter 6cm, cathode placed in the parietal lobe. (b) Current of $\pm 6mA$, electrode diameter 6cm, electrodes positioned in DLPFC and DRPFC. (c) The electrodes are divided into four parts, $\pm 0,5mA$ current to each (total $\pm 2mA$), electrode diameter 1cm, electrodes located approximately positioned in DLPFC and DRPFC. (d) The electrodes are divided into two parts, $\pm 1mA$ current to each (total $\pm 2mA$), electrode diameter 3cm, electrodes located approximately positioned in DLPFC and DRPFC.

Electric field and current density distributions were also analyzed during simulation tests using a grid-electrode setup with different numbers of blocks and also with the normal tDCS-measurement conditions (see table 1). The aim was to find out which electrode configuration best suited for in vivo measurement purposes, putting emphasis on distribution uniformity and peak values. Example results are given in tables 1 and 3 and the current distribution in the brain for different grid electrode block combinations is seen in figure 13.

This figure compares the effects of different numbers of blocks in grid-electrode (four, five, and seven) on the distribution of electromagnetic intensity (E_{magn}) in brain layers using the SimNIBS software. Images (a), (b), and (c) show the intensity distributions when four, five, and seven block electrodes are used in the grid, respectively. A greater number of blocks appears to maintain higher intensity deeper in the brain layers. This suggests that increasing the number of electrodes can achieve more effective stimulation in deeper brain tissue.

The percentile analysis of the simulation results focuses on field uniformity and peak values across different electrode configurations. This is important in selecting

an optimal grid-electrode setup that offers uniform electric field and current density distribution, allowing one to develop a configuration with small peak values in order to enhance measurement accuracy and reliability. Simulation results in tabulated form are included in detail in section C.

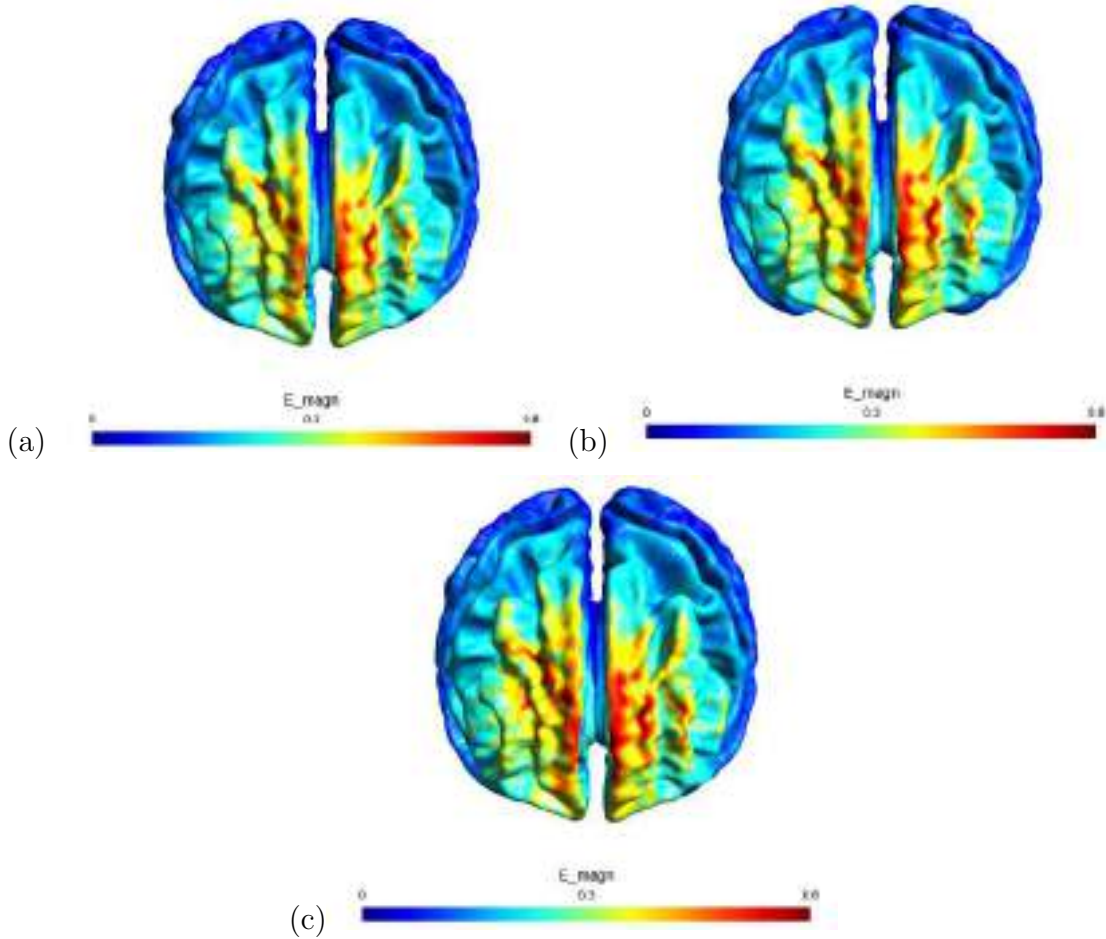


Figure 13: Effects to magnitude of electric field ($\frac{V}{m}$) with different number of segments in a grid-electrode (a) Four blocks in the grid. (b) Five blocks in the grid. (c) Seven blocks in the grid.

Table 1 gives, for some grid configurations, sample percentile values (99,9%, 99,0%, 95,0%) illustrating the homogeneity both for field strength and current density. The dependence of field strength and current density on the grid configuration at different percentiles is further detailed in tables 2 and 3, respectively.

The 99.9% percentile of the electric field and maximum current density corresponds to very high values that only a very small percentage of the measurement points achieve; this percentile will be important in determining 'hot spots' of the field intensity and current flow. The 99.0% percentile of the field value represents the threshold value below which 99% of the measurement points lie, therefore offering a good insight into the general pattern of distribution. Similarly, the 95,0% percentile gives the threshold below which 95% of measurement points lie, showing in a wider

sense the field distribution over the measurement area.

The percentile values are used because they provide an easy means of evaluating uniformity and intensity of both electric fields and current densities throughout the grid-electrode setup. Higher percentiles indicate regions that could be potential points for localized exposure, while lower percentiles indicate a more even distribution suitable for consistent measurement conditions.

Table 1: Top percentiles of the field with the normal tDCS-treatment setup.

Field	99,9%	99,0%	95,0%
E	0,446 V/m	0,352 V/m	0,275 V/m
magnJ	0,123 A/m ²	0,0969 A/m ²	0,0755 A/m ²

The optimal grid-electrode to use for measurement purposes would be determined by several factors, all related to its ability to measure and distribute accurately electric fields and current densities in the area of interest.

Uniformity of the electric field and current density is important. The simulation results can help in determining how well these parameters can be distributed among various electrode configurations. Smaller maximum values and closer percentile values (e.g., 99,0% and 95,0%) give more uniformity.

Smoothing of peak values is necessary. It is very important to reduce maximum values of electric field and current density, especially for higher percentiles - for example, 99.9% - in order to avoid risk in high-intensity areas.

In such a way, it is possible to compare and choose the best electrode configuration providing more uniform distribution of electric field and current density by using simulation results with minimum peak values.

In conclusion, choosing an electrode that offers uniform distribution of electric field and current density, minimizes peak values, and ensures the accuracy of the measurement is necessary. Simulation-driven decisions optimize electrode configurations for specific measurement requirements and also work as robust method to initiate design and compare different electrode designs in personalized treatment applications which require high-accuracy current distribution measurement and control.

Table 2: Field percentiles for different grid configurations; grids with 4, 5 and 7 blocks when the block are divided in circle shape ($d = 3cm$) area.

Field	99,9%			99,0%			95,0%		
	grid of 7	grid of 5	grid of 4	grid of 7	grid of 5	grid of 4	grid of 7	grid of 5	grid of 4
E (V/m)	0,581	0,570	0,537	0,455	0,446	0,421	0,341	0,334	0,315
magnJ (A/m ²)	0,160	0,157	0,148	0,125	0,123	0,116	0,0937	0,0918	0,0865

Field focality analysis (Tables 4 and 5) provides information on the electric field distribution extent in the brain with a focus on volume coverage and intensity concentration. Smaller focality describes more concentrated field strength, which is desirable for focalsimulation.

Table 3: Field percentiles for different grid configurations; grids with 4, 5 and 7 blocks when the blocks are divided in circle shape ($d = 6\text{cm}$) area.

Field	99,9%			99,0%			95,0%		
	grid of 7	grid of 5	grid of 4	grid of 7	grid of 5	grid of 4	grid of 7	grid of 5	grid of 4
E (V/m)	0,422	0,430	0,445	0,330	0,337	0,350	0,251	0,257	0,264
magnJ (A/m ²)	0,116	0,118	0,122	0,0907	0,0927	0,0962	0,0691	0,0706	0,0727

Considering the results of the various simulations with the alternative number of blocks in the grid-electrode and how these are divided, as shown by tables 2 and 3 we can identify that Electric field (E) values at 99,9% decrease with fewer blocks when they are divided in smaller area (7: $5,81 \cdot 10^{-1}$ V/m, 5: $5,70 \cdot 10^{-1}$ V/m, 4: $5,37 \cdot 10^{-1}$ V/m), which shows that higher maximum values are reached with more blocks. Trends for 99,0% and 95,0% are similar, which shows that fewer blocks result in more uniform current distribution. This 4-block configuration shows lower values for these percentiles, which is indicative of a more uniform electric field and reduced localized intensity, which should improve reliability in measurement and precision.

Values for current density (magnJ) at 99,9%, 99,0%, and 95,0% exhibit a trend where fewer blocks lead to lower maximum values. The lowest current density values are observed with the 4-block configuration, indicating a more uniform current distribution. The exact opposite happens when we move the electrode blocks to a larger area.

With these results, the 4-block configuration produces the most uniform distribution of electric field and current density within a circular area of diameter 3 cm. This may provide an optimal configuration from the point of view of both measurement and treatment in cases where current must be uniformly distributed. A 5-block configuration may be considered for the purposes of precision around specific measurement points, given that the use of 7 blocks does not bring great improvements over the configuration with fewer blocks.

Field focality (see tables 4 and 5) describes how extensively a strong electric field-or field strength-spreads. In this context, focality is defined as 'the volume-or area-that exceeds a certain percentage of the 99,9th percentile of the field.' For instance, 75,0% could indicate that volume of the mesh at which the field strength is at least 75% of the value at the 99,9th percentile. This will be important because focality helps in understanding how concentrated or dispersed the electric field is.

Larger focality means that the field is widely distributed in a larger volume, though with slightly weaker strength across certain areas of the region concerned, while smaller focality refers to the situation when the field is more concentrated with greater strength in a smaller volume, which may have some advantage if the aim is to concentrate the treatment on a specific region. Consequently, by selecting the proper focality for a segmentation, efficiency and precision of the tDCS treatment regarding the wanted area can be affected.

Previous studies have sought to investigate the role of both magnitude and direction of the electric field in specific cortical structures by assessing the magnitude

Table 4: Field focality for different grid configurations with electrode setup in a 6 cm diameter circular area.

Field	75,0% (mm ³)	50,0% (mm ³)
4 blocks	$1,04 \cdot 10^4$	$6,82 \cdot 10^4$
5 blocks	$9,78 \cdot 10^3$	$6,79 \cdot 10^4$
7 blocks	$9,66 \cdot 10^3$	$6,81 \cdot 10^4$

Table 5: Field focality for different grid configurations with electrode setup in a 3 cm diameter circular area.

Field	75.0% (mm ³)	50.0% (mm ³)
4 blocks	$9,59 \cdot 10^3$	$6,24 \cdot 10^4$
5 blocks	$9,58 \cdot 10^3$	$6,25 \cdot 10^4$
7 blocks	$9,62 \cdot 10^3$	$6,25 \cdot 10^4$

and distribution of both tangential and normal electrical field components across different cortical areas [11]. They have observed a general tendency, in which tangential electric fields are focussed in gyrus areas and normal electric fields in sulcus areas for almost any electrode configuration, as can be shown by the simulation results presented in this thesis [18].

Analyzing the influence of various electrode configurations on the distribution of electric fields in the brain is indeed very well done using a 5-measurement-point grid electrode configuration. From tables C2 and C3, the different configurations are quite disproportionate to each other, reflecting large differences in electric field strength between different measurement points and percentiles, for example, a 4×1 cm anode with a 6cm cathode configuration.

Simulations using SimNIBS software were utilized to understand the general trends in the distribution of electric current across the grid electrode and to ensure that the measurement system would be reliable and repeatable.

A grid electrode with 5 measurement points allows finer spatial resolution and gives a fuller picture of the field intensity over different brain regions. With electrodes placed in strategic places in a grid manner, it could be possible to cover a wide area of the electrode-skin interface quite effectively. The coverage is necessary for comprehensive data on the distribution of electrical current in tDCS treatments. Such a grid layout will provide measurements that capture the variations across the different regions of the electrode and hence provide insight into the spatial distribution of current. Besides, the simplicity of a measurement system is crucial for practical use during in vivo experiments.

For example, an increase in the number of measurement points to 7 does not translate to a proportional increase in accuracy or information enrichment in the assessment of skin current distribution. This gives a balance between complexity and data granularities. A 5-point grid can achieve sufficient spatial resolution without overburdening the measurement setup with too many electrodes, which makes it difficult to interpret the data and complicates setup time. This simplicity allows

for a much easier setup and calibration, thereby reducing sources of error during measurements. Also, working on a grid pattern has its advantages regarding statistical coverage and minimizing measurement errors, especially for neurological applications. Therefore, the choice of a 5-measurement-point grid electrode is supported because it allows doing a more detailed and reliable analysis of electric fields under several experimental conditions and across different brain regions.

In summary, heatmaps of current density showed areas of higher concentration near the electrode edges. This confirms that the segmented grid layout can effectively highlight regions where current density is uneven, reinforcing the suitability of the grid design for testing and identifying critical problem areas in future electrode studies.

3.4.2 Laboratory test

Simulation results were crucial in developing the verification testing model by providing baseline data to design the in vivo measurement setup. The patterns observed in simulations indicated potential hotspot areas, which the model needed to validate in practical conditions which were taken into account when starting grid-electrode designing and first experiments without participants.

Before starting in vivo measurements, the measurement system was first calibrated and validated without subjects in the laboratory to ensure its accuracy and reliability prior to starting the in-vivo measurements. The main purpose was to compare laboratory results with simulation data, ensuring that the simulated current distributions closely match real conditions and that the measurement setups gives reliable and reasonable data.

The measuring system includes tDCS Sooma duo device and Sooma ComfoPad electrodes to provide 2 mA direct current. The anode is a grid-style electrode with 4 different measuring points at the outer edge of the electrode cup, besides a reference electrode as shown in Figure 14.

The first tests were conducted on moisturized paper and then to an arm of a volunteer, in order to check the functionality and the accuracy of the measurement system without the complications of the head region. In this way, the measurement system can be checked to work as expected and produce reliable results before performing actual in vivo measurements.

The electrode system is put on the arm of the participant; the contact resistance and the current distribution are measured in different points. With these measurements, calibration of the system is enabled and accuracy is checked. We test data collection, too. Afterwards, we analyze in detail the results of the laboratory tests to see if the measurement system works as expected.

3.4.3 In vivo testing

In vivo measurements were conducted to evaluate the ability of the measurement system and grid electrode to identify problem areas related to electrical current distribution under real-world operating conditions.



Figure 14: The figure shows a 3D model of the electrode cup used as an anode in in vivo measurements. The anode is divided into four blocks isolated from each other by a cross in the middle. In addition, locations for the measurement electrodes are placed on the outer edge of each segment. This solution aims to minimize the risk of the measurement electrode getting wet and the electrical interaction between the blocks.

During the in vivo measurements tDCS electrodes are placed on the participants' heads similarly to how they are used in tDCS therapy. In total 15 participants were tested and 12 of them participated 2 times. Measurement electrodes are positioned on the outer edges of the anode blocks instead of within the electrode area. This outer placement mitigates interference from hydrogel short-circuiting, enhancing data integrity by focusing on current at the anode boundary rather than within its surface. A reference electrode is placed on a neutral area, such as the shoulder.

2mA direct current is supplied to the tDCS electrodes. The cable leading to anode is also divided in four parts and it delivers current in all of the segments of anode. From the outer measurement points we can see how much current ends up in each segment compared to each other and compared with other participants.

Contact resistance and current distribution are measured at the different points of the grid electrode. Measurements are taken during the direct current delivering and the results are recorded for analysis with Micro 1401 measurement device and Spike2 Software. The cables from the grid electrodes are connected first to D440 amplifier/filtering device and then to Micro 1401 device that measures converts analog signals to digital form, measuring voltage and current throughout the treatment. This, in turn, is linked in real-time to computer software called Spike2 via a USB port, into which all data is collected.

3.4.4 Reproducibility and reliability of results

Accordingly, several tests under the same conditions are performed to ensure reproducibility and that the results are reliable. The repeatability tests are checked by repeating tests on different days and with different participants. All the data collected from these repeated tests are compared to check on any variation or inconsistencies.

The grid-style electrode has 4 measurement points on its outer edges, which enable detailed analysis of the current distribution and contact resistance over sub-areas of the electrode-skin interface. Swiss Dätwyler dry, brush-like electrodes were used for measuring. Moreover, measurements on various participants take into consideration individual differences in skin properties and electrode placement. This helps in evaluating the generalization of the results.

The measurement is performed to check the reproducibility of the data. It includes artefact filtering and statistical analysis. The standard deviation and coefficient of variation are determined for all the values of contact resistance and current distribution measured with different tests. Small variations of those parameters with high reproducibility of the results show that the measuring system provides reliable and repeatable results.

Overall, this systematic approach to testing and analysis indicates that the measuring system is reproducible and reliable and engenders confidence in the valid conclusion of the findings.

3.5 Data Analysis

The critical steps for the analysis of data in the measurements of the tDCS electrode interface are presented in this section, important to result in an accurate interpretation and visualization.

The analysis will consider the extent of current dispersion outside of the anode's segmented blocks to enable insight into inter-block current flow and 'leakage.' While the outer-edge electrode placement has some limitations due to edge effects, the data should still reveal significant inter-block variations and current dispersion beyond the intended electrode area.

MATLAB is used for the data analysis part because it offers strong tools for signal processing, statistical analysis, and data visualization, suitable for analyzing complex datasets from tDCS electrode interface measurements. This will enable the study of possible current distribution patterns, identification of influential factors, and drawing of conclusions on the electrical properties at the skin-electrode interface. In addition, it could allow one to provide further insight into optimization techniques in tDCS treatments regarding individual anatomical differences and electrode configurations.

3.5.1 Signal processing and data collection

Signal processing plays a crucial role in extracting meaningful information from the raw measurements especially with such a big data set (gain frequency is set to 2kHz and the measurements are 15 minutes long). This includes high- and low-cut filtering, already partly done during the data collection process. Next, movement-induced noise or electrical noise is attenuated using artifact removal techniques such as artifact rejection algorithms or averaging methods to ensure cleaner datasets. For impedance measurements, the signal will contain baseline drifts, which are corrected for by methods such as polynomial fitting or moving average subtraction to capture impedance changes without drift in the signal.

3.5.2 Statistical analysis

Such statistical analysis will be necessary to understand the variability in current distribution and impedance characteristics among the measurement points and across different participants.

First, descriptive statistics (mean, standard deviation, and range) of impedance values at all measured sites were calculated to give an overview of the spread in data and outliers. Besides that, correlation analysis examining impedance values with respect to participant characteristics-skin thickness, hydration levels-examines their variation and impact on electrical properties at the skin electrode interface.

Using t-tests, comparative analysis carries out impedance measures across conditions, like before versus during tDCS, or electrode types, such as different sizes, that show significant differences in an attempt to validate experimental conditions.

3.5.3 Data visualization

Effective visualization enhances interpretation of complex data patterns, and the findings are communicated more effectively.

Techniques also include heatmaps, which enable visualization of the present distribution across the electrode surface by showing the spatial insights into hotspots or poor contact areas. MATLAB is used because its heatmap function represents color gradients in accordance with impedance and current density variations. Line plots and scatter plots illustrate the trends in impedance measures over time or across frequencies and show the correlations between impedance values and participant characteristics.

Comparison between groups is summarized in statistical plots such as box plots or bar charts. Error bars or confidence intervals convey information on variability in the measurements.

3.6 Participant Recruitment and Ethical Considerations

Participants will be recruited through university email channel. To ensure participants' privacy and safety, the data will be anonymized and stored on password-protected servers and in locked closets. Adverse events of participants will be closely monitored during the tDCS sessions. Any medical emergency caused by this will be covered on the spot, and for any post-study concern, participants will be provided with contact information of the research members. These are measures to ensure the ethical integrity of this study and the welfare of all participants.

3.6.1 Inclusion/Exclusion Criteria

Participants must be 18-65 years old, generally healthy with no known neurological or dermatological conditions, and they need to be capable of providing informed consent. They also need to speak fluent Finnish and/or English. Individuals with pacemakers, metallic implants in the head, or other contraindications for tDCS will be excluded. Additionally, those with a known history of epilepsy, severe skin conditions, or other medical conditions that might interfere with tDCS are excluded as well as individuals taking medications that could influence brain activity or skin condition, and those who are unable to provide informed consent. Each participant will undergo two 15-minute sessions; based on Sooma system data indicating that this duration is sufficient to detect notable variations in contact resistance without significant deterioration in hydrogel performance. This protocol, conducted with approximately 20 participants, will reveal differences in current distribution dynamics effectively.

3.6.2 Ethical Approval and Informed Consent Process

Ethical approval will be obtained from Aalto University's ethical approval committee. A detailed study protocol should be sent to them, including, among others, the purpose of the study, methodology, and its eventual risks and benefits involved in the study. Upon approval, informed consent forms with all relevant details about the

study will be provided both in the information sessions and in writing to participants. Participation is voluntary, and participants can withdraw at will.

4 Results

Results are discussed in this section. The focus will be in illustrating, through the developed verification testing model, how the objectives set out in this thesis have been met. More precisely, this section will cover analysis of the present distribution across grid electrode segments, the validation of repeatability and accuracy of the measurement system, and the assessment of influence from individual participant characteristics. Such results are interpreted in terms of their significance and implication for future electrode design.

The developed measurement system for this thesis is robust in terms of its electrical properties at the electrode-skin interface and identifying potential issues in current distribution. There is good quality in the captured signal throughout the experimental sessions without too much external interference noise. Quantitative analysis demonstrates that the system exhibits good repeatability in the multiple measurements. The measured impedance values fall within the range compared with those from earlier studies indicating reliability in the developed system [64].

A more detailed comparison of the individual measurements suggested that signal stability varied depending on participant-specific factors, such as skin type and hydration level as expected. For instance, participants with drier skin (visually estimated) had slightly higher baseline impedance values, which is in line with earlier studies, that present skin condition as one of the most influential factors affecting electrode performance [44].

First, the description of the obtained results is considered, and afterwards, the measurement system, signal quality, and noise are introduced, taking into account such aspects as measurement accuracy and repeatability, current distribution across the electrodes, and spatial distribution patterns. Furthermore, we analyze the statistical aspects of the results on the basis of quantitative measurements, significance testing, and confidence intervals. We finally address the other observations on the effects of tDCS, regarding, for instance, correlations with participant characteristics, and compare results to the theoretical background, interpreting their significance and implications.

4.1 Description of the obtained results

The thesis successfully validated the verification testing model through both simulation and in vivo measurements. The results showed consistent current flow patterns through the segmented anode, with measurable variability at the interface that correlated with individual differences in skin conductivity and scalp anatomy. And by systematically identifying uneven current distribution and its contributing factors, the model has proven its capacity to support electrode development. The consistency between simulation and in vivo results supports the reliability of the verification

testing model. The participant-reported comfort levels further indicate that the system is practical and user-friendly, making it suitable for extended testing and comparative studies of different electrode configurations.

The findings of this thesis indicate the presence of substantial temporal and spatial variations in the distribution of current across the grid electrode for different subjects. For instance, in one case, a subject with higher skin resistance was found to have stabilization beyond 300 seconds. Again, this supports the properties of skin being an important modifying variable in current uniformity. In contrast, participants who have uniform skin-to-electrode contact showed faster stabilization, indicating the practical value of improvement in electrode material and design.

From the temporal point of view, density analysis indicates that the stabilization of current happened between 100 and 300 seconds after stimulation began as seen in figure 15. After this period, the current distribution was approximately the same, though at times it was disturbed-especially in the early minutes of stimulation.

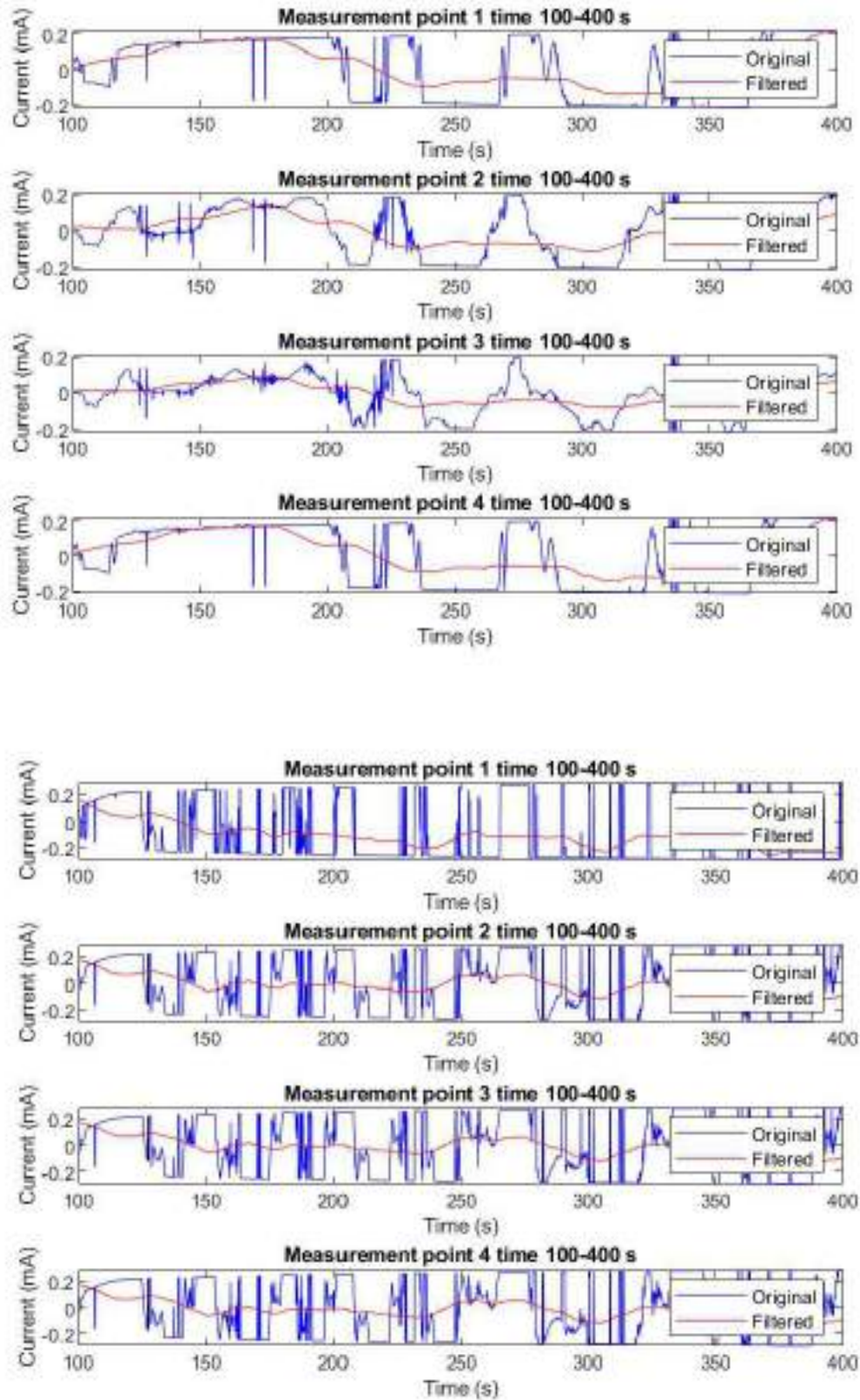


Figure 15: This figure shows the current stabilization timeline for two participants using all of their measurement points. In the upper subfigure, equilibrium is reached after approximately 200 seconds while in the lower subfigure this is reached after 150 seconds, with the differences coming from skin resistance and quality of electrode placement.

The application current from a spatial perspective was not homogenous across the electrode surface. Local "hotspots," or higher concentration areas, were evident in points where irregularities occurred either in the skin-to-electrode contact or where inhomogeneities existed in hydrogel moisture conditions. Such could be based on factors involving skin properties-thickness, resistance-pertaining geometry of the electrode structure.

Further measurements within the experimental setup maintained the view that electrical properties of the material, particularly resistance and capacitance, were relevant in determining current distribution. A more profound analysis of such properties revealed that leakage currents across the electrode segments were considerably stronger for some electrode segments relative to others.

Key Observations were:

1. Poor skin-to-electrode contact areas provided places where current concentration may occur.
2. Only minor but significant differences in the current distribution depending on their locations existed between the electrode segments.
3. The current was distributed and flowed through either density or resistance of materials, which are very basic electrical and physical properties.

4.2 Performance of measurement system

The measurement system designed for this thesis demonstrated robust performance in capturing electrical properties at the electrode-skin interface. It was evaluated based on its performance to provide correct and repeatable results, the quality of the signal, and the noise. One of the problems were although the risk of measurement electrodes getting wet. Hydrogel leakage onto the measurement electrodes however could be prevented by using materials or separators that better confine the hydrogel in the future. The challenges identified indicate that the verification model works well, but refining the protocol and controlling the hydrogel would improve its usability. Also important to note is that the measurement setup, as done, is not very exact, and significant error most likely exists in the absolute voltage values measured. The system lends itself quite well to comparison among measurement points, however. Effective evaluation of the current distribution across the segmented electrode blocks is thus easily achieved with the system.

Future designs can use some kind of physical barriers, to keep the hydrogel confined within the electrode area to reduce leakage. Also, absorbent or non-conductive barriers around the measurement points would further isolate the measurement electrodes and ensure consistent data collection. Also more measurement points combined with more advanced mathematical modelling tools could make the setup more precise.

4.2.1 Signal quality and noise

The results demonstrate that the developed verification testing model is functional and reproducible. It provides a basis for further research and development of tDCS electrodes, especially to improve current distribution and user comfort. The measurement system showed good capabilities in signal data collection, which was generally of high quality. The temporal density plot from time period 300 to 500 seconds indicated that the signal disturbances were primarily concentrated at an early phase of stimulation from 0 to 100 seconds when the electrodes were adapting to the skin. During the signal processing phase, low-frequency disturbances were removed, which thus improved the accuracy of the measurements.

The high S/N ratio in certain measuring points provided a warranty of collecting data of high quality. Besides, disturbances that occurred in an initial phase of stimulation did not influence the repeatability of long-term measurements.

4.2.2 Measurement accuracy and repeatability

The accuracy and repeatability of the measurement system were assessed by using heat map analyses along with statistical methods. Results showed that the measurements were very repeatable for individual test subjects but highly variable between subjects.

The main findings were as follows:

In most measuring points, the measuring system showed quite good accuracy, while significant variation was observed at point 4 as seen in figure 16. This could be an effect of problems with keeping the hydrogel at a constant moisture level due to migration of saline solution to hair. It is seen that the degree of variability is higher in measurement points 1 and 4 compared to points 2 and 3, which is a clear indication of edge effects in the distribution of current.

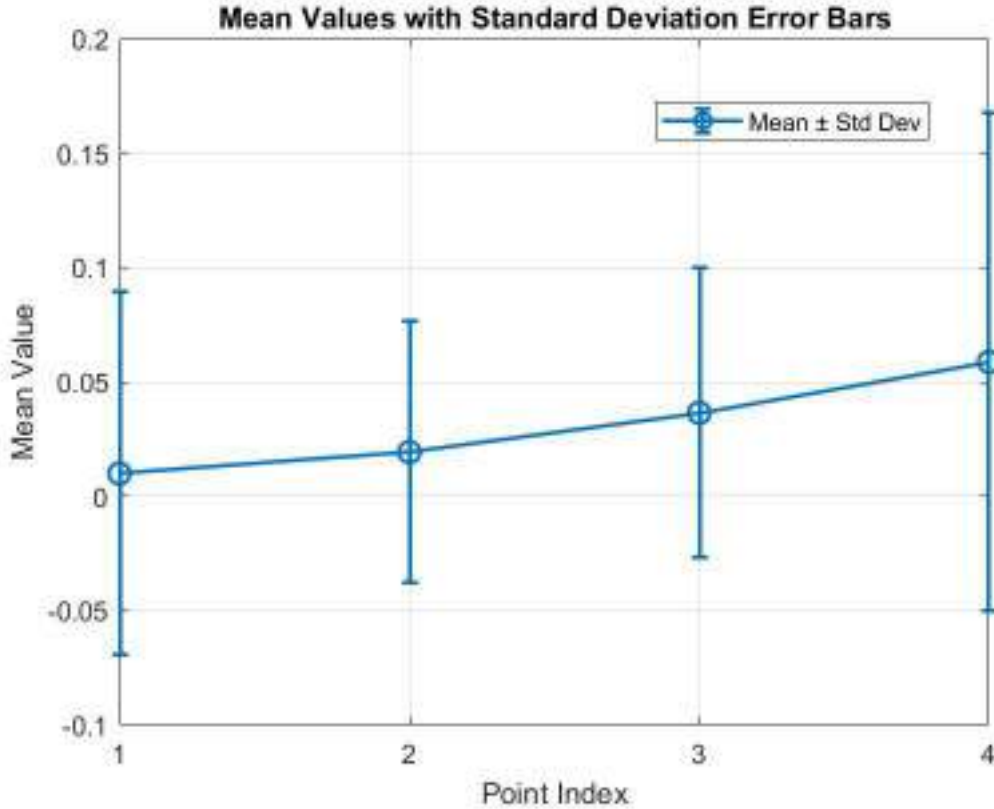


Figure 16: This figure illustrates measurement accuracy at different points. Measurement points 2 and 3 showed higher consistency compared to points 1 and 4, where deviation was greater, highlighting the edge effects and hydrogel challenges.

4.3 Distribution of current across the grid electrodes

One of the key questions of the thesis is the distribution of current across the grid electrode area. As indicated in earlier studies, the distribution of current within the grid electrode area is a complex and often heterogeneous process influenced by various factors such as electrode geometry, skin moisture, and contact quality. The analysis also showed inter-individual differences in the distribution of stimulation current on the skin surface. This result underlines the impact of individual anatomy, such as skin thickness and hydration level, on current flow. In addition, it was pointed out that even with the use of grid electrodes, which helps guide the current, the final distribution may still be subject to patient-specific variability. This emphasizes the need for personalized stimulation parameters to maximize control and accurate measurement of current in future feedback loop based stimulation methods.

In this thesis, measurements were taken from outside the electrode, without extending into the electrode area itself. This limits any direct inference about the distribution currently on the electrode itself. The simulation and in vivo measurements of the current distribution across grid electrodes however showed clear patterns and irregularities. Simulations using SimNIBS software demonstrated a generally

uniform current density within electrode segments, though regions of increased current density (or "hotspots") were observed near the edge of some segments. Such findings corresponded to localized increases in skin resistance or incomplete contact of the electrode to the skin. These findings point out that there is potential in personalized treatment where electrode current is controlled and accurately measured to achieve uniform current distribution on skin-electrode interface. While the grid electrode setup already offers improved control, the variability in current distribution suggests that future developments should be adaptive systems capable of adjusting in real-time to individual anatomical characteristics.

It should be underlined that several previous studies, by the use of mathematical modeling, have indicated that the current is highest at the edges of the electrode itself-edge effect. Consequently, placing measurement points only outside the electrode may distort the picture of current distribution, particularly in areas near the skin-electrode contact boundary. The edge effect might lead to the leakage of current outside the electrode and distort the measurement results by indicating a high value of current whereas, practically most of the portion is actually flowing towards where it actually should. This setup however still gave an idea as to how different segments of anodes differ from one another. The results of this project confirm the previous mathematical models about the edge effects, in which the maximum current density is at the edges of the electrodes because of small surface area and poor contact. One possible solution involves redesigning the electrode geometry to include tapered edges or increased padding thickness, reducing the current density at the borders.

4.3.1 Hotspot Identification and Mapping

The focus was on the analysis of the places of the highest current densities observed at the measurement points, as illustrated in figure 17. This heat map gives the visualization of a current distribution for different participants at the current distribution, and also shows that there is much variability. A number of points that show high maximum current might be the concerns, but again, this also was out of the electrode area where, indeed, it could be low compared with underneath the electrode areas. This means that while the heatmap does indicate some "hotspots," it is not possible to determine the nature and intensity of these "hotspots" within the electrode region directly from this data. It is most probably caused by irregularities in the skin-electrode interface and/or improper placement of the electrode. This observation calls for a re-evaluation of electrode design for better contact to be made, reducing the inhomogeneities in current distribution. While the highest current densities were recorded in some areas, these findings may partly be attributed to the fact that the measurement points were outside the electrode, and no data was available from the electrode area itself.

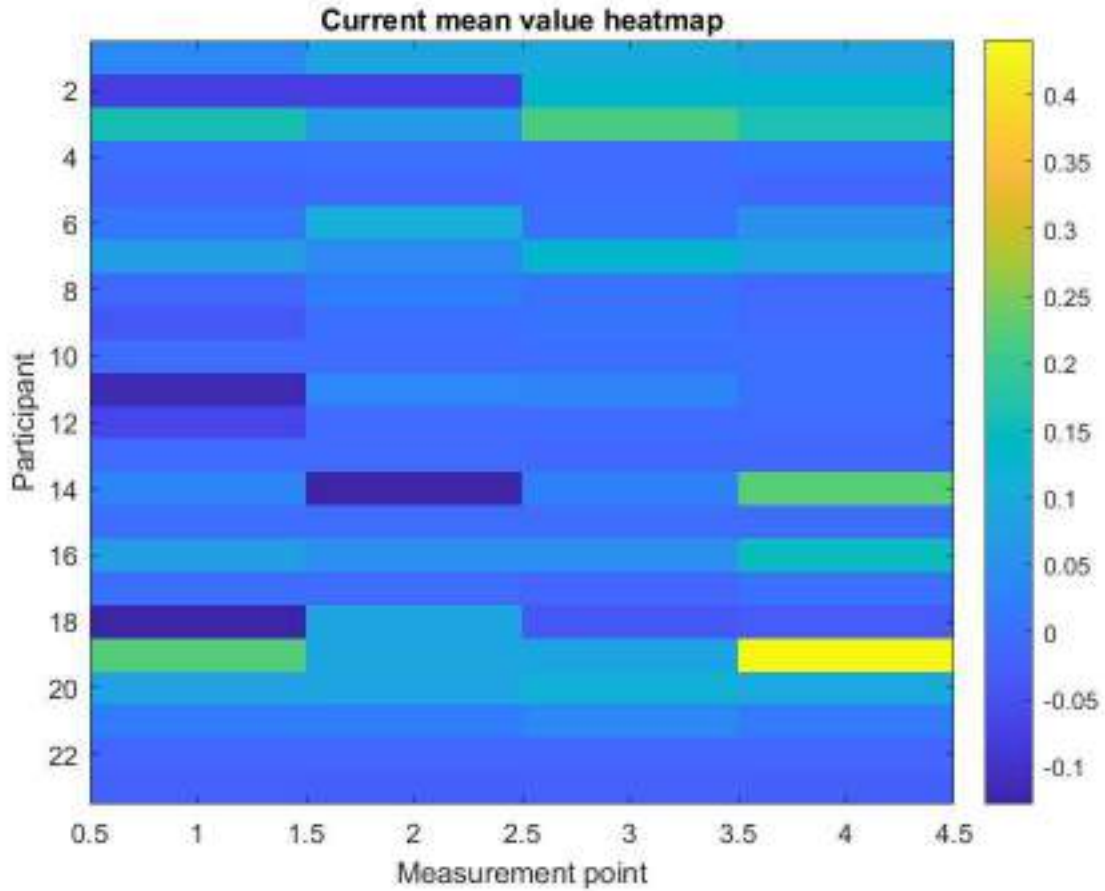


Figure 17: A heat map showing the distribution of current across different participants.

From the heatmap (Figure 17), it clearly appears that the current distribution for any particular individual may be significantly different, especially for individuals with higher skin resistance and/or different scalp structures. The edge effects also influence the shape of the heatmap, since these are known to give higher current densities in proximity to the margins of electrodes. Although data from the region over the electrode itself is unavailable for this study, mathematical modeling of tDCS in prior studies has suggested that this can strongly affect the distribution of current. Therefore, the results support the development of verification testing models that are able to capture this nuance and provide a more complete understanding of current behavior across the electrode-skin interface.

4.4 Statistical Analysis of Data

The section presents the statistical results as well as the statistical methods used in analyzing the results of measurements. These findings provide a deeper understanding of the interplay between individual anatomical characteristics and electrode performance. Since the results of tDCS measurements often include data with significant dispersion and variability between individual measurements, proper statistical

tests and confidence intervals need to be applied to establish the significance and reliability of the results.

4.4.1 Quantitative Results and Measurements

By applying the appropriate statistical treatment, large variability could be found in the means and dispersion at subjects and measurement points. Looking at the averaged data from both graphs above, the following general trends emerge:

The range of values is larger at points 1 and 4 than it is at points 2 and 3.

There is a more symmetric distribution at measurement point 3 with respect to positive and negative current values, hence greater variability.

The distribution of the values of current density in various subjects is presented in the figure 18. The chart presents a more detailed view of the measurement results.

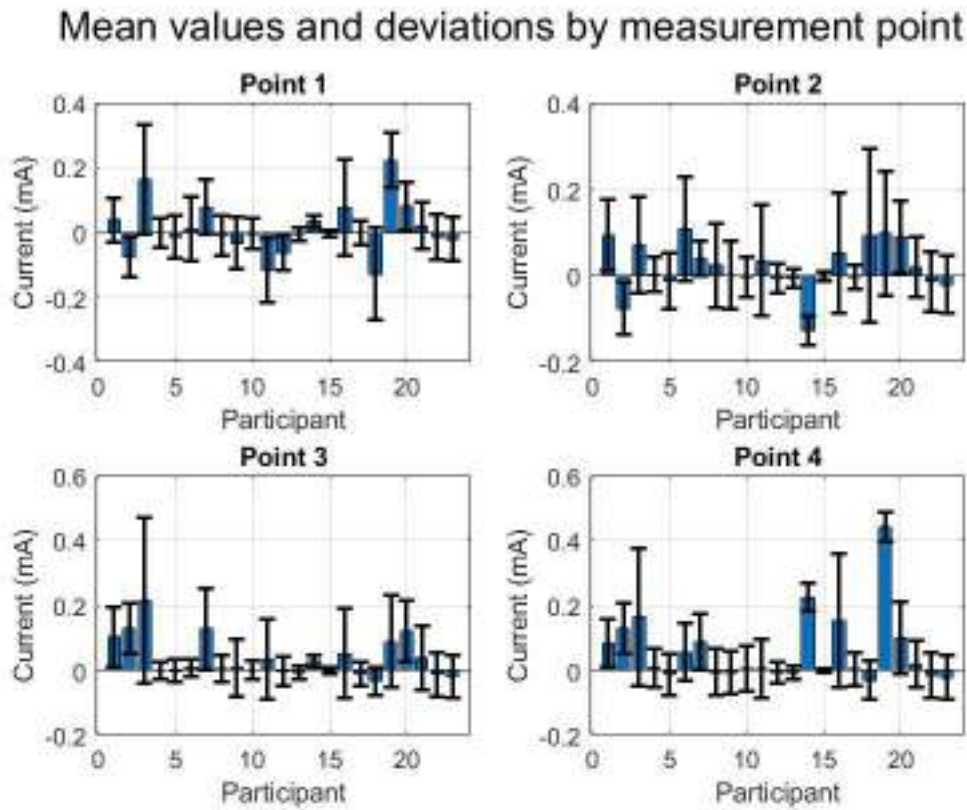


Figure 18: The distribution of the mean values of current for each participant in different measurement points.

A general observation was that the higher the impedance at the site of electrode placement, the more uneven was the distribution of current. This suggests that individual factors such as skin moisture and thickness of the fat layer could be affecting the repeatability and accuracy of the results of measurement.

4.4.2 Significance Testing and Confidence Intervals

Significance of these findings were confirmed using t-tests. Although the p-values indicate no significant differences at all between measurement points, trends do point to edge effects and inter-individual variability. These findings highlight that a larger sample size in any successive studies will be highly useful in confirming these observations. Key statistical findings agree on individual skin property and electrode geometry differences being a plausible cause for the discrepancies that were observed.

Table 6: p-values for estimating the difference between measurement points.

Measurement points	p-value
1 vs. 2	0.64769
1 vs. 3	0.21559
1 vs. 4	0.089245
2 vs. 3	0.33937
2 vs. 4	0.13153
3 vs. 4	0.40141

The table 6 presents the test results for current comparisons between measurement points. Even though none of the results was statistically significant, they pointed out the trends of variability between points.

4.5 Other observations regarding the effects of tDCS

This section describes some further observations concerning the effects of tDCS and their possible relations to participant characteristics. These findings represent further support for the optimization of the electrode-skin interface as one factor in enhancing user experience and therapeutic efficacy. Moreover, the observed individual differences underline the potential benefit of tailoring tDCS protocols to participant-specific characteristics.

4.5.1 Correlations with Participant Characteristics

Individual differences in the current distribution seemed to significantly relate to physiological factors, especially skin characteristics. For instance, those subjects with lower skin resistance exhibited a more homogenous current distribution. This result suggests that skin resistance may be an influential factor in the flow and distribution of current during tDCS stimulation. These individual differences could also be explained by skin moisture content and structural variations in the scalp, such as hair density, affecting electrode-skin contact. This inter-individual variability highlights the importance of the verification model in identifying how individual physiological factors, such as skin moisture and resistance, interact with electrode performance. Future iterations of the model could include pre-measurement skin preparation protocols to standardize these effects.

A heat map showing the current distribution across different participants makes a person's current distribution highly variable, especially in the case of higher skin resistance and/or different scalp structures. Individual characteristics, like skin resistance or moisture, are important aspects in designing and optimizing the set-up or form-factor electrode geometry with treatment options or protocols in tDCS. This heat map puts into light how the spread or distribution of currents may well resultantly differ among participants and through their factors of how the effects actually come into the precision level required in tDCS-the spread of current over that area and its impact on the part of the brain covered.

This further emphasizes the need to consider individual factors like skin resistance and moisture when designing and optimizing electrodes for tDCS treatments. Second, the heat map visualization may provide insight into exactly how current distribution can differ between individuals. Again, accounting for these individual factors in understanding may help better the treatment and its accuracy of tDCS, since these factors affect how well the current is distributed over any one area and how well the stimulation affects the desired portion of the brain.

The correlation matrices for voltage and current, can be seen in Figures 20 and 19 and in table 7, also give interesting information regarding the relationship between the variables of current distribution and participant characteristics. Voltage behavior correlation is shown in the following voltage correlation matrix, where large values indicate strong electrical coupling or exposure to common external conditions. Large values of 0.6071 and 0.6058 represent points 1 and 4 and points 1 and 2, respectively. In contrast, a relatively weaker correlation of 0.2411 is returned for points 2 and 4, indicating less electrical interaction or divergence in current pathways between these points. In summary, the matrix brings into focus both well-connected points and regions of variability that could be due to differences either in circuit placement or physiological factors like skin impedance.

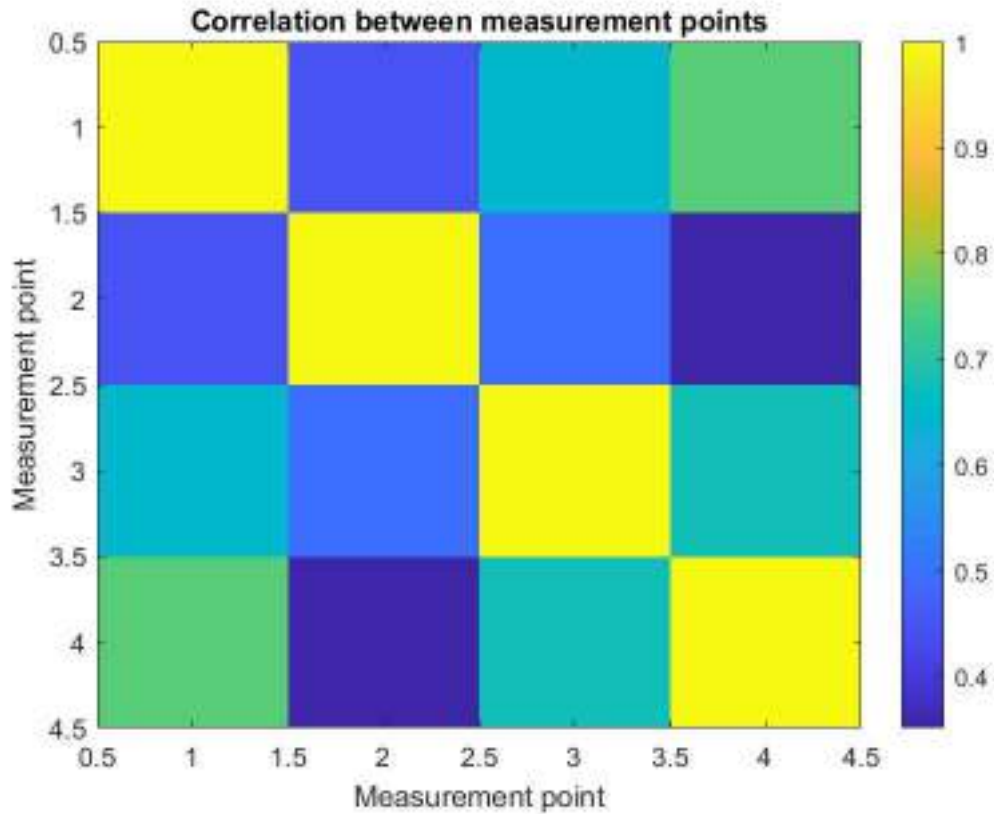


Figure 19: Pearson correlation matrix of currents between measurement points. Strongest correlations are between 1 and 4.

The present correlation matrix complements this analysis by showing how currents behave across measurement points.

	Point 1	Point 2	Point 3	Point 4
Point 1	1.0000	0.4494	0.6470	0.7548
Point 2	0.4494	1.0000	0.4961	0.3507
Point 3	0.6470	0.4961	1.0000	0.6751
Point 4	0.7548	0.3507	0.6751	1.0000

Table 7: Correlation matrix values of currents between measurement points.

The highest correlation, at 0.7548, is between points 1 and 4; these points are likely in the same circuit or have strong mutual electrical interaction. Similarly, points 1 and 3 show a moderate correlation of 0.6470, and between points 3 and 4, it is 0.6751, whereas the weakest is again between points 2 and 4 with 0.3507, as consistent with the voltage matrix. These patterns give insight into how currents flow through the system and suggest areas of strong and weak connectivity.

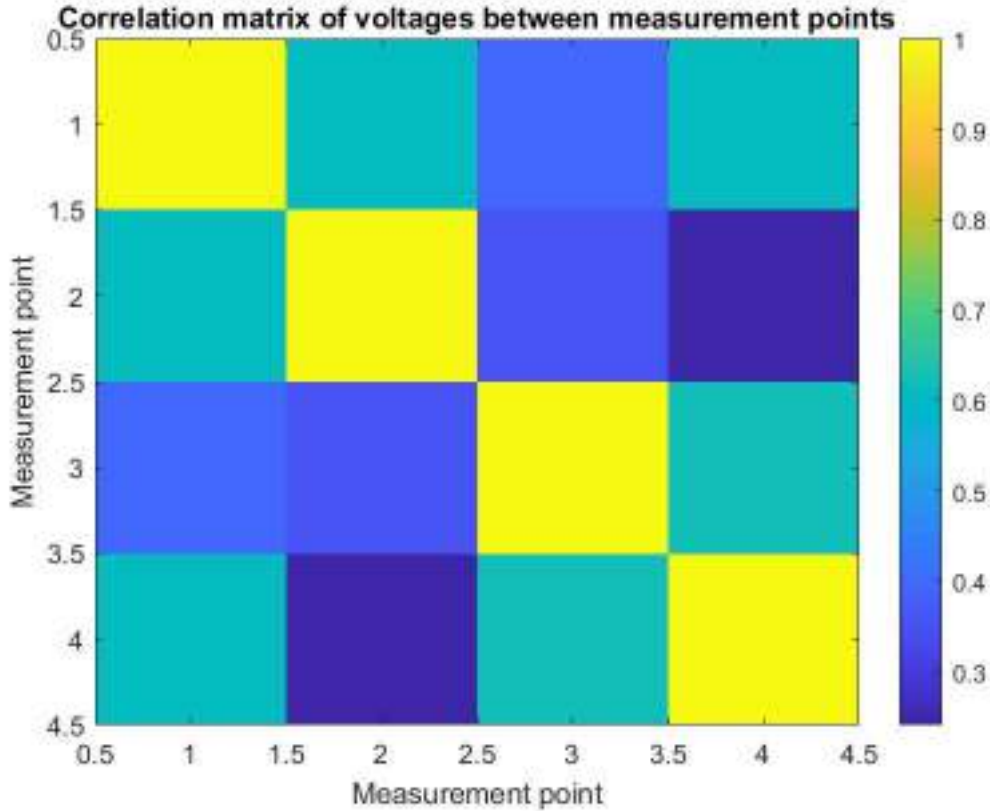


Figure 20: Pearson correlation matrix of voltages. Notably, measurement points 2 and 4 are weakly correlated with each other, having a correlation value below 0.3.

Correlations reflect how similarly voltage behaves at different measurement points throughout the treatment and may uncover possible causes of current inhomogeneities or deviations. Low correlation, for example between points 2 and 4 (as shown in figure 20) may indicate that the present distribution between these points is very different. It is possible that the pathways of current between these points diverge—for example, near the edges of the electrode or because of structural differences, causing that different voltage behaviors are produced in the skin. On the other hand, there are high correlations between measurement point pairs 4 and 1 are about 0.63, indicating the points where voltage behaves more alike.

A comparison of the voltage and current correlation matrices seems to indicate that, in most cases, correlations are stronger for current than for voltage; this may reflect the more strongly interconnected nature of current flow than voltage variations. In both matrices, measurement points 1 and 4 show up consistently with strong relationships, pointing out their important electrical interaction. In contrast, points 2 and 4 have weaker correlations in both contexts, indicating only a limited connectivity or functional divergence of these regions. Voltage is less correlated, 0.2411, than current, 0.3507, and this may reflect the greater impact of structural or physiological differences on voltage distribution.

These findings therefore support the hypothesis that scalp structure and individual physiological properties have a significant effect on current distribution and voltage

behavior. Understanding such correlations and driving factors behind these will be important in the development of better electrode placement and optimization of tDCS treatments with enhanced efficacy and reduced discomfort.

4.6 Analysis of results

This section discusses the results of the research, contrasting the outcomes against the theoretical framework and other related works. Such work would seek to contextualize the results among already existing tDCS set-ups and theoretical models relevant to the study, before evaluating the importance of these findings both for practical implementation and for further research.

4.6.1 Comparison to theoretical background

The good agreement between the results and the theoretical models confirms the validity of the testing framework. However, the deviations at the edges in the distribution of current emphasize practical issues with achieving uniform stimulation that should be improved by new designs through iterative testing. Symmetry of current distribution and current staying close to zero for most of the time suggests no significant unilateral current leakages or inhomogeneities in the measurements. This agrees with the estimations using tDCS equipment and simulation models, which assume that current distribution would be uniform if the interface of the electrode-skin was perfect.

As was seen in figure 18, average current and standard deviation at every measurement point depict how current is distributed and varies among subjects; this might also reflect individual differences in electrode placement or skin characteristics.

The result is further supported by the simulation models of current distribution across different tissues, which predict a symmetrical distribution of current in cases where there is uniform contact between the electrode and skin. In the same way, variations across different measurement points are indicative that individual factors may produce deviation in the current distribution of the subject - an important design consideration when developing clinical application of tDCS.

4.6.2 Interpretation and significance of results

These findings have several important implications, both in terms of treatment efficacy and user experience. They form a strong foundation for these methods of developing and verifying of tDCS electrode designs. For example, edge effects observed highlight the need for segmented electrode designs that are able to more effectively constrain current flow. High variability between subjects indicates the importance of treatment protocols that will be individualized for skin resistance and other physiological factors. Future studies should be geared toward testing other electrode materials and geometries in a range of populations to generalize these findings and further optimize the development of electrode designs to more personalized and accurate controlled systems. This project serves to validate this type of experimental setup and demonstrates the potential of using grid electrodes

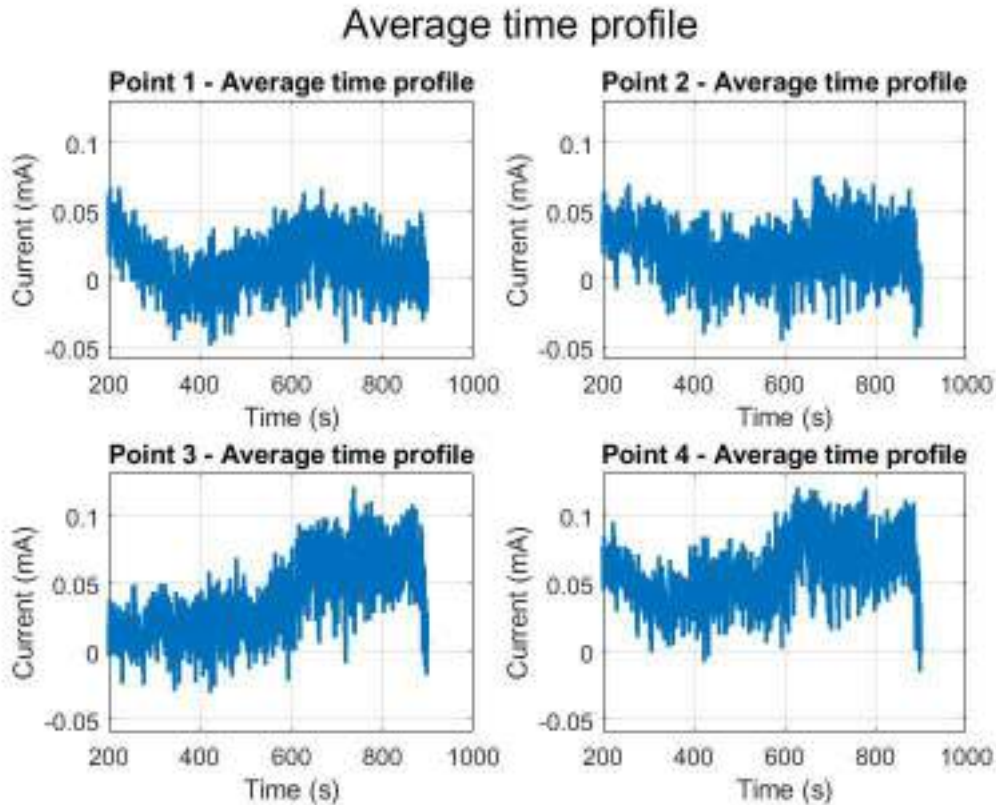


Figure 21: Mean current values of all participants in a time interval 200-900 seconds. This figure represents the changes in measurement points across individuals and how the mean current values can vary near the anodal region.

in tDCS treatments, as they seem to maintain current distribution irregularities at a reasonable level. Such an experiment with a traditional anode in similar settings will determine whether this is, in fact, improved due to the segmented electrode design. The relatively small skin irritation, pain, or burning sensation of the subjects in vivo is, nevertheless, reassuring. This is actually the most important step in bringing more precision and reliability into the development of novel tDCS treatments.

This thesis also reinforces previous findings that current distribution may vary individually, which means that optimization of the electrode-skin interface is needed. For example, in measurement points where the current distribution was less homogeneous, such as at the edges of the anodal region, potential edge effects and leaking points were observed that might be more personalized and accurately controlled feedback loop systems. That means any future design of electrodes should consider finer tuning to manipulate the distribution of current to ensure most of the electrical current is concentrated on the ROI rather than stagnating on the skin by virtue of a bad contact.

The practical significance is also considerable: inhomogeneous current distribution, especially at the edges of the anodal region, could lead to discomfort, like tingling or skin irritation, which may lower the user experience. This can be a critical factor,

especially in clinical settings where treatment customization and patient satisfaction are crucial.

Filtered and averaged current within a time frame between 200–900 s can be seen in figure 21. Local current deviations provide insight to optimizing the electrode layout configuration. One can see that in the measurement point 4 the risk of current leakage is the biggest, since the measured values of current are greatest.

The findings indicate that the grid electrodes have a chance to further improve the optimized comfort during tDCS treatments and revealed that interindividual differences in electrode configuration optimization may play major roles in tDCS skin-electrode current distribution variability. These insights will help toward better tDCS treatment recommendations and research approaches that consider individual variability.

5 Discussion and conclusions

This thesis provides new insights into the specific factors that determine current distribution, including electrode segmentation and participant-specific variables such as skin hydration. Systematic mapping of current distribution across grid-style electrodes furthers understanding of how best to optimize electrode-skin contact for a diverse user group [64].

The discussion and conclusions will be presented in this section. First, the interpretation of the results within the framework of the existing literature is presented, highlighting consequences for the optimization of tDCS electrode design and for the advance of stimulation methodologies. Further, this is followed by comparing the findings with the earlier related studies to outline the consistencies and discrepancies. This was followed by a discussion on the possible sources of error and limitations of the thesis, adding that from such insights, valuable lessons could be gleaned where improvements may be affected. This is followed by the scientific and practical contributions of the research, emphasizing the importance for future developments in tDCS technology. Further, recommendations are made on the direction of future research work in ways to increase the accuracy and control of applications and clinical and therapeutic use of tDCS.

5.1 Discussion of results

The findings of this thesis are in line with previous studies that identify electrode-skin interface properties as a key determinant of current distribution [45], [14]. The leakage currents at the edges, agree with previous mathematical modeling predictions that edge effects enhance uneven current density [8] [25]. While these findings extend previous studies in important ways, providing for the first time in vivo measurements of how such patterns are modulated by physiological factors-skin moisture and anatomical variability-in real-world conditions.

This thesis represents the importance of leakage current in tDCS treatments. The results show that leakage current from the anode can occur, which may reduce

the targeting of the stimulation in the target area. This finding points to the need for optimization in electrode design and materials. It agrees with the literature that uniform current distribution within the anodal region is important for maximizing accuracy and control of current distribution on skin-electrode interface. With the grid-type electrode, there is a more precise analysis of current distribution, offering new possibilities to improve current control and, with it, precision targeting.

Correlation between measurement points also shows that current distribution under the electrode is dynamic, having great spatial variability. For instance, a high value of the correlation factor (about 0.63) for points 4 and 1 indicates the same kind of behavior of the current in those places. On the other hand, the weak correlation below 0.3 between points 2 and 4 reveals great local differences in the current behavior. These observations might suggest leakage currents at the edges of the electrodes or structural changes at the skin-electrode boundary, such as different moisture and resistance conditions. Besides, small differences in head size and geometry resulted in minor variations in the anatomical positioning and skin contact at the measuring points.

The findings highlight the need for future electrode designs to consider individual anatomical variations. The distribution of current showed dependence on conditions such as skin moisture and hair density, as shown in previous studies too. These factors are practically impossible to standardize between different user groups. Such results emphasize how the addition of for example closed-loop systems might benefit tDCS devices. Biosignal measurements could be brought into future setups to automatically adjust the location and amplitude of stimulation to account for individual variability, thus yielding more precise and effective personalized current distributions.

In line with previous studies, significant individual variability in current distribution was further observed, pointing to personalized tDCS protocols. Temporal changes in current flow, likely due to increased skin moisture during stimulation, further indicated dynamic interactions at the electrode-skin interface, as suggested by previous work highlighting skin impedance as a variable factor. A better understanding of these factors is crucial for the further development of tDCS therapies.

5.1.1 Comparison with previous studies

The results are in line with previous studies, reporting individual differences in the current distribution, these studies have relied heavily on computational simulations for the study of current distribution [15]. However, this thesis brings new insights with an added dimension of in vivo measurements using a grid electrode, that allows a finer resolution of current hotspots and edge effects. The use of grid-type electrodes, especially in the identification of hotspot regions and a more detailed analysis of current distribution. Unlike previous studies that have relied largely on mathematical modeling [18], this study incorporates in vivo measurements, adding a practical dimension that makes the results more realistic.

These results complement the existing knowledge while challenging some assumptions- for example, a near-homogeneous current distribution across the anodal region underlying many mathematical models. The observation could be one that would

contribute to more accurate simulation design and practical measurement setup in the future. These results point out individual factors, such as head anatomy and skin properties, as the most influential factor affecting individual current distributions on skin-electrode interface. These further reinforce the need for developing adaptive stimulation systems that take into consideration each patient's unique variability.

5.1.2 Potential sources of error and limitations

A major source of error involved placing the measurement points outside the electrode area, thus limiting insight into the actual current distribution at the skin-electrode interface. Another significant issue could be current leakage, particularly around electrode edges, right next to the measurement points. Also the contact quality of the measuring electrodes with the skin and possible displacements of the hydrogel could interfere with the signals [26] [36]. Keeping measurement electrodes dry was hard; at some points in the measurement of some of the subjects, the spread of saline solution was uncontrolled and needed to be excluded from analysis. At points 1 and 4, such spreading happened; at measurement point 3, the anatomy is such that many people do not obtain stable contact there. Designing the containment strategies for conductive gels and using materials with higher adhesiveness in the future will be of great benefit.

Individual physiological differences among participants, such as skin resistance or moisture, may have led to some deviation in the results. The simulations performed supported the experimental results only partially since they were not truly in vivo conditions. Sensitivity of the measurement system to interference could also have been a factor affecting the accuracy of the results.

As stated in its technical specifications, Digitimer made the D440 amplifier somewhat poor for DC measurements [17]. The susceptibility of this amplifier to noise and signal artifacts most probably became an issue of some kind, here, consistent with earlier misgivings over motion artifacts in the recording of biopotentials, which raise demands on the strength of measuring systems [69]. These artifacts were however filtered in MATLAB after-hand. It is conceivable, however, that the filtering methods used inadvertently remove usually some relevant information too.

One of the limitations included a small sample size, $n=15$, and participants being a homogeneous group of healthy volunteers with little variation in age range, which limits the generalization for the findings to patient populations of different ages. Besides, there were technical limitations in the equipment, especially regarding signal accuracy and performance of the measurement electrodes. Future studies should explore methods for standardizing these variables to further enhance the reliability of the findings.

All of this analysis illustrates that the use of modern mathematical modeling methods, for instance, the Laplace method, could be significantly important for further improvement of the measurement system and the enhanced study of multi-input configurations. These models could provide a stronger foundation for analyzing and improving current flow.

5.2 Contribution and significance of the study

This thesis provides the first systematic framework to develop and validate the electrode-skin interface properties for future research as well as important insights into the optimization of tDCS electrode design and highlights the critical nature of electrode material and shape selection as well as the configuration for current distribution control. The grid-style electrode design and the associated measurement setup allow for an accurate mapping of the current distribution and thereby fill in some of the gaps in the literature, which lacks in vivo validation of simulation results. These results point to great possibilities in enhancing treatment accuracy and control by minimizing current leakage and accounting for physiological variability between individual. This optimizes electrode design by reducing hotspots and increasing user comfort, further improving the state of tDCS technology.

In addition, the thesis has underscored the need for personalization of electrode configurations and protocols in relation to individual anatomical characteristics. This represents a major development in the personal approach to tDCS design in the quest for consistency of therapeutic outcomes across diverse populations. Thus, the innovative use of grid electrodes points toward a promising approach for more homogeneous current flow to enable future tDCS technologies in the development of improved therapeutic applications and new design strategies.

5.2.1 Scientific and practical implications

The thesis provides scientific and practical insights into the development of tDCS electrodes as pointed out also in previous studies [7]. The grid electrode design described herein represents a very promising approach to minimizing current leakage and increasing targeting precision. Moreover, this study extends earlier studies on the importance of electrode placement and montage in optimizing stimulation effects [14]. Beyond that, the measurement setup was functional and flexible; it could be used to test different models and configurations of electrodes for the development of future electrodes.

It opens up opportunities in practical applications for the development of personalized tDCS treatments, where the distribution of the current could be optimized using real-time feedback. Such a development would be particularly beneficial for user groups such as neurological patients whose therapeutic needs may have specific requirements. Skin irritation due to the grid electrode might further be reduced in possible subsequent developments of the electrode model.

5.2.2 Potential applications and future research directions

There are disadvantages to the developed model with grid electrodes, which mainly is the leakage current between each segment of the grid. Controllability may also improve by a multi-input model where each grid segment feeds separately controlled current. However, these models also suffer from similar drawbacks as the existing structure since current leakage persists in multi-input configurations. In practice, however, this would also call for advanced measurement techniques and accurate

measurement of current in every segment. The problem requires extensive research whether the existing models would serve the purpose or all new methodologies are to be invented.

These insights of the thesis therefore have implications in both the clinical and research uses of tDCS. Further research is thus warranted to examine whether scaling up the developed framework within the thesis can be generalised to a larger population that is heterogeneous. Also, advanced integration of bio-compatible materials in electrode design might include conductive polymers or graphene that can contribute to better distribution of currents and biocompatibility. This may also hint at the direction of personalized tDCS protocols, given certain anatomical and physiological features.

Furthermore, the further refining of the measuring electrodes-for instance, utilizing models based on Laplace would better position enhancing the effectiveness of the multi-input model assessment with an increase in preciseness within current distribution. This can minimize problems associated with uneven flow. Also involved are the possibilities of improving the simulation models and making a detailed study of the properties specific for particular tissues. Precise modeling of the number and place of measurement electrodes is rather important in view of current targeting optimization and minimization of hotspot effects. The contribution of measurement electrodes in the reduction of hotspot phenomena by nonuniform current distribution needs to be considered in the models, too.

Active adaptive algorithms for real-time monitoring and current distribution adjustment based on impedance profiles of participants may also provide a new revolution in applications. These could provide dynamic optimization of stimulation parameters for both efficacy and improved user experience.

So, further refinement of electrode design, improvement in the treatment outcomes, and the extension of clinical applicability are possible through the following:

Advanced Computational Modeling and Simulation

The numerical modeling, like Finite Element Method analysis, is useful in simulating the distribution of current for different electrode configurations. These simulations could allow one to obtain the degree of distribution of current flow across different layers-skin, skull, and brain-along with the inclusion of realistic anatomical and physiological parameters [11]. Such a model could be the foundation for the design of geometrical electrodes to maximize the delivery of current at a target area with a minimum spreading of unwanted current. In silico simulations supported by in vivo and in vitro validations will enhance the reliability of the findings. Real-time modeling, in combination with adaptive stimulation systems, can further dynamically optimize current flow based on patient-specific responses.

Biocompatible and Efficient Electrode Material Development

The potential subject of new material development is research into new material development offering better conductivity, lower impedance, and high biocompatibility. Novelty might relate to multilayer electrodes, hybrid composite materials, or new hydrogels. Such developments aim at eliminating the current hotspots and irritation of the skin. Furthermore, advances in materials development may enable designing single-use or multiple-use electrodes suitable for different clinical uses, even for

long-term therapy [2].

Clinical Research in Greater Perspective

Most of the current studies, including this thesis, are being conducted in healthy volunteers, which seriously limits the generalization to wider clinical contexts [10]. Further research should be extended to a variety of demographic and clinical groups, including elderly participants or neurological patients. Such investigations may reveal specific problems and opportunities for optimizing tDCS for specific groups, thus guaranteeing the effectiveness and safety of treatments. tDCS application research for stroke rehabilitation, depression, or neurodegenerative diseases may yield more specific details related to electrode design and stimulation protocols which could be applied on an individual basis.

Innovations in Electrode Geometry and Design

Indeed, this thesis has shown a potential of grid electrodes to lessen current leakage. Further studies might be geared towards further refinements in electrode geometries, including tapered and multi-compartment geometries in optimizing current distribution. Dynamic electrode systems able to make real-time adjustments to individual physiological variations could optimize treatment efficacy. Furthermore, this could be further improved regarding both safety and comfort through the inclusion of sensor technologies that would monitor skin impedance and temperature during stimulation.

Simulation to Clinical Outcome

Future studies shall move in the direction of relating these computational findings with clinical observations. This would drive electrode designs to optimize current delivery into the brain tissue with reduced hotspot effects, but needs to be complemented with a more in-depth look at tissue-specific variability for better usability and precision in the models.

In these respects, studies on whether the homogeneous distribution of current also improved therapeutic efficacy in different groups of patients would give more practical significance to electrode optimization. Long-term studies regarding how electrode configurations affect longevity and effectiveness of treatment for the purpose of providing data necessary in the refinement of stimulation protocols.

In conclusion, a focus on these research directions will move the field of tDCS toward personalized effective neuromodulation therapies. Improved simulations, new materials, and electrode designs will continue in tandem with extended clinical validation: the technologies of tDCS are bound to improve. The patients will show better results, and so will be opened up newer therapeutic avenues in neuroscience and beyond.

5.3 Conclusions

This thesis fulfilled its aims in developing and testing a rigorous framework for electrode-skin interface analysis in tDCS, and it determined several factors that modify current distribution, including that grid electrodes can reduce the risk of hotspots. These will provide a strong basis for future development regarding the technologies of tDCS and ultimately for enhanced therapeutic outcomes and broader

clinical applicability.

It also indicates the value of embedding simulation and empirical measurement approaches. The dual methodology robustly strengthens the findings and leaves a replicable model for further research in neuromodulation. This grid electrode proved to be of great value in the current leakage investigation and gave new insights into how tDCS treatments can be further improved. It appeared that current distribution is dynamic and depends on both individual factors and properties of electrode materials.

This framework can further enable detailed investigation of electrode-skin interface properties, thus offering considerable potential to advance both research and clinical practices in tDCS.

5.3.1 Summary of key findings

This thesis showed the usefulness of grid electrodes for the detailed current mapping, especially hotspots, and the physiological influence on the distribution of current. Thus, these results confirm the preliminary insights of previous studies regarding multi-electrode configurations, adding new evidence by in vivo experiments in practice [18].

So the key findings are heterogeneity of the current distribution at the measurement points, individual differences among the participants, and effectiveness of the grid electrode in locating hotspot areas as well as that the designed measurement setup performed well and can be used in future research when examining new electrode models.

5.3.2 Success of the study in relation to objectives

The thesis successfully achieved its aim of developing the verification measurement setup, validating it and designing grid electrode and considering its limitations, which could be used in optimizing tDCS treatments and further development of electrode materials. The findings align with the broader goal of optimizing tDCS treatments and electrode designs as in previous studies [15] [53].

This thesis provides the basis for future research for example with closed-loop tDCS systems, where current targeting may be updated in real time based on biosignal feedback. Such systems would improve the precision and safety of tDCS, enabling Therapeutic interventions can be more personalized and effective, but this still requires further research, since such a closed-loop system does demand the ability to measure precisely and have deep insight into the behavior of the current distribution.

5.3.3 Recommendations and future outlook

Future studies can be performed toward real-time optimizing systems that would automatically adjust to the physiological characteristics of the user in real time. Reliable current measurement during development could enable this integration of future personalized treatments with a system that can dynamically adjust itself through closed-loop feedback based on biosignal monitoring. This naturally involves designing

sophisticated electrode configurations that are capable of real-time monitoring and stimulation parameter adjustments. These include multi-input models in which each segment of the grid is provided with precisely controlled current, and its distribution is measured independently for high-resolution feedback. High-resolution, detailed tissue-specific properties will further enhance simulation models and allow for the accurate prediction of pathways of current and inform design improvements. This would open broader clinical applications where advanced current distribution systems could improve therapeutic outcomes in personalized treatments.

A more inclusive demographic approach is also required, with a wide range of user groups, to widen the generalization of results. Future studies involving patient groups and using advanced materials are needed to study [72]. Research regarding electrode materials and geometries should be directed toward achieving a more homogeneous distribution of current.

Further refinements in simulation models should be done to include, for example, detailed tissue properties of skin, skull, and brain conductivity. This kind of model should already make reliable predictions of how the current flows into the brain tissue. In the long run, a theoretical framework has to be developed that combines measurement data with simulation results, enabling the prediction of the current distribution within the whole brain. This probably will need new methodologies beyond those tried here.

These systems can utilize biosignal measures to dynamically adjust stimulation amplitude and targeting, personalized to a particular patient. This can dramatically enhance efficacy while minimizing discomfort. Once reliable current measurement technologies are established in development settings, sophisticated, personalized closed-loop tDCS systems could emerge that would provide both highly accurate current targeting and real-time adaptability to biosignal inputs, driving major advances in neuromodulation therapies.

This thesis lays a very good groundwork for the advancement of tDCS treatments and electrode technology in improving the accuracy and control of current distribution on skin-electrode interface of these therapies.

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A Measurement data

Mean Values (mA) (participant $\times$ point)

Participant	Point 1	Point 2	Point 3	Point 4
1	0.0381	0.0934	0.1015	0.0830
2	-0.0765	-0.0766	0.1288	0.1278
3	0.1628	0.0702	0.2149	0.1633
4	-0.0006	0.0022	-0.0018	0.0077
5	-0.0136	-0.0125	-0.0033	-0.0143
6	0.0111	0.1078	0.0065	0.0549
7	0.0783	0.0402	0.1263	0.0858
8	-0.0110	0.0228	0.0034	-0.0077
9	-0.0341	0.0006	0.0079	-0.0078
10	-0.0042	-0.0041	0.0005	0.0045
11	-0.1175	0.0349	0.0320	0.0050
12	-0.0645	-0.0064	-0.0030	-0.0070
13	-0.0078	-0.0079	-0.0087	-0.0087
14	0.0338	-0.1273	0.0270	0.2241
15	-0.0029	-0.0024	-0.0024	-0.0017
16	0.0766	0.0517	0.0497	0.1510
17	-0.0015	-0.0046	-0.0129	0.0024
18	-0.1283	0.0914	-0.0344	-0.0318
19	0.2226	0.0972	0.0875	0.4409
20	0.0819	0.0901	0.1183	0.0992
21	0.0201	0.0187	0.0362	0.0174
22	-0.0137	-0.0143	-0.0139	-0.0148
23	-0.0212	-0.0216	-0.0215	-0.0228

Deviations (participant $\times$ point)

Participant	Point 1	Point 2	Point 3	Point 4
1	0.0676	0.0826	0.0914	0.0731
2	0.0611	0.0612	0.0779	0.0781
3	0.1700	0.1111	0.2566	0.2123
4	0.0455	0.0394	0.0268	0.0593
5	0.0661	0.0651	0.0344	0.0652
6	0.0985	0.1198	0.0274	0.0899
7	0.0830	0.0397	0.1266	0.0869
8	0.0630	0.0971	0.0400	0.0722
9	0.0802	0.0798	0.0885	0.0674
10	0.0492	0.0461	0.0288	0.0704
11	0.0978	0.1279	0.1229	0.0896
12	0.0549	0.0340	0.0449	0.0338
13	0.0209	0.0214	0.0214	0.0214
14	0.0190	0.0336	0.0170	0.0440
15	0.0110	0.0111	0.0109	0.0075
16	0.1490	0.1391	0.1379	0.2064
17	0.0378	0.0276	0.0360	0.0512
18	0.1407	0.2011	0.0439	0.0590
19	0.0840	0.1436	0.1419	0.0455
20	0.0745	0.0834	0.0943	0.1128
21	0.0708	0.0695	0.0982	0.0718
22	0.0700	0.0698	0.0695	0.0704
23	0.0668	0.0670	0.0670	0.0677

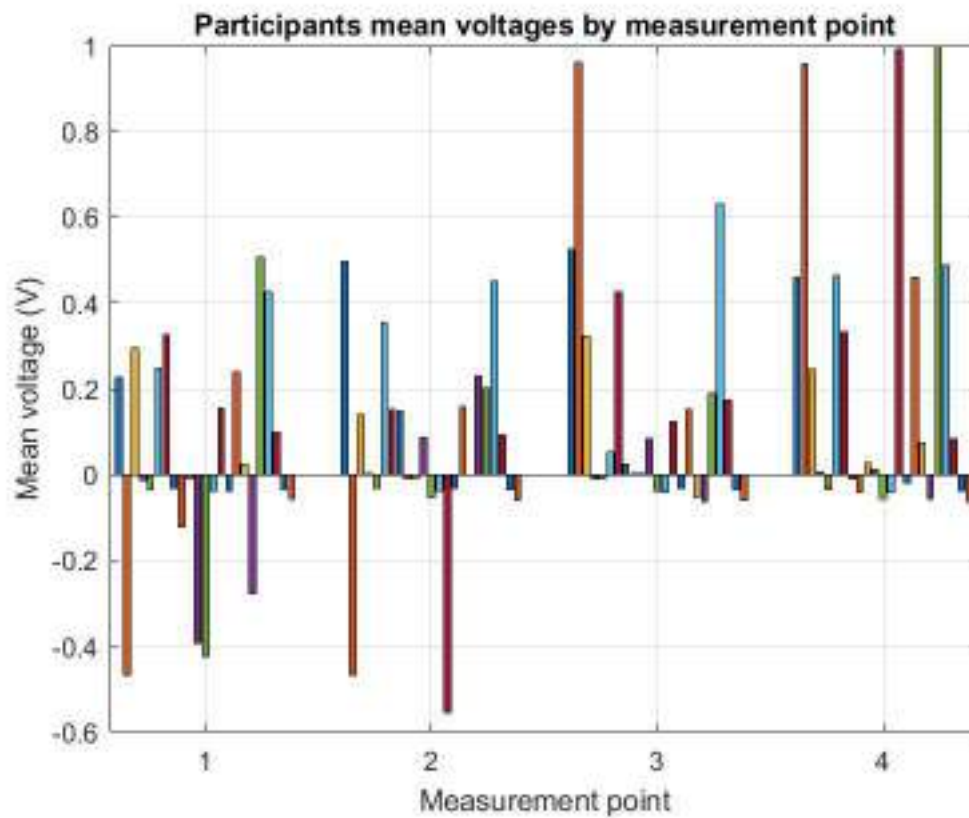


Figure A1: Participant mean voltages (V) by measurement point. Each color corresponds one participant.

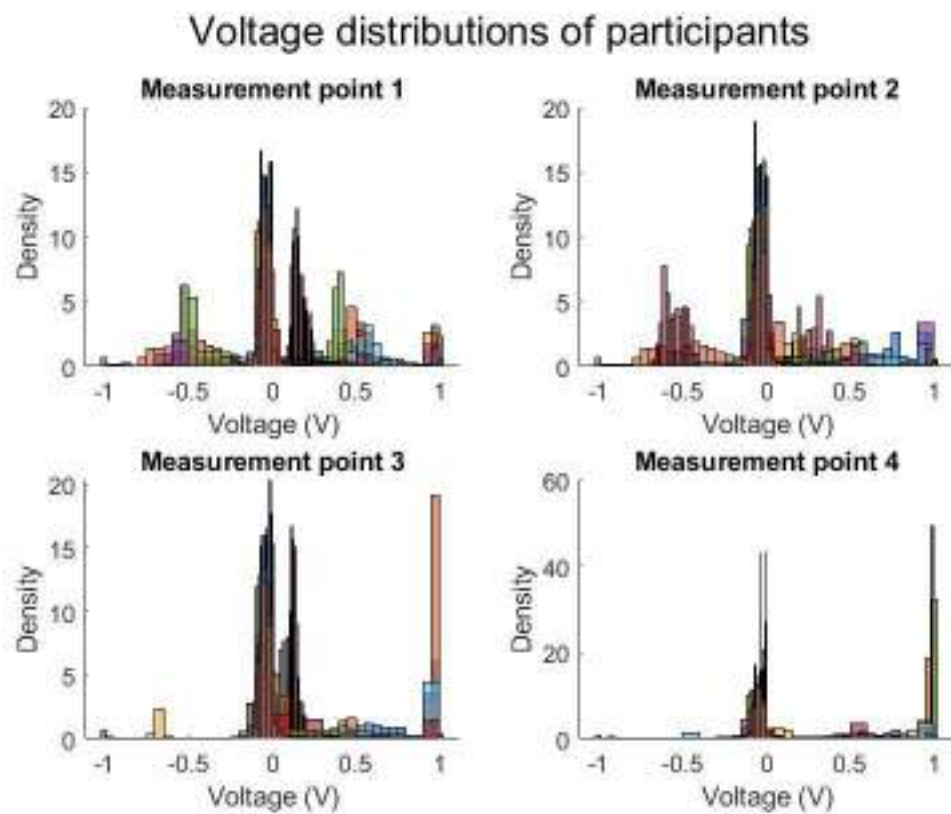


Figure A2: Voltage distributions of participants. Each color represents one participant. Note the different scale in measurement point 4 vs. others.

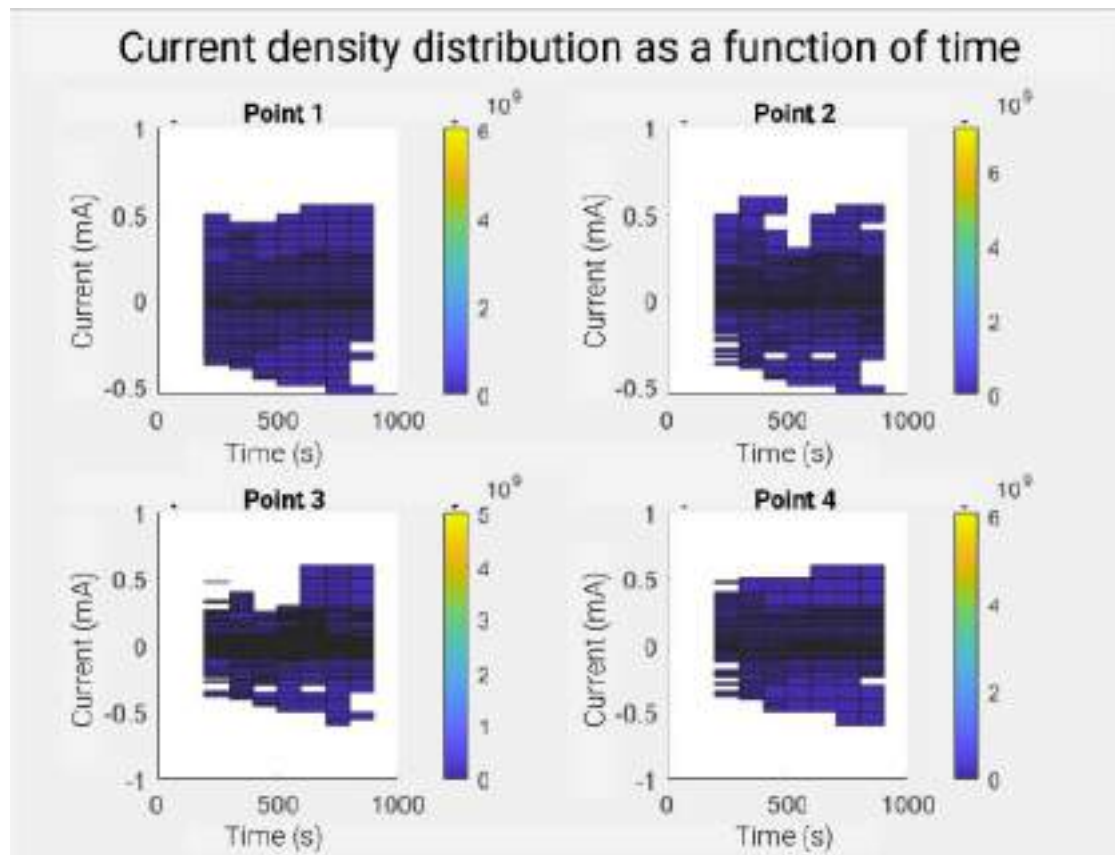


Figure A3: Current density distribution as a function of time by measurement points.

B Detailed experimental setups

Description of Experimental Setup

The experimental setup was made to provide as precise as possible and reproducible analysis of the tDCS electrode-skin interface current distribution. The measurement system consisted of the following:

Measuring Devices and Software:

- tDCS Source: The Sooma tDCS device delivered a 2mA direct current.
- Data Acquisition: Micro 1401 for converting analog signals into digital form.
- Amplifiers and Filters: Digitimer D440 instrument amplifier with accompanying filter.
- Software: Spike 2 was used in real time data acquisition, MATLAB was used for signal processing and visualization of data.

System Configuration:

- Grid-style segmented anode, Sooma's electrode cup cathode and Dätwyler long brush dry comb electrodes were used.

- A detailed wiring schematic was used to attach the electrodes to the instrumentation, so that at all times the signal interference was at a minimum.

Measuring Protocols

Measurement Procedure:

The participants signed consent form and each of them received one or two 15-minute sessions of tDCS (2mA) over the left DLPFC. They stayed still during measurement in order to avoid artifacts from the movement. Resistance was measured before, after and during the 15min session.

Calibration:

Pre calibration of all devices was effected with regards to standardized measurement before the measuring in order to ensure accuracy.

Handling Artefact:

- Noise reduction includes shielded cables and earth ground for all equipment.
- D440 was isolated.
- For moving artefacts, the fixing of electrodes is to be securely taped and subjects were asked not to move during measurements.
- High- and low-pass filtering was used.

Scheme and electrode placement

The color-coded wiring was detailed in the schematics, with electrode placement included to make setup replication clear (measurement points in Fig. B1).

The grid electrode array was placed in a configuration approximating a square on the scalp. Measurement points were positioned as follows:

- Points 1 and 4: Located most anteriorly, near the frontal region of the scalp.
- Points 2 and 3: Positioned posteriorly, closer to the vertex (parietal region).
- Points 1 and 2 were most distally and 3 and 4 were medially positioned.

These schematics also illustrated the flow of current from the tDCS device through the electrodes and into the participant's skin, as well as the routing of signals to the data acquisition module.

The present measurement electrode placement allows for the thorough examination of current distribution across both medial and lateral regions of the scalp. Anatomical placement was performed to cover a wide range of scalp topography for assessing electrode performance in different conditions.

C Additional materials



Figure B1: Grid-electrode used as an anode with measurement points numbered.

Electrode Size (radius)	Field	99.9%	75.0% Focality (mm³)	50.0% Focality (mm³)
2 cm, 5 cm	E (V/m)	0.482	12,400	75,200
	magnE (V/m)	0.482	12,400	75,200
	magnJ (A/m ²)	0.194	3,640	13,400
1 cm, 5 cm	E (V/m)	0.497	12,200	73,700
	magnE (V/m)	0.497	12,200	73,700
	magnJ (A/m ²)	0.200	3,610	13,300
1 cm, 6 cm	E (V/m)	0.485	11,800	73,800
	magnE (V/m)	0.485	11,800	73,800
	magnJ (A/m ²)	0.194	3,630	13,300
2 cm, 6 cm	E (V/m)	0.469	12,200	75,700
	magnE (V/m)	0.469	12,200	75,700
	magnJ (A/m ²)	0.188	3,650	13,400
1 cm, 3 cm	E (V/m)	0.512	11,600	69,200
	magnE (V/m)	0.512	11,600	69,200
	magnJ (A/m ²)	0.203	3,680	13,700
1 cm, 1 cm	E (V/m)	0.505	11,200	66,400
	magnE (V/m)	0.505	11,200	66,400
	magnJ (A/m ²)	0.198	3,680	13,700
3 cm, 3 cm	E (V/m)	0.499	11,900	71,600
	magnE (V/m)	0.499	11,900	71,600
	magnJ (A/m ²)	0.198	3,580	13,500

Table C1: Summary of SimNIBS Simulation Results for Different Electrode Configurations

Table C2: Field Percentiles for Different Anode and Cathode Configurations

Configuration	Field	99.9%	99.0%	95.0%
A 4×1 cm, C 6cm	E	4.52e-01 V/m	3.59e-01 V/m	2.74e-01 V/m
	magnE	4.52e-01 V/m	3.59e-01 V/m	2.74e-01 V/m
	magnJ	1.24e-01 A/m ²	9.88e-02 A/m ²	7.54e-02 A/m ²
A 4×1 cm, C 5cm	E	4.69e-01 V/m	3.69e-01 V/m	2.82e-01 V/m
	magnE	4.69e-01 V/m	3.69e-01 V/m	2.82e-01 V/m
	conductivity	2.75e-01	2.75e-01	2.75e-01
	magnJ	1.29e-01 A/m ²	1.02e-01 A/m ²	7.75e-02 A/m ²
A 4×1 cm, C 3cm	E	4.94e-01 V/m	3.85e-01 V/m	2.91e-01 V/m
	magnE	4.94e-01 V/m	3.85e-01 V/m	2.91e-01 V/m
	magnJ	1.36e-01 A/m ²	1.06e-01 A/m ²	7.99e-02 A/m ²
A 4×0.5 cm, C 6cm	E	4.75e-01 V/m	3.76e-01 V/m	2.85e-01 V/m
	magnE	4.75e-01 V/m	3.76e-01 V/m	2.85e-01 V/m
	magnJ	1.31e-01 A/m ²	1.03e-01 A/m ²	7.85e-02 A/m ²
A 4×0.5 cm, C 3cm	E	5.10e-01 V/m	4.00e-01 V/m	3.02e-01 V/m
	magnE	5.10e-01 V/m	4.00e-01 V/m	3.02e-01 V/m
	magnJ	1.40e-01 A/m ²	1.10e-01 A/m ²	8.29e-02 A/m ²
A 5×1 cm, C 6cm	E	4.64e-01 V/m	3.68e-01 V/m	2.81e-01 V/m
	magnE	4.64e-01 V/m	3.68e-01 V/m	2.81e-01 V/m
	magnJ	1.28e-01 A/m ²	1.01e-01 A/m ²	7.73e-02 A/m ²
A 5×1 cm, C 3cm	E	4.99e-01 V/m	3.92e-01 V/m	2.97e-01 V/m
	magnE	4.99e-01 V/m	3.92e-01 V/m	2.97e-01 V/m
	magnJ	1.37e-01 A/m ²	1.08e-01 A/m ²	8.17e-02 A/m ²
A 5×1.5 cm, C 6cm	E	4.47e-01 V/m	3.55e-01 V/m	2.72e-01 V/m
	magnE	4.47e-01 V/m	3.55e-01 V/m	2.72e-01 V/m
	magnJ	1.23e-01 A/m ²	9.77e-02 A/m ²	7.48e-02 A/m ²

Table C3: Field Focality for Different Anode and Cathode Configurations

Configuration	Field	75.0%	50.0%
Anode $4 \times 1\text{cm}$, Cathode 6cm	E	$1.08\text{e}+04 \text{ mm}^3$	$7.31\text{e}+04 \text{ mm}^3$
	magnE	$1.08\text{e}+04 \text{ mm}^3$	$7.31\text{e}+04 \text{ mm}^3$
	magnJ	$1.08\text{e}+04 \text{ mm}^3$	$7.31\text{e}+04 \text{ mm}^3$
Anode $4 \times 1\text{cm}$, Cathode 5cm	E	$1.03\text{e}+04 \text{ mm}^3$	$6.99\text{e}+04 \text{ mm}^3$
	magnE	$1.03\text{e}+04 \text{ mm}^3$	$6.99\text{e}+04 \text{ mm}^3$
	conductivity	$7.23\text{e}+05 \text{ mm}^3$	$7.23\text{e}+05 \text{ mm}^3$
	magnJ	$1.03\text{e}+04 \text{ mm}^3$	$6.99\text{e}+04 \text{ mm}^3$
Anode $4 \times 1\text{cm}$, Cathode 3cm	E	$9.46\text{e}+03 \text{ mm}^3$	$6.38\text{e}+04 \text{ mm}^3$
	magnE	$9.46\text{e}+03 \text{ mm}^3$	$6.38\text{e}+04 \text{ mm}^3$
	magnJ	$9.46\text{e}+03 \text{ mm}^3$	$6.38\text{e}+04 \text{ mm}^3$
Anode $4 \times 0.5\text{cm}$, Cathode 6cm	E	$1.04\text{e}+04 \text{ mm}^3$	$7.04\text{e}+04 \text{ mm}^3$
	magnE	$1.04\text{e}+04 \text{ mm}^3$	$7.04\text{e}+04 \text{ mm}^3$
	magnJ	$1.04\text{e}+04 \text{ mm}^3$	$7.04\text{e}+04 \text{ mm}^3$
Anode $4 \times 0.5\text{cm}$, Cathode 3cm	E	$9.88\text{e}+03 \text{ mm}^3$	$6.49\text{e}+04 \text{ mm}^3$
	magnE	$9.88\text{e}+03 \text{ mm}^3$	$6.49\text{e}+04 \text{ mm}^3$
	magnJ	$9.88\text{e}+03 \text{ mm}^3$	$6.49\text{e}+04 \text{ mm}^3$
Anode $5 \times 1\text{cm}$, Cathode 6cm	E	$1.06\text{e}+04 \text{ mm}^3$	$7.14\text{e}+04 \text{ mm}^3$
	magnE	$1.06\text{e}+04 \text{ mm}^3$	$7.14\text{e}+04 \text{ mm}^3$
	magnJ	$1.06\text{e}+04 \text{ mm}^3$	$7.14\text{e}+04 \text{ mm}^3$
Anode $5 \times 1\text{cm}$, Cathode 3cm	E	$9.68\text{e}+03 \text{ mm}^3$	$6.52\text{e}+04 \text{ mm}^3$
	magnE	$9.68\text{e}+03 \text{ mm}^3$	$6.52\text{e}+04 \text{ mm}^3$
	magnJ	$9.68\text{e}+03 \text{ mm}^3$	$6.52\text{e}+04 \text{ mm}^3$
Anode $5 \times 1.5\text{cm}$, Cathode 6cm	E	$1.10\text{e}+04 \text{ mm}^3$	$7.40\text{e}+04 \text{ mm}^3$
	magnE	$1.10\text{e}+04 \text{ mm}^3$	$7.40\text{e}+04 \text{ mm}^3$
	magnJ	$1.10\text{e}+04 \text{ mm}^3$	$7.40\text{e}+04 \text{ mm}^3$

Table C4: Simulation results with different electrode and gel thicknesses

Blocks	El.+G. (mm)	Field	99.9%	99.0%	95.0%	Focality 75%	Focality 50%
4	1 + 5	E	4.53e-01	3.57e-01	2.71e-01	1.04e+04	6.82e+04
		magnJ	1.25e-01	9.83e-02	7.44e-02	1.04e+04	6.82e+04
	3 + 3	E	4.54e-01	3.58e-01	2.71e-01	1.03e+04	6.83e+04
		magnJ	1.25e-01	9.84e-02	7.46e-02	1.03e+04	6.83e+04
	4 + 4	E	4.54e-01	3.58e-01	2.71e-01	1.03e+04	6.81e+04
		magnJ	1.25e-01	9.85e-02	7.46e-02	1.03e+04	6.81e+04
5	1 + 5	E	4.42e-01	3.46e-01	2.64e-01	9.78e+03	6.79e+04
		magnJ	1.22e-01	9.53e-02	7.25e-02	9.78e+03	6.79e+04
	3 + 3	E	4.43e-01	3.47e-01	2.64e-01	9.77e+03	6.80e+04
		magnJ	1.22e-01	9.53e-02	7.26e-02	9.77e+03	6.80e+04
	2 + 4	E	4.43e-01	3.47e-01	2.64e-01	9.76e+03	6.78e+04
		magnJ	1.22e-01	9.54e-02	7.26e-02	9.76e+03	6.78e+04
7	4 + 4	E	4.23e-01	3.31e-01	2.52e-01	9.66e+03	6.81e+04
		magnJ	1.16e-01	9.11e-02	6.94e-02	9.66e+03	6.81e+04