Evaluating the electrode measurement sensitivity of subdermal electroencephalography electrodes

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Take your time, think a lot, think of everything you got, for you will still be here tomorrow but your dreams may not

Cat Stevens
Acknowledgments

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Miguel Mendes
Abstract

Electroencephalography (EEG) is an invaluable neuroimaging modality. It is used to register the electrical activity of the brain with millisecond temporal resolution. The EEG recording depends on the electrode sensitivity distribution which has considerable variations among the electrode montages available. This thesis assesses the EEG measurement sensitivity distribution for both surface and subdermal electrodes.

A five-layered head model was constructed based on magnetic resonance data. We added 21 surface electrodes to the model according to the traditional 10-20 EEG system, and 5 × 5 subdermal electrode grids on the skull in seven reference locations: FZ, CZ, OZ, T3, T4, P3, and P4. The half-sensitivity volume (HSV) was measured for all the configurations studied. The surface leads concentrate the measurement in $1 \text{cm}^3$ volume. The sensitivity measurement is improved with the subdermal leads which can focus the measurement in regions ten times smaller. However, the improvement was registered only for the subdermal grids centred on CZ, T3 and T4 locations. This suggests that the electrode performance is highly dependent on thicknesses of the underlying matter, such as the skull and cerebrospinal fluid (CSF). Studying the electrode sensitivity provides a deeper knowledge regarding the EEG electrode montages which leads to clinical improvements in the diagnostics of brain functionality.

Keywords

Electroencephalography (EEG), half sensitivity volume (HSV), lead field, subdermal electrodes
Resumo

O electroencefalograma (EEG) é uma técnica de neuroimagiologia de valor inestimável. O EEG é utilizado para registar a actividade eléctrica cerebral com uma resolução temporal na ordem do milisegundo. O registo electroencefalográfico depende da distribuição de sensibilidade dos eléctrodos, a qual apresenta diferenças consideráveis entre as configurações de eléctrodos existentes. Este trabalho estima a medida de distribuição de sensibilidade do EEG, tanto para eléctrodos de superfície como para eléctrodos subdermais.

A partir de imagens de ressonância magnética construiu-se um modelo da cabeça humana com cinco tecidos. Adicionaram-se 21 eléctrodos de superfície de acordo com o sistema tradicional 10-20, e grelhas de $5 \times 5$ eléctrodos subdermais no crânio em sete posições de referência: $F_Z, C_Z, O_Z, T_3, T_4, P_3,$ e $P_4$. O half-sensitivity volume (HSV) foi medido para todas as configurações estudadas. A medição dos eléctrodos de superfície concentra-se em volumes de $1 \text{ cm}^3$. A medida de sensibilidade apresenta melhorias com os eléctrodos subdermais, cuja medição de sensibilidade chega a concentrar-se em volumes dez vezes menores. Contudo, esta melhoria apenas se observou para as grelhas subdermais situadas em $C_Z, T_3$ e $T_4$. Tal facto sugere que o desempenho dos eléctrodos depende bastante da espessura dos tecidos subjacentes, tais como o crânio e o líquido cefalorraquidiano. O estudo da sensibilidade dos eléctrodos proporciona conhecimento mais aprofundado das configurações de eléctrodos de EEG, o que conduz a melhorias a nível clínico e de diagnóstico do funcionamento cerebral.

Palavras Chave

Electroencefalograma (EEG), half sensitivity volume (HSV), lead field, eléctrodos subdermais
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Abbreviations

3D  three-dimensional

aFDM  anisotropic finite difference method

AMG  algebraic multigrid

BEM  boundary element method

BIC  Brain Imaging Center, McConnell

CG  conjugate gradient

CSF  cerebrospinal fluid

CT  computed tomography

DW-MRI  diffusion weighted magnetic resonance images

EEG  electroencephalogram

EIT  electrical impedance tomography

FDM  finite difference method

FEM  finite element method

fMRI  functional magnetic resonance imaging

FSV  fifth sensitivity volume

HSV  half sensitivity volume

iFDM  isotropic finite difference method

MEG  magnetoencephalogram

MRI  magnetic resonance imaging

PCG  preconditioned conjugate gradient

SNR  signal to noise ratio

SOR  successive over-relaxation
List of Symbols

$A_1$ auricular 1
$A_2$ auricular 2
$C_3$ central 3
$C_4$ central 4
$C_5$ central 5
$C_6$ central 6
$F_3$ frontal 3
$F_4$ frontal 4
$F_7$ frontal 7
$F_8$ frontal 8
$F_{p1}$ frontopolar 1
$F_{p2}$ frontopolar 2
$F_{Z}$ frontal zero
$O_1$ occipital 1
$O_2$ occipital 2
$O_{Z}$ occipital zero
$P_3$ parietal 3
$P_4$ parietal 4
$T_3$ temporal 3
$T_4$ temporal 4
$T_5$ temporal 5
$T_6$ temporal 6
Γ_Ω  surface of the head
Φ_{LE}  lead field electric potential
ε  electrical permittivity
σ  electric conductivity tensor
Ω  volume of the head
c  lead vector
k  finite number of locations within the volume conductor
p  location inside the volume conductor
J  total current density
J^i  volume source current density
J_{LE}  lead field current density
n  vector normal to the surface
V_L  lead voltage
V_{LE}  reciprocal lead voltage
υ  volume of the conductor model
μ  magnetic permeability
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1.1 Motivation and Goals

The human brain is a complex organ and highly susceptible to many types of damages and diseases. The interest of the human being in treating and healing neurological disorders started in early times. The first evidences of cranial surgery date back thousand of years ago [1]. In 1875, Richard Caton, a scientist from Liverpool, England, recorded electrical signals from living brains in animal subjects [2], but the first human electroencephalogram (EEG) was recorded only in 1929 by the German psychiatrist Hans Berger [3, 4]. Since then, neuroscientists have used the EEG to study the brain rhythms, to investigate neurological diseases and to learn how the mental activities affect the brain activity. In 1947 the American EEG Society was founded and the First International EEG Congress was held in London, United Kingdom [2]. During the following decades, the work on EEG rapidly expanded: microelectrodes with diameters shorter than 3 \( \mu m \) were invented, and the first depth electroencephalography of a human was obtained using intracerebral electrodes by Mayer and Hayne in 1948 [2, 5]. The electrocorticogram (ECoG) became a reality due to the development of very delicate needle-type electrodes that measure the brain activity directly from the cerebral cortex.

The EEG and the magnetoencephalogram (MEG) detect the neuroelectrical activity with millisecond temporal resolution [6]. However, both EEG and MEG need to be combined with other neuroimaging techniques to get sufficient spatial resolution, which is needed to correlate the neuroelectrical events with brain structures [7]. Functional magnetic resonance imaging (fMRI), positron emission tomography (PET), and single photon emission computed tomography (SPECT) offer the improved spatial resolution but worse temporal resolution than the EEG [6, 7].

Nowadays, EEGs are recorded invasively and noninvasively but the knowledge regarding the measurement sensitivity is limited. Researchers have used volume conductor head models to simulate the sensitivity distributions of the EEG electrode configurations [8–12]. They have shown the importance of multilayered realistic models, the benefits of subdermal electrode implantation, and they also described the impact of modelling the EEG in diagnostic and management of neurological disorders like epilepsy. Nevertheless, these studies barely compare the measurement sensitivity among the existing electrode configurations.

The main goal of this thesis is to evaluate the sensitivity measurement of subdermal EEG electrodes. Seven subdermal electrode grids, placed on the skull surface of a five-layered realistic head model, are studied and compared to the traditional electrodes located on the scalp surface. A secondary objective is to identify the model characteristics, such as the thickness and electrical conductivity of the underlying tissues, that explain the variations between the performance of the two types of electrodes.

1.2 Contribution of Thesis

The successful completion of this thesis provides researchers and neurophysiologists useful information regarding the EEG. Firstly, the results obtained in this work confirm the relationship between the measurement sensitivity distributions and the geometry of the head tissues beneath the mea-
surement sites, that were previously reported in the literature. Secondly, comparing the sensitivity distributions provides neurophysiologists information on how to optimize the recording according to the location of the subdermal grid. The ultimate contribution is the improvement of the EEG recordings on individuals with neurological disorders, which will help clinicians to localize neuroelectric sources and better evaluate brain pathologies.

1.3 Thesis Outline

The Chapter 2 reviews the theoretical concepts required to fully understand how to evaluate the sensitivity of EEG electrodes using volume conductor models. In Chapter 3, the methodology involved on the construction of the 3D head model is described, the electrode montages studied are presented and the sensitivity measurement is explained. The Chapter 4 contains the results of the sensitivity measurements for both surface and subdermal leads, which are discussed in the following Chapter 5. The Chapter 6 includes the last considerations regarding this project, as well as possible new approaches for the related future work.
Background

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This chapter provides the background knowledge required to study the sensitivity of EEG electrodes. The first section reviews the neuroanatomy and neurophysiology and introduces the electric dipole model. The second section briefly summarizes the EEG characteristics, emphasizing the improvements obtained due to the newer invasive techniques, and also describes the traditional surface EEG montage. The third section focuses on head volume conductors and explains the reciprocity theorem applied to the forward problem. It also reviews the conductivity and anisotropy of the head tissues and provides a synopsis of the numerical methods and solvers available.

2.1 Brain Anatomy and Neurophysiology

Anatomically, the human brain is divided in four major regions: brain stem, cerebellum, diencephalon and cerebrum. Each region is specialized in different functions. The cerebrum is the largest part and consists of an outer cerebral cortex, an internal region of cerebral white matter, and gray matter nuclei deep within the white matter. The cerebral cortex, whose thickness ranges from 2 to 4 mm \[13\], is formed by convolutions and fissures. The convolutions are named gyri and the superficial fissures are named sulci. The longitudinal fissure divides the cerebrum into right and left hemispheres that are internally connected through the corpus callosum (a broad band of white matter). Each cerebral hemisphere is further subdivided into four lobes (Fig. 2.1) that are named after the bones (Fig. C.1) that cover them: frontal, parietal, temporal, and occipital. The central sulcus, the lateral cerebral sulcus and the parietal-occipital sulcus separate the frontal lobe from the parietal lobe, the frontal lobe from the temporal lobes, and the parietal from the occipital lobes, respectively. The insula is the inner lobe that lies within the lateral cerebral sulcus. \[13\]

![Figure 2.1: Right lateral view of the cerebrum. Reproduced from \[13\].](image)

The human brain contains about $10^{11}$ neurons and $10^{13}$ glial cells known as neuroglia \[13\]. Although they may differ in size and shape, all the neurons possess the same anatomical subdivision (Fig. 2.2). The cell body or soma is the core of the cell and contains the nucleus, the dendrites are branching projections of the soma and are specialized in receiving inputs from other neurons, and the
The axon is responsible for the electric impulse transmission to the following neurons [14]. Each neuron is further connected to approximately $10^4$ other neurons [2].

The electric impulse flows from one neuron to the next through a specialized interface named synapse. About $10^{15}$ synapses exist in the human brain [13]. A synapse consists of a cleft, between a presynaptic and a postsynaptic neuron, where the impulse transmission depends upon chemicals called neurotransmitters that interact with the intracellular environment. In the resting state, the typical polarization of the intracellular space varies between $-60$ and $-70 \, \text{mV}$ [5]. This potential is altered when an action potential reaches the cell. When an action potential travels along one neuron that ends in an excitatory synapse, an excitatory post-synaptic potential (EPSP) occurs in the following neuron and the action potential is propagated if its intracellular compartment reaches the $-55$ to $-50 \, \text{mV}$ threshold [2]. On the other side, if the neuron ends in an inhibitory synapse, then it will result in hyperpolarization, corresponding to an inhibitory postsynaptic potential (IPSP). The restraint of the nervous function is only possible due to the inhibitory receptors [14]. Figure 2.3 describes both EPSP and IPSP cases.

In the two largest portions of the brain, cerebrum and cerebellum, the gray matter is located externally while the white matter is an internal tissue. The gray matter is considered the brain source region where the neuroelectric activity is generated [11], and it mainly consists of neuronal cell bodies, dendrites, unmyelinated axons, axon terminals, and neuroglia. The white matter is composed primarily of myelinated axons that propagate the electric impulse.

The cortical gray matter is organized in a six layered structure with a thickness that ranges between 1 mm and 4 mm [15]. The pyramidal cells are the neurons responsible for generating the EEG signal. They contain both apical and basal dendrites. The former are located in the two outermost layers of the cortical gray matter, while the latter are located in deeper layers. The apical dendrites are orthogonally oriented to the brain surface, defining the electrical flow direction. Further, the pyramidal cells are capable of producing strong current dipoles, whereas weaker currents result from the other neurons found in the cortical gray matter. [12, 15, 16]

Further readings on the brain anatomy and physiology may be found in [5, 13, 14].
Figure 2.3: Action potentials in the excitatory and inhibitory presynaptic fibre respectively lead to EPSP and IPSP in the postsynaptic neuron. Reproduced from [5].

2.1.1 The electric dipole model

To produce a measurable EEG signal, a considerable amount of pyramidal neurons needs to be synchronously active [17]. Since these cells are arranged orthogonally to the cortical surface, the electric superposition results in the amplification of the potential distribution. Therefore, when several neurons are active in a restricted area of the cortical gray matter, the resulting synchronous electrical activity may be macroscopically represented by an equivalent current dipole [18]. The current dipole has been widely used as a source model in both forward and inverse applications, not only in electroencephalography [19], but also in magnetoencephalography studies [20].

The equivalent electric dipole consists of a current source and a current sink, with opposite current strength, located infinitesimally close to each other [18]. The direction of the current is defined by the direction along which positive charges are transported. Thus, at the level of the synapse, a positive inward current represents an excitatory post-synaptic current (EPSC), while a negative one describes an inhibitory post-synaptic current (IPSC) [21]. Consequently, in the extracellular environment, an active current sink is caused by an EPSC and an active current source by an IPSC. As a result, a dipole configuration rises due to the source and the sink created by EPSC and IPSC, respectively. From the macroscopic point of view, the activation of a set of parallel neurons is capable of creating a dipole layer, but synchronous activation of the neuronal population is required [21].

Besides being used to study the EEG source localization problem [22], the source and sink model
has been also used as a neuroelectric generator to solve the EEG forward problem using, not only a finite element method model [9, 23], but also a finite difference method model [11]. With the dipole model, the sensitivity distributions on the cortex may be successfully computed using the forward problem.

2.2 Electroencephalography

In 1924, the electric activity of the human brain was recorded for the first time by the German psychiatrist Hans Berger who named the recording, the electroencephalogram [3]. The electroencephalogram or EEG is the clinical procedure that measures the brain electrical activity using electrodes placed on the forehead and scalp [13], and it results from the summed electrical activity of populations of cortical neurons, with a modest contribution of glial cells [21]. Regarding the metallic composition, the EEG electrodes are commonly manufactured using silver, silver-chloride, tin, gold, platinum or stainless steel [24–26]. Generally, the diameter of surface electrodes rounds 10 mm, while the shape of the electrode depends on the vendor [27]. Despite the high temporal resolution (in the millisecond range), the EEG has an insufficient spatial detail to correlate brain electrical events and anatomic structures [7]. To overcome this issue, the number of electrodes may be increased and the distortion, caused by skull, should be corrected.

The EEG not only allows the study of the normal brain functions, such as changes that occur during sleep, but also helps diagnosing brain disorders such as epilepsy, tumours, trauma, hematomas, metabolic abnormalities, sites of trauma, and degenerative diseases [13]. It is also an important tool for studying the temporal dynamics of the neuronal circuits [6] and allows to establish the different activation patterns of brain neurons: alpha, beta, theta and delta rhythms [13]. In the epilepsy world, the EEG is extremely helpful as diagnosis tool and plays a crucial role in the anatomical localization of the epileptogenic zone [4, 19, 28].

Nevertheless, the scalp electrodes are remote, detecting the summed activities of a large number of neurons which are synchronously electrically active [16]. Thus, these electrodes hardly identify the epileptic source with the accuracy needed to perform a surgical intervention. Although the EEG measurement is highly dependent on the electrode size – mainly on the cross section area [29] – the scalp electrodes detect potentials with only 100 µV of amplitude, which is considerable low when compared to the 1 – 2 mV potentials detected by electrodes placed on the brain surface [3].

The electrocorticogram (ECoG) is the technique used to record the electric activity of the brain by placing electrodes on cortical surface [21]. It can be performed using either needle-type electrodes or electrode grids or strips, and it has been shown very effective since the attenuation and nonlinearity effects induced by the skull are eliminated [2]. ECoG electrodes are widely used in epilepsy centers in patients who are diagnosed with drug-refractory partial epilepsy and need epilepsy surgery [30]. By placing the electrodes over suspected areas of epileptogenicity, long-term intracerebral recording of seizures is possible and, consequently, the delineation of the epileptogenic zone is improved. This invasive technique utilises depth electrodes that inserted surgically under stereotactric MRI guidance.
or subdural electrode strips or grids [4]. Due to its limited field-of-view, placing a electrode grid is a delicate intervention that requires priori estimation of the epileptogenic zone [31]. Therefore, the traditional EEG is required to estimate the epileptogenic brain lobe before the implantation of a ECoG grid [31].

One minimally invasive alternative for recording the brain activity is the subdermal or subcutaneous EEG measurement. In this new approach, the electrodes are implanted on the skull beneath the skin, fat, and muscles. Besides bypassing the artifact-prone skin, the subcutaneous implantation notably enhances the accuracy and specificity of EEG measurement when compared with the scalp measurement. Similarly to the ECoG, the subcutaneous measurements enable specific monitoring of small source volumes of the brain. [10, 32]

2.2.1 10-20 electrode system

The traditional EEG is recorded using 21 surface electrodes placed on the scalp according to the international 10-20 system (Fig. 2.4) [3]. This electrode system uses the nasion and the inion as anatomical references. The former corresponds to the delve at the top of the nose levelled with the eyes, and the latter is the bony lump at the base of the skull on the midline at the back of the head. From these points, the skull perimeters are measured in the transverse and median planes and the electrode locations are determined by dividing these perimeters into 10% and 20% intervals. Additionally, three more electrodes are placed on each side equidistant from the neighbouring points. Both locations and nomenclature of the electrodes are standardized by the American Electroencephalographic Society [3]. Optionally, electrodes may be introduced halfway between each of the traditional montage – the 10% system – in order to improve the accuracy and the spatial resolution [5, 12, 33].

![Figure 2.4: The sagittal (A) and the superior transverse (B) views of the international 10-20 electrode system. Reproduced from [3].](image-url)
The EEG is often recorded with high number of electrodes, like 64, 128, 256 or even 512, to obtain a finer spatial sampling of the electrical activity at the scalp and, thus, acquire high precision source locations [19]. Dense EEG systems are extremely sensitive to noise [33] due to the proximity of the adjacent electrodes that mainly measure the lead field current that flows within the skin [3]. Consequently, the EEG primarily detects electric sources that are radial to the scalp surface with sufficiently distant electrodes and tangential components when the leads are located near to each other (Fig. 2.5) [33].

Furthermore, there are two methods to measure the potential of the EEG electrodes. While bipolar electrodes register the potential difference between a pair of electrodes, the unipolar method records the potential of each electrode compared either to a reference electrode or to the average of all electrodes. [3]

**Figure 2.5:** The Sensitivity Distributions of EEG. (Left) An EEG setup measuring the tangential components of neuroelectrical activity, where each bipolar lead is located relatively close to each other. (Right) An EEG setup measuring the radial components of neuroelectric activity, where the measuring electrode is located far from the reference electrode. The arrows in both figures represent macrocolumns of cellular architecture not dipolar sources. Reproduced from [33].

### 2.3 Head Volume Conductors

The head model as a volume conductor is a key element to study the measurement principle of the EEG. The model parameters, such as geometry and conductivity, play an important role when measuring the sensitivity distributions [27, 33]. Complex models with realistic features provide more precise results, whereas simple models only generate theoretical results. Therefore, knowing the shape and thickness of the different biological tissues, as well as the realistic conductivity values of the tissues is mandatory to successfully evaluate the sensitivity distributions.

#### 2.3.1 Forward Problem

The forward problem consists of finding the effects given a source information. When applied to the study of the brain activity, the head volume conductor is considered to be the head model, the EEG measurement is the data and the neuroelectric sources are the model parameters [27]. Consequently,
the forward problem consists of finding the electrostatic potentials within the head volume conductor when a set of neuroelectric sources is given. The potential distribution is computed by solving the Poisson’s equation (Eq. 2.1), where $\sigma$ is the electrical conductivity tensor, $\Phi$ is the electrical potential, $J^i$ is the current source distribution and $\Omega$ is the volume of the head [34]. Additionally, it is common to define Neumann boundary conditions equal to zero (Eq. 2.2) on the outer layer of the model, the scalp surface $\Gamma_\Omega$, where $n$ is the vector normal to this surface [34]. As an alternative, the Dirichlet boundary condition can be specified by fixing the electric potential on the outer surface instead of its derivative [35].

$$\nabla \cdot (\sigma \nabla \Phi) = \nabla \cdot J^i \ (in \ \Omega) \quad (2.1)$$

$$\sigma (\nabla \Phi) \cdot n = 0 \ (on \ \Gamma_\Omega) \quad (2.2)$$

On the other side, the inverse problem is commonly used in neuroscience to estimate the internal current source that fits with a given potential distribution measured on the scalp. However, the inverse problem goes beyond the scope of this study since it is not needed to evaluate the electrode sensitivity.

Regarding the sensitivity measurement of EEG electrodes, there is an alternative to the forward problem that allows one to determine the sensitivity distributions, based on the reciprocity theorem. The reciprocal problem consists of injecting an unitary current on the electrode to measure its sensitivity. The following section explains the reciprocal problem by introducing the concept of lead field.

### 2.3.2 Lead Field and the Reciprocity Theorem

The reciprocity theorem, introduced into biophysical areas by Hermann von Helmholtz in 1853 [36], states that the sources and measurement locations may be exchanged without affecting the results. In 1969, Rush and Driscoll [37] adapted this theory to the EEG problem. To measure the sensitivity of EEG leads, they used the reciprocity theorem to calculate the potential and current density distributions within the brain. The lead field theory – that is based on the reciprocity theorem of Helmholtz – was extensively described by Malmivuo and Plonsey in 1995 [3].

First of all, the concept of lead field requires two measurement sites, i.e. a pair of electrodes. Then, each dipole source is characterized by a dipole source location $p$ inside the volume conductor and a lead vector $c$. The lead field $J_L$ is therefore, the field composed by all the lead vectors $c_k$ of the locations $p_k$ of the volume model. Since the voltage $V_k$ of each elementary dipole is given by the inner product between $c_k$ and $p_k$, the total lead voltage $V_L$ has, according to the principle of superposition, the contribution of all dipole elements (Eq 2.3).

$$V_L = \sum c_k \cdot p_k \quad (2.3)$$

One important property of the lead field, that results from the reciprocity theorem, is that the lead field $J_{LE}$ is exactly the same as the electric current field raised by introducing a reciprocal unitary
current to the lead. Consequently, the lead voltage $V_{LE}$ due to a volume source of $J^i$ is obtained by integrating the dot product between the lead field current density and the source density throughout the volume source with the conductivity tensor $\sigma$ (Eq. 2.4). Further, the lead field may be visualized either as a field of lead vectors or with lead field current flow lines. The lead vectors are tangents to the lead field current flow lines and their length is proportional to the density of the flow lines.

$$V_{LE} = \int \frac{1}{\sigma} \mathbf{J}_{LE} \cdot \mathbf{J}^i \, d\mathbf{v}$$ (2.4)

Concluding, the measurement of the sensitivity distribution within a volume conductor may be performed by feeding a reciprocal current to the lead. This is a major advantage since one single reciprocal calculation replaces all the $k$ forward calculations [15] [37].

### 2.3.3 Geometry

The geometry of the volume conductor is a parameter that highly influences the lead field within the model [9]. Despite the geometrical complexity of the head tissues, the first models of the human head consisted of three concentric spheres that represented the brain, skull and scalp [37]. Posteriorly, researchers realized the importance of including a fourth shell in the models to represent the cerebrospinal fluid [35] [38]. However, the spherical shape barely represents the geometry of the head tissues. In this way, investigations towards the development of head models more realistically shaped increased to improve the accuracy of neuroelectric problems, like the source localization [39], scalp potentials [40], and source imaging [6].

More recently, the segmentation of image slices from medical imaging modalities, such as computed tomography (CT) and Magnetic Resonance Imaging (MRI), has been used to obtain the realistic geometries of head tissues.

The McConnell Brain Imaging Center (BIC) from Montreal Neurological Institute, Canada, created a database from realistic simulated MRI volumes. Broche et al. [41], from McConnell BIC, developed a second version of the digital brain phantom in 2005. This newer version was built using mostly automated techniques and contains three new layers (vessels, dura matter and marrow) in addition to the layers included in the previous phantom: gray matter, white matter, cerebrospinal fluid, muscles, skull, skin and fat. Five of the tissues segmented by Broche et al. [41] are shown in Figure 2.6. The accurate segmentation resulted from the high spatial resolution of MRI and they have a great value in the study of imaging modalities, such as MRI, functional MRI, CT, positron emission tomography (PET) and single photon emission computed tomography (SPECT).

Segmentation of head tissues using structural imaging techniques, such as CT and MRI, leads to the construction of multi-layered models used as tools to study the electrode sensitivity in the EEG. While MRI provides better images of soft tissues (skin, gray matter and white matter), CT is more sensitive to hard tissues like the skull [42]. Moreover, T1-weighted MRI is well suited for the segmentation of soft tissues and tissue boundaries like outer skull and skin, but the classification of the skull is problematic. Currently, it is common to use T1 images to segment white matter, gray matter and CSF, while skull and scalp may be obtained from the proton-density images [16] [43].
The accurate segmentation of both CSF and skull tissues is mandatory since the geometry of these layers have a great influence on modelling the EEG [9, 44]. On one hand, the continuity of the CSF layer, between brain and skull compartments, is a parameter that should be included in a realistic model when measuring the sensitivity of subdermal electrodes [11, 45]. On the other hand, large inaccuracies in skull geometry might lead to errors of 20 mm on EEG source analysis [44].

![Figure 2.6: Five tissues obtained from MRI segmentation: CSF, GM, WM, Skin-Muscles and Skull. Reproduced from the website of the McConnell Brain Imaging Centre (BIC) of the Montreal Neurological Institute, McGill University.](image)

### 2.3.4 Tissue conductivity and anisotropy

The head tissues are electrically inhomogeneous, anisotropic, dispersive, and nonlinear [33]. They have different conductivities $\sigma$, permittivities $\epsilon$ and magnetic permeabilities $\mu$, and they present multi-layered structures where the value of these parameters highly depends on the direction [45]. Therefore, to create a realistic multilayered model, inhomogeneous and anisotropic properties need to be assigned to the tissues. Additionally, since in vivo or living in vitro and postmortem measurements show differences [46], the conditions under which the values are acquired must be described and preference should be given to in vivo measurements. For instance, the resistivity values may be obtained based on in vivo electrical impedance tomography (EIT), as performed by the team lead by Lopes da Silva [12]. The anisotropic conductivity profile can be derived from diffusion weighted magnetic resonance images (DW-MRI), even though it is not a straightforward procedure [47].

Anisotropicity is an important issue in forward problem as it influences the lead field and this is discussed in detail elsewhere [18, 22, 32, 35, 47-54]. In 2005, Hallez et al. [18] firstly studied the influence skull and white matter anisotropy in the EEG source localization using a spherical model. In the following years, the team lead by Hallez [22, 47] used more realistic models with anisotropic white
matter to study the same problem. In these studies it was shown that the anisotropy influences the EEG dipole source localization, not only in source position, but also in its direction. During the same years, Cook et al. [48, 50] solved both forward and inverse EEG problems and found out that the anisotropy of the tissues cannot be neglected from the head models in order not to compromise the accuracy of the scalp potentials or source location, respectively. The work developed by Wolters et al. [49] in 2006 may be resumed to three major findings. Firstly, the influence of white matter and skull anisotropy on the EEG forward problem was confirmed. Secondly, they discovered that the deeper a source lies and the more it is surrounded by anisotropic tissue, the larger is the influence of this anisotropy on the resulting electric and magnetic fields. Finally, the team realised the importance of the highly conducting cerebrospinal fluid compartment, underscoring the need for accurate modelling of this space. In 2008, Bashar et al. [52] also investigated the anisotropy of both white matter and skull. By dividing the conductivity of these tissues in the transverse and longitudinal components, the researchers realised that the anisotropy of white matter is mainly due to the inhomogeneous transversal conductivity, whereas on the skull the main anisotropy effect is caused by the radial inhomogeneous conductivity. The same result regarding the importance of the radial conductivity of the skull was obtained by Restrepo [53] in 2010.

Despite being an essential feature when modelling the head, the anisotropy is not always included in the models. Nevertheless, it is crucial to find the most suitable value for the isotropic conductivity of each head compartment. The two most discussed layers, and likely to have different reference conductivity values across the literature [32, 45, 46, 54, 55], are the CSF and skull. However, the six ultra thin cortical layers of the gray matter also contribute to the electrical attraction and shunting of lead field currents. Though, further investigation on the conductivity of the gray matter is still needed and recommended by Wendel et al. [45].

The skull is formed by a trilayered structure that contains a highly conducting layer sandwiched in between the two lower conducting skull layers. The middle porous layer – the diplöe – is formed by spongy bone, while the two layers around are composed of compact bone. The porous diplöe has a higher conductivity since more higher-conducting fluid plausibly perfuses this region than the denser outer and inner skull surfaces [46]. When studying the EEG, the skull is known due to its shunting behaviour, even though the lead field is channelled and reshaped in the middle porous layer [45]. Wendel et al. [32, 54] also correlated the ageing of skull and the consequent decrease of conductivity in the sensitivity measurement of both surface and subcutaneous bipolar EEG lead. According to the research, the sensitivity ratio between the surface and subdermal measurements, is smallest with the lowest skull resistivity, or highest conductivity. Consequently, since the bone ossification is completed after adolescence, this is when the conductivity of the skull reaches the highest value. Afterwards the conductivity starts to decrease and the EEG measurement loses sensitivity. Despite, considering the skin conductivity constant from adolescence onwards, further investigations on its influence on EEG leads are still needed. The overall conclusion was that the electrode sensitivity benefits from subdermal implantation since it provides more localised measurements. Furthermore, Malmivuo and Suihko found out that the more realistic is the resistivity of the skull, the more improved is the spatial
resolution of EEG when compared to the magnetoencephalogram (MEG) [55].

Finally, realistic conductivity values of the skull compartment were registered by Wendel and Malmivuo [46]. The researchers recommend a skull conductivity value of 0.0053 S/m, since it reflects the average skull conductivity, but they also indicate that 0.0063 S/m may be considered to include the younger population. The mentioned values were obtained by Hoekema et al. [58] by measuring the conductivity of temporarily removed skull fragments during epilepsy surgery. Comparing these live measurements with the post mortem values commonly reported on the literature, Wendel and Malmivuo concluded that the in vivo confirmation is advisable to decide the most realistic conductivity values for head tissues. In addition, they also suggested to include skull measurement recording locations in order to confirm the decreased conductivity near sutures and conductivity variations due to localized skull thickness variations.

On the other side, the CSF, which surrounds the brain, is often neglected either due to the difficulty to correctly segment this layer from magnetic resonance images or due to the insensitivity of CT images to soft tissues. As a result, the skull thickness is increased beyond the realistic value because it is considered the space in the model between the scalp and the brain, or, on the other side, the brain is exaggeratedly enlarged to compensate the absent CSF layer. Nevertheless, it is known that the CSF attracts and concentrates the lead field current, thus partially shunting the current flow to the brain and, as a result, the sensitivity distributions depend on the correct determination of the CSF boundaries. Wendel et al. [9, 45] have shown the importance of including the CSF compartment in multilayered models of the human head. The conductivity used to prove the need of including this compartment in head models was 1.79 S/m, which corresponds to the body temperature conductivity measured by Baumann et al. [57]. The value included in the model created by Subramaniyam et al. [11] to study the subdermal sensitivities of EEG electrodes was slightly higher, 1.82 S/m, but still pretty realistic when compared to the underestimated values measured at room temperature [57].

### 2.3.5 Numerical Methods

The bioelectric forward problem needs to be solved to find the current density distribution. Three numerical methods are used to address the forward problem by solving Poisson’s equation (Table 2.1). The boundary element method (BEM) [12, 58, 59] consists of computing the electric potential at the boundaries between homogeneous isotropic conducting regions, commonly considered the interfaces between tissues. The finite difference method (FDM) [15, 22] is useful when anisotropy needs to be included in the model, and the finite element method (FEM) [48, 49, 60] is known for its flexibility regarding the geometry and conductivity distribution of the volume conductor model.

The number of computational points depends on the method. While BEM only computes solutions for the points located in the boundaries between compartments, FDM and FEM need to calculate for all the points within the volume, leading to a larger number of computational points. On the other side, these two methods allow the determination of the potential at an arbitrary point through interpolation, which cannot be done with BEM. Secondly, the computational points of FDM lie fixed in the cube centers for the isotropic approach and at the cube corners for the anisotropic approach, whereas
BEM and FEM benefit from the freedom to choose the positions of the vertices or tetrahedrons, respectively. Consequently, using the same amount of nodes, FEM provides a better representation of irregular interfaces between compartments than FDM. The computational efficiency is highly dependent on the number of points. On one hand, when the number of compartments increases, more boundaries are defined leading to a large full system matrix in the BEM and, as a result, the numerical efficiency is compromised. On the other hand, the number of regions is transparent to FEM and FDM since it is possible to give each tetrahedron or cube a different conductivity. Furthermore, tessellation algorithms are required in the FEM and BEM in order to obtain the tetrahedron elements and the surface triangles, which is not needed in the FDM. Additionally, FDM and FEM can handle anisotropic tissues while BEM cannot. The FDM is subdivided into isotropic and anisotropic methods, or iFDM and aFDM, respectively. [27] [33]

Table 2.1: Comparison of the four methods for solving Poisson’s equation in a realistic head model is presented: boundary element method (BEM), finite element method (FEM), isotropic finite difference method (iFDM), and anisotropic finite difference method (aFDM). Reproduced from [33].

<table>
<thead>
<tr>
<th>Position of computational points</th>
<th>BEM</th>
<th>FEM</th>
<th>iFDM</th>
<th>aFDM</th>
</tr>
</thead>
<tbody>
<tr>
<td>Free choice of computational points</td>
<td>Surface</td>
<td>Volume</td>
<td>Volume</td>
<td>Volume</td>
</tr>
<tr>
<td>System matrix</td>
<td>Full</td>
<td>Sparse</td>
<td>Sparse</td>
<td>Sparse</td>
</tr>
<tr>
<td>Solvers</td>
<td>Direct/Iterative</td>
<td>Iterative</td>
<td>Iterative</td>
<td>Iterative</td>
</tr>
<tr>
<td>Number of conductivity compartments</td>
<td>Small</td>
<td>Large</td>
<td>Large</td>
<td>Large</td>
</tr>
<tr>
<td>Requires tessellation</td>
<td>Yes</td>
<td>Yes</td>
<td>No</td>
<td>No</td>
</tr>
<tr>
<td>Handles anisotropy</td>
<td>No</td>
<td>Yes</td>
<td>No</td>
<td>Yes</td>
</tr>
</tbody>
</table>

2.3.6 Solvers

Two major types of solvers, direct or iterative, may be used to solve the forward EEG problem. Sparse matrices, which may reach hundreds of thousands of equations, are quite challenging for direct methods and, as a result, iterative solvers are recommended when FDM and FEM are applied. In 2003, Mohr and Vanrumste [61] compared the most representative iterative solvers. The comparison analysed four solvers: successive over-relaxation (SOR), conjugate gradient (CG), preconditioned conjugate gradient (PCG) by symmetric SOR, and algebraic multigrid (AMG).

The SOR method represents the classical stationary methods, but it is not an optimal choice when convergence is needed. The CG method is especially suited for symmetric positive definite matrices, for which it was originally devised. It descends from the steepest descent method and it avoids repeated search by orthogonalising the research directions within the matrix. The CG method
is a special case of the PCG method obtained when the preconditioner matrix is the unity matrix. The convergence of the PCG method depends, not only on the condition of the problem matrix, but also on preconditioner method that might be the SOR. The CG method is seldom used without preconditioning. Finally, the AMG method is considered a very efficient solver for elliptic boundary problems and it can decrease the computing time of EEG forward problems.

Summing up, iterative solvers are essential in FEM and FDM, while they are optional in BEM if there is a low number of computational points. Nevertheless, the main disadvantage is the need of applying the iterative solver for each configuration. Further, the FEM and FDM would be computationally inefficient when an iterative solver needs to be used for each dipole, but the reciprocity theorem is used to overcome this inefficiency. Also, preconditioning is extremely important when a large number of compartments or anisotropic domains are included in the model decreasing the sparseness of the stiffness matrix and leading to an unstable system with slow convergence. [33] [61–64]
Design of the Head Model

Contents

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3.2 Montages and Lead Pairs .............................................................. 25
3.3 The Sensitivity Measurement ......................................................... 26
This chapter is divided in three sections that describe the methodology followed during this project. First, the construction steps of the realistic finite element method model based on MRI data are specified, the conductivity values assigned for the different compartments are listed and the model evaluation based on the reciprocity theorem is described. Secondly, the different electrode systems studied are presented, including the surface EEG and the subdermal grids. The third section explains the sensitivity measurement method used to evaluate the electrode montages.

3.1 Construction of the Five-Layered Model

Magnetic resonance images are often segmented to identify the different regions of the head to create realistic 3D models [40, 41, 43, 49, 65–73]. Due to its high resolution, MRI is preferable than other imaging modalities, such as PET and SPECT, to construct head models. This nonionizing technique also has a full three-dimensional capabilities and excellent soft-tissue contrast [42]. Besides that, MRI is the most appropriate imaging modality in the initial investigation of patients with epilepsy, even though PET and SPECT are also common. To create our model, the MRI database provided by the McConnell BIC from the Montreal Neurological Institute, which is publicly available, was used. Known as BrainWeb\(^1\) this database has been used also by Salvador et al [43] and also by Acar and Makeig [59] with the same purpose. From the two datasets available, the normal and the multiple sclerosis, the former was chosen since it corresponds to the healthy individual. The MRI data volumes were produced by an MRI simulator using three sequences (T1, T2, and proton density weighted) and a variety of slice thickness, noise levels, and levels of intensity non-uniformity. The slice thickness selected was 1 mm, the noise level 3%, and the intensity non-uniformity 20%. These combination of values was tested and shown to produce good segmentation results. The data is available for viewing in three orthogonal views: transversal, sagittal, and coronal (Fig. 3.1).

\(\text{Figure 3.1: Orthogonal slices of the MRI database from McConnell BIC.}\)

Several computational tools are involved in creating the realistic FEM model of the head from the MRI database. First of all, since the MRI data provided has a specific file format, \textit{minc} (*.mnc), not compatible with the segmentation software, a file format conversion was required. The Laboratory of

\(^1\)http://www.bic.mni.mcgill.ca
Neuro Imaging (LONI) from Los Angeles, developed Debabeler[^2], a tool that manages the conversion of medical imaging data among different file formats. We chose this tool to convert the original MRI data into nifti (*.nii) which was then readable on BrainSuite, the segmentation tool. Secondly, before the segmentation an additional step was performed to avoid the incorrect scalp segmentation. The original MRI data contains an image artefact near the scalp on the top of the head that needs to be corrected, otherwise the segmentation would have originated an extra non-realistic structure connected to the scalp (Fig. 3.2). We performed the manual correction using ImageJ[^3], an image processing program.

![Figure 3.2: Scalp surface segmented when the artefact is not removed from the MRI data.](image)

The head segmentation was executed using BrainSuite[^4]. This tool results from the collaboration between Shattuck and Leahy [74] at the UCLA Ahmanson-Lovelace Brain Mapping Center and the Biomedical Imaging Group, respectively, and the Laboratory of Neuro Imaging from the University of Southern California. BrainSuite is a collection of image analysis tools designed to process MRI data of the human head. The version BrainSuite13 provides an automatic sequence to extract cortical surface mesh models from the MRI data, tools to register these to a labelled atlas to define anatomical regions of interest, and tools for processing diffusion imaging data.

The first step of the segmentation process is the skull stripping. This stage reveals to be a crucial point of the segmentation, since the success of the segmentation depends on defining the adequate values of two parameters. The edge constant defined was 0.8, and the erosion size was 2 that is the appropriate value for high resolution MR images. The number of diffusion iterations and the value of diffusion constant used were the default values, 3 and 25, respectively. To achieve segmentation needed for this project, only the cortex surface extraction needed to be performed. After defining those parameters, the segmentation was done taking around 14 minutes on a personal computer running OS X, with 8 GB of RAM and a dual-core 2.9 GHz processor. After this step, the five *.dfs files created corresponded to the inner cortex, the brain, the inner and outer skull and the scalp surfaces (Fig. 3.3). Then, the information contained in these files was imported into MATLAB. The space

[^2]: http://www.loni.usc.edu/Software/Debabeler
[^3]: http://imagej.nih.gov/ij/
[^4]: http://brainsuite.org
between the inner skull and the brain surface was considered filled in with CSF, since there is no CSF surface that results from the BrainSuite segmentation. The importance of including one continuous CSF layer to evaluate the sensitivity of subdermal electrodes was demonstrated by Wendel et al. [9] and recommended by Subramaniyam et al. [11].

The data processing in MATLAB allowed to gather the five surfaces together in one single mesh, transform the meshed surfaces into meshed volumes, without loosing the information about the respective domain, and finally save the information in one mesh file compatible with COMSOL Multiphysics (*.mph.txt). The iso2mesh toolbox is a free matlab/octave-based mesh generation and processing toolbox that contains a long list of useful functions concerning surface and volume meshing and several operations between meshes. This open source software was developed by Qianqian Fang and David Boas [75] of the Martinos Center for Biomedical Imaging from the Massachusetts General Hospital. Using the iso2mesh toolbox, the finite element model was created, containing tetrahedral elements, the number of nodes was downsampled to 10 % of the initial value. The final FEM model (Fig. 3.4) contained 149792 nodes, 206472 faces and 883430 elements, and realistically represents the five head layers.

As discussed in the background chapter, the FEM is known for its flexibility regarding the geometry and conductivity distribution of the volume conductor model. Thus, using less computational points than FDM and allowing the analysis of, not only boundary but also domain nodes, which is not possible with BEM, the FEM appears as a suitable option for studying the EEG electrode sensitivity. Additionally, Lanfer et al. [31] mentioned that the FEM is adequate for solving the forward problem in head volume conductors incorporating thin compartments, such as subdermal electrodes. Also, to add the electrodes to the FEM model, they needed to be moved around the theoretical position. This procedure was done to avoid intersecting problems during the meshed volume generation. The electrodes were prebuilt on the FEM model using the iso2mesh toolbox, so they were already included in the final mesh file.

The last step concerning the model construction is performed in COMSOL Multiphysics, a FEM modelling platform commonly used to simulate neuroelectric problems. Salvador et al. [43] and Datta et al. [76] used a 3D finite element model to study the electric fields induced in the brain.
during transcranial current stimulation (TCS). Once the tetrahedral mesh file `.mphtxt` is imported into COMSOL, it is necessary to define the electrical conductivity and relative permittivity values required by the stationary electrical current physics within the AC/DC module.

Figure 3.4: The five-layered finite element model created using the `iso2mesh` toolbox. The realistic model contains white matter (white), gray matter (green), cerebrospinal fluid (blue), skull (red) and scalp (yellow).

3.1.1 Tissue Conductivities

As discussed in the background chapter, the conductivity of the biological tissues plays an important role in modelling the head since it highly influences the current density distributions and, consequently the lead field. Therefore, it is advisable to carefully select the most suitable conductivity values that better characterize the living tissues. Although the head tissues are anisotropic, the conductivity values of the modelled tissues were isotropically defined in this work. Despite not being the most realistic approach, the isotropic values were selected after an extensive literature review, with particular care for the skull conductivity. The chosen values for the five modelled compartments, and respective bibliographic references, are listed in Table 3.1.

![Figure 3.4](image)

Despite being required by COMSOL Multiphysics within the AC/DC module, the electrical permittivity does not influence the result of solving the quasi-static Poisson’s equation and, thus their value do not change the stationary lead field that is measured during this study. Therefore, they might be randomly chosen, as long as they are kept minimally realistic so the solvers can run.
### Table 3.1: Electrical properties of the tissues included in the head model.

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Conductivity [S/m]</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>White matter</td>
<td>0.14</td>
<td>[27]</td>
</tr>
<tr>
<td>Grey matter</td>
<td>0.33</td>
<td>[27]</td>
</tr>
<tr>
<td>CSF</td>
<td>1.82</td>
<td>[11]</td>
</tr>
<tr>
<td>Skull</td>
<td>0.058</td>
<td>[56]</td>
</tr>
<tr>
<td>Scalp</td>
<td>0.43</td>
<td>[11]</td>
</tr>
</tbody>
</table>

#### 3.1.2 Model Evaluation

The model evaluation was performed to assure the bioelectric model is properly working, which means the electric current is propagated through the model respecting the Poisson’s equation. One possible way to confirm this is by using the reciprocity theorem like Rush and Driscoll [37] did in 1969. The reciprocity theorem validation consists of solving two modelling problems, the forward and the reciprocal, and verifying if the results are consistent: the reciprocal lead voltage (Eq. 2.4) corresponds to the electric potential measured between the lead pair in the forward problem. Before the validation, it was necessary to add two surface electrodes, preferentially in opposite sides of the model and aligned along one axis. The electrodes \( C_5 \) and \( C_6 \) were added in the left and right sides of the model, respectively. These scalp positions are located slightly inferiorly to the \( C_3 \) and \( C_4 \) references, respectively, shown in Figure 2.4. Two half spherical platinum electrodes with 12 mm diameter were considered, with an electrical conductivity of \( 9.44 \times 10^6 \) S/m.

To solve the forward problem using COMSOL Multiphysics, an electric point dipole with unitary magnitude was placed in the midpoint, \((0.00 \ 0.79 \ 26.02) \) mm, between the electrode pair and it was aligned along \( xx \) axis, \( \mathbf{J}^f = [1 \ 0 \ 0] \) A m. The electric potential difference, measured between the half spherical surfaces of the electrode pair, was 148 V. Secondly, to perform the reciprocal calculation, the unitary current was injected in the outer surface of \( C_6 \) using the Terminal interface of the software. By measuring the current density \( \mathbf{J}_{LE} = [21.22 \ 0.46 \ 1.13] \) A m\(^{-2}\), in the same point where the electric dipole was previously placed, the final lead voltage reaches the value of 151.57 V, according to Eq. 2.4, taking into to account that the point lies within the isotropic white matter domain. Despite not being exactly the same, the values of the lead voltage calculated and the electric potential measured can be considered similar enough to prove the reciprocity theorem since they only differ 2.4%.

#### 3.1.3 COMSOL Multiphysics Solver

The mathematical solver predefined by COMSOL Multiphysics was the iterative algebraic multigrid solver or simply AMG. This is the default solver used by the software and its features were discussed on the background chapter. Additionally, it is important to highlight that the default solver configuration of COMSOL Multiphysics performs a fully coupled analysis which is not as accurate as a segregated analysis. However, the latter sharply increases the computing time and, to overcome this undesired fact, the non-linearity parameter can be added to the default fully coupled analysis. This feature was only applied on the forward simulation used to validate the model, because all the other simulations

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6http://www.engineeringtoolbox.com
were performed by injecting current on the electrodes. For the reciprocal problem, no differences were registered between linear and non-linear analysis.

3.2 Montages and Lead Pairs

Although the main goal of this work is to evaluate the sensitivity of subdermal EEG electrodes, the surface electrodes were also studied in order to perform comparison among the two types of electrodes. The surface montage was based on the 10-20 traditional systems, whereas the invasive electrodes were displayed in grids with 25 electrodes each placed in specific regions of the skull surface. All the electrodes were considered made of platinum with the same conductivity of the ones used to validate the model. Further, all the electrodes were insulated, using the Neumann boundary conditions equal to zero, to assure no electric current would flow through the outer half-spherical surface of the electrode.

![COMSOL Multiphysics head model with 21 surface electrodes.](image)

**Figure 3.5:** COMSOL Multiphysics head model with 21 surface electrodes.

3.2.1 Surface EEG

The surface montage contains 19 electrodes that were placed on the scalp according to the traditional 10-20 EEG system (Fig. 2.4). Two electrodes, A_1 and A_2, were not included in the model since the head volume used is inferiorly limited by a transverse plane at the same levels as the ears. In addition, two extra electrodes in the C_5 and C_6 locations were added and, in order to keep the
electrode distribution uniform, the $C_3$ and $C_4$ were slightly moved towards the center $C_Z$ electrode. In fact, $C_5$ and $C_6$ are placed above a region of particular interest for the EEG recording. Both locations are close to the squamous sutures which separate the parietal bones from the temporal bones and, as a result, the spongy middle layer of bone, the diplôme, is almost absent in these areas \cite{13}. Since EEG bipolar electrodes were considered, the reference electrode was decided to be the $C_Z$. Finally, all of the 21 platinum surface electrodes were half spherical shaped with 12 mm contact diameter (Fig. 3.5).

### 3.2.2 Subdermal Grids

Subdermal grids were geometrically created using a layout similar to the epilepsy ECoG grids commercialized by PMT Corporation\footnote{http://pmtcorp.com/index.html} (Fig. 3.6). The invasive electrodes were organized in $5 \times 5$ square grids with 10 mm spacing center-to-center of adjacent electrodes. Similarly to the surface electrodes, the invasive electrodes were also designed with a half-spherical shape, but with a smaller contact diameter of 4 mm instead. Seven subdermal grids were designed and centered on the reference locations: $C_Z$, $F_Z$, $O_Z$, $T_3$, $T_4$, $P_3$ and $P_4$ (Fig: 3.7). For each grid, the center electrode was considered the reference.

![Figure 3.6](image)

**Figure 3.6:** Subdural epilepsy electrode grid commercialized by PMT Corporation.

### 3.3 The Sensitivity Measurement

The sensitivity distributions were obtained based on the half-sensitivity volume (HSV) concept introduced by Malmivuo et al. \cite{8} to investigate the EEG and MEG detectors’ ability to concentrate their measurement sensitivity. During the last years, this concept has been also used by other researchers to study the sensitivity distributions of subdermal electrodes \cite{10, 11, 32}. The HSV is the volume of the source region of the volume conductor where the magnitude of the detector’s sensitivity is more than one half of its maximum value in the source region. If a source is homogeneously distributed, the smaller the HSV is, the smaller is the region from which the detector’s signal arises.
Figure 3.7: Subdermal electrode grids.
The gray matter was considered the source region since the neuroelectric activity is mainly generated in this domain. In fact, Subramaniyam et al. [11] have measured the HSV also considering the gray matter the domain of interest for this reason.

The sensitivity distributions of the gray matter were computed for 20 bipolar electrode pairs placed on the scalp surface and for 24 bipolar subdermal lead pairs for each subdermal grid. The sensitivity distributions were plotted in pictures with logarithmic scale in order to better evidence the current density gradient.

Note: The MMendes.MScThesis code repository contains the MATLAB scripts used to create the FEM head models and the COMSOL Multiphysics (LiveLink for MATLAB) scripts used to run the neuroelectric simulations. High computational power and memory are required to use the scripts.

https://github.com/MMendes/MMendes_MScThesis
Results

Contents

4.1 Surface 10-20 EEG System ......................................................... 30
4.2 Subdermal Electrode Grids ......................................................... 32
This chapter contains the HSV values (in $mm^3$) obtained for the electrode configurations studied. In addition, the measurement sensitivity distributions on the cortex, of the surface and subdermal leads, are illustrated in the figures found in the Appendices [A] and [B], respectively. Slices containing the gray matter cross section plane where the current density reaches the maximum values are also attached to the corresponding appendix. This chapter also contains the figures that show the most representative measurement distributions.

### 4.1 Surface 10-20 EEG System

The Table 4.1 shows the HSV values of the 20 surface bipolar leads whose reference is the $C_Z$ electrode. Apart from the values obtained for $C_5$, $C_6$ and $P_Z$ leads, the results are uniform and show that the surface electrodes detect at least half of the magnitude of an electrical source located in 1 $cm^3$ of gray matter. The mean value of the HSV registered for these 17 bipolar leads is 1014 $mm^3$. The electrodes $C_5$ and $C_6$ are sensitive to gray matter volume 45% and 62% larger than this value, respectively. The highest value obtained for the surface electrodes was registered on the $P_Z$ location where the HSV exceeds 1.3 times the mean value.

**Table 4.1:** HSV [$mm^3$] results of the gray matter of the head model bipolar lead pairs. The table is spatially organized according to the 10-20 traditional locations used on the surface EEG.

<table>
<thead>
<tr>
<th>Sinister</th>
<th>Center</th>
<th>Dexter</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anterior</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>$F_T$</td>
<td>1007</td>
</tr>
<tr>
<td></td>
<td>$F_S$</td>
<td>1011</td>
</tr>
<tr>
<td></td>
<td>$P_T$</td>
<td>1039</td>
</tr>
<tr>
<td>Center</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>998</td>
<td>1470</td>
</tr>
<tr>
<td></td>
<td>1019</td>
<td>971</td>
</tr>
<tr>
<td></td>
<td>$P_Z$</td>
<td>1039</td>
</tr>
<tr>
<td></td>
<td>$O_T$</td>
<td>992</td>
</tr>
<tr>
<td>Posterior</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>$C_T$</td>
<td>1061</td>
</tr>
<tr>
<td></td>
<td>1019</td>
<td>1019</td>
</tr>
<tr>
<td></td>
<td>$P_T$</td>
<td>1039</td>
</tr>
<tr>
<td></td>
<td>$O_T$</td>
<td>992</td>
</tr>
</tbody>
</table>

4.1 Figure 4.1 illustrates the measurement sensitivity distributions on the cerebral cortex, and the corresponding axial cross sections of gray matter, of the surface leads. The distributions focus the measurement in the brain lobes between the source and reference scalp electrodes. Regardless the volume near the reference $C_Z$, the bipolar leads measure neuroelectric activity in the lobes covered by the source electrode. Therefore, their measurement can identify the brain lobe where one electric source is active. The exceptions are the $F_Z$, $P_Z$, $C_5$ and $C_6$ leads. The measurement sensitivity of $F_Z$ and $P_Z$ electrodes (Fig. [A.2] b & c) is clearly spread on both hemispheres. The $C_5$ and $C_6$ leads (Fig. [A.1] b & h) concentrate the measurement sensitivity in the brain areas where the central sulcus finds the lateral cerebral sulci, covering part of parietal, frontal and temporal lobes.
Figure 4.1: Gray matter sensitivity distributions (columns 1 and 4) and axial cross section (columns 2 and 3) when the reference electrode is placed on C2 and the source electrode is located on Fp1, Fp2, F7, F8, T3, T4, T5, T6, O1 and O2.
4.2 Subdermal Electrode Grids

The HSVs registered for the seven subdermal grids are shown in Tables 4.2 to 4.8. In general, the results show that the grids centred on $C_Z$, $T_3$ and $T_4$ locations concentrate the sensitivity measurement in gray matter regions at least one third smaller than the surface leads. Contrastingly, the leads of the other grids do not focus the measurement in smaller volumes than the surface leads.

The HSVs of the leads in the $F_Z$ grid (Table 4.2) are much larger than the values of the surface leads, reaching the maximum volume of $3095 \text{ mm}^3$. This value was registered for the source electrode located in the center of the most inferior row. The gray matter HSV indicate an increase in the volume from 54% to 272% from which the neuroelectric activity is measured when the source electrode is laterally moved from 10 mm to 20 mm far from the reference electrode. The same variation of the electrode spacing in the inferior direction results in an increase of the HSV by 79%, but there is almost no effect when the spacing is superiorly increased. The four diagonal measurements increased in the HSV from 36% to 228% when the electrode spacing duplicated. The measurement is uniformly distributed within a large region of the frontal gray matter from both hemispheres (Fig. 4.2). When aligned with the longitudinal fissure, the leads may measure deep neuroelectric sources (Fig. B.1).

Table 4.2: HSV $[\text{mm}^3]$ results of the gray matter of the head model bipolar lead pairs of the subdermal grid centered on $F_Z$ location.

<table>
<thead>
<tr>
<th>Locations</th>
<th>Dexter $-20 \text{ mm}$</th>
<th>Dexter $-10 \text{ mm}$</th>
<th>Center $0 \text{ mm}$</th>
<th>Sinister $10 \text{ mm}$</th>
<th>Sinister $20 \text{ mm}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior 20 mm</td>
<td>2345</td>
<td>1475</td>
<td>1550</td>
<td>1643</td>
<td>2181</td>
</tr>
<tr>
<td>Superior 10 mm</td>
<td>2049</td>
<td>1071</td>
<td>1580</td>
<td>1271</td>
<td>2111</td>
</tr>
<tr>
<td>Center 0 mm</td>
<td>1642</td>
<td>1069</td>
<td>REF</td>
<td>523</td>
<td>1943</td>
</tr>
<tr>
<td>Inferior $-10 \text{ mm}$</td>
<td>1263</td>
<td>696</td>
<td>1726</td>
<td>1413</td>
<td>1078</td>
</tr>
<tr>
<td>Inferior $-20 \text{ mm}$</td>
<td>2285</td>
<td>1010</td>
<td>3095</td>
<td>561</td>
<td>1919</td>
</tr>
</tbody>
</table>

The HSV results of the subdermal grid centered on $C_Z$ (Table 4.3) show that subdermal electrodes can focus and concentrate the HSV within gray matter regions as small as $44 \text{ mm}^3$. When the electrode spacing doubles laterally, the HSV increases from 4% to 16%. Increasing the spacing along the anterior direction also results in an increase of the HSV (51%), but along the posterior direction the value decreases 7%. Contrarily, the diagonal measurements increase posteriorly and decrease anteriorly when the electrode spacing doubles. The measurement sensitivity of the leads of the $C_Z$ grid (Figs. 4.3 and B.2) concentrates in small areas of the superficial gray matter. The extension of sensitivity distributions increases when the comparing the 20 mm to the 10 mm subfigures.

Table 4.3: HSV $[\text{mm}^3]$ results of the gray matter of the head model bipolar lead pairs of the subdermal grid centered on $C_Z$ location.

<table>
<thead>
<tr>
<th>Locations</th>
<th>Sinister $-20 \text{ mm}$</th>
<th>Sinister $-10 \text{ mm}$</th>
<th>Center $0 \text{ mm}$</th>
<th>Dexter $10 \text{ mm}$</th>
<th>Dexter $20 \text{ mm}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anterior 20 mm</td>
<td>61</td>
<td>66</td>
<td>139</td>
<td>126</td>
<td>72</td>
</tr>
<tr>
<td>Anterior 10 mm</td>
<td>55</td>
<td>66</td>
<td>92</td>
<td>78</td>
<td>70</td>
</tr>
<tr>
<td>Center 0 mm</td>
<td>50</td>
<td>48</td>
<td>REF</td>
<td>55</td>
<td>64</td>
</tr>
<tr>
<td>Posterior $-10 \text{ mm}$</td>
<td>49</td>
<td>44</td>
<td>55</td>
<td>52</td>
<td>58</td>
</tr>
<tr>
<td>Posterior $-20 \text{ mm}$</td>
<td>53</td>
<td>50</td>
<td>51</td>
<td>54</td>
<td>57</td>
</tr>
</tbody>
</table>
Figure 4.2: Gray matter sensitivity distributions (a - d) and transverse cross section (e - h) when the reference electrode is placed on $F_{Z}$ and the source electrode is moved laterally 10 mm and 20 mm.

Figure 4.3: Gray matter sensitivity distributions (a - d) and coronal cross section (e - h) when the reference electrode is placed on $C_{Z}$ and the source electrode is moved laterally 10 mm and 20 mm.
In the $O_Z$ grid (Table 4.4), the HSV values obtained increase when the electrode spacing is either laterally increased or increased along the vertical axis. Also, the diagonal measurements also show positive variations with the HSV increasing from 21% to 393% when the electrode spacing duplicates. However, comparing electrodes with non parallel lead fields, it is possible to verify that the HSV not always increase when the source electrode is further from the reference. This fact is clear on the superior and sinister part of the grid. There, the HSVs first increase when the lateral spacing is increased from 0 mm to 10 mm, but then they decrease with the 20 mm spacing. Figures 4.3 and 3.2 show that the $O_Z$ grid is slightly displaced towards the dexter side of the skull because the measurement sensitivity is predominant on the right side of the cortex. The leads of the $O_Z$ grid detect the neuroelectric activity originated in a wide area of the occipital lobe.

Table 4.4: HSV [mm$^3$] results of the gray matter of the head model bipolar lead pairs of the subdermal grid centered on $O_Z$ location.

<table>
<thead>
<tr>
<th>Locations</th>
<th>Sinister</th>
<th>Sinister</th>
<th>Center</th>
<th>Dexter</th>
<th>Dexter</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior 20 mm</td>
<td>755</td>
<td>1042</td>
<td>931</td>
<td>1188</td>
<td>1587</td>
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<tr>
<td>Superior 10 mm</td>
<td>111</td>
<td>512</td>
<td>230</td>
<td>322</td>
<td>913</td>
</tr>
<tr>
<td>Center 0 mm</td>
<td>616</td>
<td>342</td>
<td>REF</td>
<td>271</td>
<td>752</td>
</tr>
<tr>
<td>Inferior -10 mm</td>
<td>529</td>
<td>421</td>
<td>268</td>
<td>282</td>
<td>594</td>
</tr>
<tr>
<td>Inferior -20 mm</td>
<td>513</td>
<td>480</td>
<td>429</td>
<td>439</td>
<td>476</td>
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</tbody>
</table>

Figure 4.4: Gray matter sensitivity distributions (a - d) and transverse cross section (e - h) when the reference electrode is placed on $O_Z$ and the source electrode is moved laterally 10 mm and 20 mm.
Similarly to $C_Z$ grid, both $T_3$ and $T_4$ grids show that subdermal electrodes can concentrate the lead field in small regions of the gray matter. The maximum HSV value registered for the $T_3$ grid was $159 \text{ mm}^3$ (Table 4.5) while the maximum value of the $T_4$ grid was $304 \text{ mm}^3$ (Table 4.6). Tables 4.5 and 4.6 indicate a decrease in the volume of $13\%$ and $33\%$, respectively, when the electrode spacing is posteriorly increased from $10 \text{ mm}$ to $20 \text{ mm}$. The inferior and posterior leads of both grids also show a decrease of the volume when the electrode spacing is posteriorly increased. The diagonal measurement in this part of the grids decreases $34\%$ in the $T_3$ grid, but increases $5\%$ in the $T_3$ grid. The distributions of the leads in $T_3$ grid (Figs. 4.5 and B.4) and $T_4$ grid (Figs. B.5 and 4.6) concentrate the measurement sensitivity in a restricted volume of the temporal gray matter between the measurement pair. The orientation and distance between the leads highly influence the portion of the temporal lobe being measured.

Table 4.5: HSV [$\text{mm}^3$] results of the gray matter of the head model bipolar lead pairs of the subdermal grid centered on $T_3$ location.

<table>
<thead>
<tr>
<th>Locations</th>
<th>Anterior 20 mm</th>
<th>Anterior 10 mm</th>
<th>Center 0 mm</th>
<th>Posterior 10 mm</th>
<th>Posterior 20 mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior 20 mm</td>
<td>86</td>
<td>89</td>
<td>133</td>
<td>89</td>
<td>147</td>
</tr>
<tr>
<td>Superior 10 mm</td>
<td>88</td>
<td>83</td>
<td>66</td>
<td>84</td>
<td>159</td>
</tr>
<tr>
<td>Center 0 mm</td>
<td>93</td>
<td>103</td>
<td>REF</td>
<td>117</td>
<td>102</td>
</tr>
<tr>
<td>Inferior −10 mm</td>
<td>120</td>
<td>138</td>
<td>130</td>
<td>81</td>
<td>78</td>
</tr>
<tr>
<td>Inferior −20 mm</td>
<td>126</td>
<td>143</td>
<td>102</td>
<td>88</td>
<td>85</td>
</tr>
</tbody>
</table>

Figure 4.5: Gray matter sensitivity distributions (a - d) and coronal cross section (e - h) when the reference electrode is placed on $T_3$ and the source electrode is moved inferior and superiorly 10 mm and 20 mm.
Table 4.6: HSV $[mm^3]$ results of the gray matter of the head model bipolar lead pairs of the subdermal grid centered on $T_4$ location.

<table>
<thead>
<tr>
<th>Locations</th>
<th>Posterior $-20$ mm</th>
<th>Posterior $-10$ mm</th>
<th>Center $0$ mm</th>
<th>Anterior $10$ mm</th>
<th>Anterior $20$ mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior 20 mm</td>
<td>217</td>
<td>304</td>
<td>198</td>
<td>175</td>
<td>163</td>
</tr>
<tr>
<td>Superior 10 mm</td>
<td>172</td>
<td>199</td>
<td>203</td>
<td>272</td>
<td>175</td>
</tr>
<tr>
<td>Center $0$ mm</td>
<td>163</td>
<td>244</td>
<td>REF</td>
<td>121</td>
<td>158</td>
</tr>
<tr>
<td>Inferior $-10$ mm</td>
<td>179</td>
<td>213</td>
<td>180</td>
<td>192</td>
<td>225</td>
</tr>
<tr>
<td>Inferior $-20$ mm</td>
<td>141</td>
<td>142</td>
<td>205</td>
<td>199</td>
<td>204</td>
</tr>
</tbody>
</table>

Figure 4.6: Gray matter sensitivity distributions (a - d) and transverse cross section (e - h) when the reference electrode is placed on $T_4$ and the source electrode is moved anterior and posteriorly 10 mm and 20 mm.

The results obtained for $P_3$ and $P_4$ grids show that the HSV measurement is spread along a large volume of gray matter that varies between $451$ $mm^3$ to $2553$ $mm^3$ (Table 4.7). The values of the $P_3$ grid (Table 4.7) indicate an increase of the HSV from 156% to 229% when the electrodes increase their spacing from 10 $mm$ to 20 $mm$ from the reference electrode. The same variation of the electrode spacing on the $P_4$ grid (Table 4.8) results in an increase of the HSV from 88% to 235%.

Table 4.7: HSV $[mm^3]$ results of the gray matter of the head model bipolar lead pairs of the subdermal grid centered on $P_3$ location.

<table>
<thead>
<tr>
<th>Locations</th>
<th>Lateral $-20$ mm</th>
<th>Lateral $-10$ mm</th>
<th>Center $0$ mm</th>
<th>Medial $10$ mm</th>
<th>Medial $20$ mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior 20 mm</td>
<td>2427</td>
<td>1814</td>
<td>1746</td>
<td>1526</td>
<td>1590</td>
</tr>
<tr>
<td>Superior 10 mm</td>
<td>2056</td>
<td>922</td>
<td>598</td>
<td>1001</td>
<td>857</td>
</tr>
<tr>
<td>Center $0$ mm</td>
<td>1518</td>
<td>593</td>
<td>REF</td>
<td>451</td>
<td>1315</td>
</tr>
<tr>
<td>Inferior $-10$ mm</td>
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<td>874</td>
<td>531</td>
<td>926</td>
<td>1718</td>
</tr>
<tr>
<td>Inferior $-20$ mm</td>
<td>2553</td>
<td>1820</td>
<td>1747</td>
<td>1499</td>
<td>808</td>
</tr>
</tbody>
</table>
Table 4.8: HSV \( [mm^3] \) results of the gray matter of the head model bipolar lead pairs of the subdermal grid centered on \( P_4 \) location.

<table>
<thead>
<tr>
<th>Locations</th>
<th>Medial −20 mm</th>
<th>Medial −10 mm</th>
<th>Medial 0 mm</th>
<th>Lateral 10 mm</th>
<th>Lateral 20 mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior 20 mm</td>
<td>1736</td>
<td>1870</td>
<td>1540</td>
<td>1880</td>
<td>2520</td>
</tr>
<tr>
<td>Superior 10 mm</td>
<td>1045</td>
<td>887</td>
<td>819</td>
<td>1038</td>
<td>2045</td>
</tr>
<tr>
<td>Center 0 mm</td>
<td>1767</td>
<td>528</td>
<td>REF</td>
<td>557</td>
<td>1299</td>
</tr>
<tr>
<td>Inferior −10 mm</td>
<td>2372</td>
<td>925</td>
<td>565</td>
<td>831</td>
<td>1499</td>
</tr>
<tr>
<td>Inferior −20 mm</td>
<td>1154</td>
<td>1576</td>
<td>1073</td>
<td>1578</td>
<td>1966</td>
</tr>
</tbody>
</table>

Figure 4.7: Gray matter sensitivity distributions when the reference electrode is placed on \( P_4 \) and the source electrode is moved, either laterally or inferior and superiorly, 10 mm and 20 mm.

The measurement sensitivity distributions of the \( P_3 \) grid (Fig. 4.6) and \( P_4 \) grid (Fig. 4.7) illustrate the wide regions of the parietal lobes where the measurement sensitivity lies.

Figure 4.8 shows the differences in the measurement sensitivity distributions that result from the varying thicknesses of the underlying tissues. The \( F_Z \) grid covers an area where the bone and CSF layers are very thick, while the \( C_Z \) covers a skull region with thin bone and CSF layers underneath. In the \( F_Z \) grid, the electric current is first shunted by the thick skull and then concentrates within the high conductive CSF between the lead pair (Fig. 4.8a). In the \( C_Z \) grid, the shunting effect of the skull is reduced and the thin CSF layer channels a small part of the electric current (Fig. 4.8b). The results show that the current density in the gray matter under the \( F_Z \) grid spreads uniformly along a wide area (Fig. 4.8a), while in the \( C_Z \) grid the current is concentrated in the gray matter between the lead sites (Fig. 4.8b).
Figure 4.8: Sagittal cross section (with skull, CSF and gray matter layers) of the sensitivity distribution for $F_Z$ and $C_Z$ subdermal leads.
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- **5.2 Subdermal Electrode Grids** ........................................... 41
- **5.3 Comparison Surface vs. Subdermal Leads** .......................... 42
- **5.4 Influence of the mesh on the measurements** .......................... 43
The HSV measurement reflects the lead capability of focusing the lead field within the neuroelectric source volume. Therefore, low HSV values are registered when the lead field is concentrated in a small region of the gray matter, whereas high HSV identify the lead fields that are spread along a large volume of gray matter. Our results show that the surface bipolar leads are sensitive to gray matter volumes about 1 cm³ when the Cz is the reference electrode. In addition, this study also shows that the electrode measurement sensitivity benefits from subdermal implantation, but the localization of the electrode grid is a crucial point.

5.1 Surface 10-20 EEG System

The mean HSV of the surface leads, 1014 mm³, excludes three electrodes (C5, C6 and Pz), and it is almost one fourth of the result reported by Vaisanen et al. [10], 4002 mm³. This difference is clearly due to the source region considered. While we have considered exclusively the gray matter as the neuroelectric source, Vaisanen et al. measured the sensitivity of the electrodes to electric sources within the whole brain (gray and white matter included). Therefore, our study measures smaller volumes that do not include the white matter as neuroelectric source.

The large values obtained for both C5 and C6, 1470 mm³ and 1641 mm³, respectively, are explained by the volume of gray matter that lies bellow these locations. These electrodes measure the gray matter volume from gyri that belong to three distinct brain lobes: parietal, frontal and temporal (Fig. A.1 b & h). These areas contain more gray matter than white matter. Consequently, C5 and C6 leads are sensitive to larger gray matter volumes than the rest of the electrodes. Nevertheless, their measurement can hardly distinguish one active source located in the parietal lobe from another located in the temporal or frontal lobes. The large volumes detected are also due to the segmentation performed that did not create a brain surface where the sulci between the brain lobes are evidenced. Therefore, the lateral cerebral sulcus and the central sulcus are overfilled with gray matter that belongs to the HSV detected by C5 and C6 electrodes (Fig. A.1 e & k).

The highest HSV registered for the surface montage, 2374 mm³, corresponds to the Pz location, which is aligned along the longitudinal fissure. However, this physical division between the cerebral hemispheres is not evidenced in the posterior part of the brain surface segmented (Fig. D.1 a). Therefore, this electrode detects sources from the non realistic gray matter portion located in the posterior brain. Contrarily, the brain surface shows a slight division between hemispheres in the frontal area (Fig. D.1 b), which is detected by the Fz electrode and explains its smaller HSV.

The measurements sensitivity distributions of Fz and Pz leads spread in the two cerebral hemispheres. The Fz distribution (Fig. A.2 c) focuses the lead field within the parietal lobes, while the Pz distribution (Fig. A.2 b) concentrates the lead field within the frontal lobes. This implies that their measurements can hardly distinguish between neuroelectric sources located near the longitudinal fissure but in different hemispheres.

Apart from C5, C6, Fz and Pz, the distributions of the other sixteen surface leads show to focus the measurement sensitivity in specific brain lobes. Concluding, scalp electrodes are capable to detect
and distinguish between neuroelectric sources originated from different brain lobes, as long as they are not placed in the scalp surface that lies above the brain sulci. Further, according to the cross sections shown on the figures of the Appendix A, scalp leads measure neuroelectric activity from superficial and deep sources.

5.2 Subdermal Electrode Grids

In general, the leads of C2, T3 and T4 grids concentrate the HSV in a gray matter region at least one third smaller than the surface leads. The maximum HSV registered for these grids, 304 mm$^3$ (Table 4.6), differs 14% from the maximum gray matter HSV registered by Subramaniyam et al. [11], 266 mm$^3$, and it is almost half of the subdermal HSV obtained by Vaisanen et al. [10], 516 mm$^3$.

The slight variation from the value reported by Subramaniyam et al. [11] might be due to the different geometry of the brain surface, while the difference from Vaisanen's results is clearly due to our neuroelectric source volume that only contains gray matter.

The HSVs of the leads of the Fz grid (Table 4.2) are much larger than the values of the surface leads, which is not expected. The subdermal HSVs should be inferior that the surface values since the smearing effect of the scalp should be reduced [10]. These large values are explained by the thick CSF layer that lies under the frontal bone where the grid was placed. Since the CSF is the biological tissue modelled with the highest electrical conductivity, the electric current that does pass through the skull preferentially spreads along the CSF layer and concentrates between electrodes before entering the less conductive brain tissues (Fig. 4.8a) [9, 45]. Consequently, the detectors sensitivity to sources located in the gray matter is very poor and the HSVs become larger [8, 45]. This means that the electrodes detect a source located in a volume of 1 to 3 cm$^3$ of the gray matter near the lead (Figs. 4.2 & 5.1).

Secondly, the results of the subdermal grid centered on Cz (Table 4.3) show that the sensitivity of subdermal electrodes might be improved when the grid is placed is specific regions of the skull. Around the Cz location, the CSF that lies between the skull and the cortex is very thin. The small HSVs result from the fact that the measurement sensitivity concentrates within the gray matter between the measurement leads (Fig. 4.8b). However, the center column of the grid shows high values which is not accordingly to the results obtained by Subramaniyam et al. [11]. These values are high due to extra gray matter inaccurately placed in the location of the longitudinal fissure, which is a consequence of the wrong gray matter segmentation. Nevertheless, the data show that subdermal electrodes eliminate much of the scalp and skull smearing, which affect the surface measurements, since the HSV are clearly much smaller than the results of the scalp measurements.

The HSV fluctuations on the superior and sinister part of the Oz grid are the consequence of the grid misplacement. The grid is not aligned with the longitudinal fissure; it slightly covers more the dexter brain than the sinister. Thus, the longitudinal fissure lies between the 0 mm center column and the 10 mm sinister column, instead of being exactly under the center column. The leads on the superior 10 mm sinister column are then sensitive to the gray matter located in the longitudinal fissure.
reaching high HSV values; the leads on the superior 20 mm sinister column are distant enough from the fissure which results in smaller HSVs. In the inferior part of the grid, this phenomenon is silenced due to the proximity to the thicker bone and CSF layers. The lead field of the inferior electrodes is firstly shunted by the thick skull and then spread along the brain due to the high conductive CSF that becomes thicker in this zone [8, 9, 45].

The HSVs of the leads of both $T_3$ and $T_4$ grids (Tables 4.5 and 4.6) show that the EEG detectors sensitivity may benefit from subdermal implantation. The temporal squama (Fig. C.1) is a thin and flat part of the temporal bone [13], and it is the exact region where the temporal grids were placed. Consequently, the smearing effect caused by the low conductive skull is reduced and the lead field is focused on a small volume of gray matter which results in small HSVs. Furthermore, the skull smearing effect is minimized on both grids when the source electrode gets closer to the squamous suture (Fig. C.1) where the bone layer is thinner [13]. This explains why the HSVs decrease when the source electrode is posteriorly moved on the inferior part of the grids.

The $P_3$ and $P_4$ grids studied (Figs. 3.7f & g) cover the thicker portion of the parietal bones (Fig. C.2). Therefore, in these locations, the skull shunting effect is amplified as well as the smearing effect cause by its low conductivity [32, 54]. It results in small current density values uniformly distributed in the gray matter which is reflected by the large HSV values.

### 5.3 Comparison Surface vs. Subdermal Leads

The subdermal electrodes are capable to measure more accurate and localized source regions within the gray matter than the traditional surface EEG, because they can reach HSVs smaller than the volumes measured by surface electrodes. The subdermal implantation benefits from the reduced smearing effect. The scalp smearing effect is totally bypassed while the smearing effect caused by the skull is partially eliminated. Additional, the subdermal recordings take place closer to the source region which also improves the sensitivity of the electrodes.

Nevertheless, this improved accuracy is only reachable when the subdermal grids are implanted in specific regions of the skull surface. From the seven subdermal locations analysed, the temporal bones and the skull near $C_2$ reference have shown to be good locations to implant subdermal electrodes. These regions benefit from the thin underlying bone and CSF layers. On the other side, the frontal and parietal bones have shown to be bad implantation places due to the thick bone or CSF layer.

Concluding, the surface electrodes can give information regarding the brain lobe where the neuroelectric source lies, but subdermal implantation is needed to get more spatially precise data. The surface detectors can measure brain activity from regions of 1 cm³, while subdermal electrodes can focus the measurement on volumes smaller than 0.1 cm³. This means that sensitivity of electroencephalography electrodes can be improved by a factor of 10 when the electrodes are subdermally implanted instead of placed on the scalp.
5.4 Influence of the mesh on the measurements

Lastly, the refinement of the mesh included in the head model might be also an important factor that influence the results obtained, more specifically for the models containing the subdermal grids. When creating the FEM model with the 25 electrode grids, the mesh generation step often took longer periods of time. This factor is related to the intersection of the head and electrode meshes. With the decrease of the electrode diameter, from surface to subdermal electrodes, the elements of the electrode mesh become smaller and, as a result, the number of nodes in the interface between the electrode and skull is increased. Consequently, this portion of the final model contains very tiny elements with small volume. During the simulations, these elements were sometimes so small that the solvers could not run due to mathematical indeterminations. During this project the models were many times redone because of this fact, but the mesh was not controlled. Therefore, the element size differences in the interface between the electrode and the skull might be the reason why the current density is not so realistically distributed through the tissues and the measurements may not be so accurate.
Conclusions and Future Work
This thesis focused on the sensitivity measurement of the surface and subdermal electroencephalography electrodes. The results obtained provide both modelling and clinical specialists useful information regarding the EEG. The geometry of the layers of the model was shown to influence the volume of gray matter detected by the electrodes because it affects the lead field. Thus, the implantation of subdermal grids should take into account the thickness variations of the underlying tissues in order to optimize the electrodes performance. According to their location, the subdermal electrodes may provide more focused measurements than the scalp electrodes, which is desirable for the source localization EEG problem.

Concluding, a realistic head model, containing at least the five layers modelled, is an essential tool to study the sensitivity distributions of electroencephalography electrodes. Such a model may be used to demonstrate the improved accuracy obtained when subdermal implanted electrodes are used instead of the traditional surface EEG.

Nevertheless, further investigation on the influence of the thickness of head tissues on the HSVs is required. One approach should consist of placing electrode grids on the scalp and confirm the results obtained for the subdermal grids. Then, ECoG grids may be added to the head model to compare subdermal and subdural leads and further investigate the influence of the low conductive skull layer.
Bibliography


Sensitivity Distributions of Surface Electrodes
Figure A.1: Gray matter sensitivity distributions (a - c ; g - i) and coronal cross section (d - f ; j - l) when the reference electrode is placed on $C_2$ and the source electrode is located on $T_3, C_5, C_3, C_4, C_6$ and $T_4$. 

Minimum to maximum sensitivity
Figure A.2: Gray matter sensitivity distributions (a - d) and sagittal cross section (e - h) when the reference electrode is placed on CZ and the source electrode is located on O1, Pz, Fz and FP1.
Sensitivity Distributions of Subdermal Electrodes
Figure B.1: Gray matter sensitivity distributions (a - d) and midsagittal cross section (e - h) when the reference electrode is placed on $F_Z$ and the source electrode is moved along the central sulcus 10 mm and 20 mm inferior and superiorly.

Figure B.2: Gray matter sensitivity distributions (a - d) and midsagittal cross section (e - h) when the reference electrode is placed on $C_Z$ and the source electrode is moved along the central sulcus 10 mm and 20 mm anterior and posteriorly.
Figure B.3: Gray matter sensitivity distributions (a - d) and sagittal cross section (e - h) when the reference electrode is placed on $O_Z$ and the source electrode is moved along the central sulcus 10 mm and 20 mm inferior and superiorly.

Figure B.4: Gray matter sensitivity distributions (a - d) and transverse cross section (e - h) when the reference electrode is placed on $T_3$ and the source electrode is moved anterior and posteriorly 10 mm and 20 mm.
Figure B.5: Gray matter sensitivity distributions (a - d) and coronal cross section (e - h) when the reference electrode is placed on $T_4$ and the source electrode is moved inferior and superiorly 10 mm and 20 mm.

Figure B.6: Gray matter sensitivity distributions when the reference electrode is placed on $P_3$ and the source electrode is moved, either laterally or inferior and superiorly, 10 mm and 20 mm.
Skull Anatomy
Figure C.1: Right lateral view of the skull. Reproduced from [13].

Figure C.2: Medial view of sagittal section of the skull. Reproduced from [13].
Segmented Brain Surface
Figure D.1: The longitudinal fissure is slightly evidenced on the frontal brain but not on the posterior brain.