ABSTRACT

SINGH, VINAY. Implantable Devices for a Retinal Prosthesis: Design and Electromagnetic and Thermal Effects. (Under the direction of Dr. Gianluca Lazzi.)

A retinal prosthesis, wherein electrical stimulation is provided to the retina of a person inflicted with outer retinal degenerative diseases such as Retinitis Pigmentosa and Age-related Macular degeneration, has been clinically tested and has succeeded in providing limited vision; such as shape recognition. It is hoped that an increased electrode count will improve visual acuity.

The retinal prosthesis considered in this work is a dual-unit system with an external (outside the human body) unit and an internal (inside the body) unit with a wireless link for power and data transfer between them. Such a system poses possible health risks due to the incident electromagnetic energy of the wireless link and the power dissipated by the internal components, particularly the processing chip which drives the electrodes responsible for eliciting a neural response from the retina. Tissue damage via heating is one the primary concerns for such a system making it necessary to obtain via simulation and in-vivo and in-vitro experiments, accurate estimates of thermal elevation due to the operation of the such devices.

In this work, numerical methods have been developed to compute temperature increases and electromagnetic effects due to the prosthesis components in anatomically correct human head models. The explicit and the Alternating-Direction Implicit (ADI) Finite-Difference Time-Domain (FDTD) have been used. Further, a hybrid explicit-ADI method was developed for the heat equation which provided simulation speedup of more than 10x over the conventional ADI method for the models considered.

FDTD methods were employed to compute the induced current densities and Specific Absorption Rate (SAR) in the human head due the inductive link comprising the primary coil (external) and a secondary coil (internal). Different orientations of the primary coil were considered in a frequency range of 1 MHz-20 MHz to provide guidelines for choosing eventual frequency and power parameters to conform to international safety standards. A novel displacement field excitation
method was used for the spiral primary coil and verified with analytical results.

In an effort to reduce the size of the internal unit and to allow integration of a patch antenna (for a separate data link), and the active devices on a single substrate, a 3-D trench inductor geometry was investigated. To enable patterning of structured surface, a custom experimental setup was designed and maintained to process a positive tone PEPR2400 electro-depositable photoresist.
Implantable Devices for a Retinal Prosthesis: Design and Electromagnetic and Thermal Effects

by
Vinit Singh

A dissertation submitted to the Graduate Faculty of North Carolina State University in partial fulfillment of the requirements for the Degree of Doctor of Philosophy

Electrical Engineering

Raleigh, NC
2008

APPROVED BY:

Dr. Griff Bilbro
Dr. Zhilin Li

Dr. Gianluca Lazzi
Dr. Mehmet Ozturk
Chair of Advisory Committee
DEDICATION

To my brother, Amit; and my parents.
BIOGRAPHY

Vinit Singh was born in Jamshedpur, India. In 1997, he joined St. Stephen’s College, Delhi for a bachelors in mathematics. After a year, he moved to the Indian Institute of Technology at Roorkee (formerly University of Roorkee), where he completed his bachelors in Electronics & Communication engineering in 2002. Subsequently, he joined University of Cincinnati where he obtained a masters in electrical engineering. Since 2004, he has been a graduate student in the department of Electrical & Computer engineering at North Carolina State University.

His research interests are in applied electromagnetics: including computational electromagnetics, human interaction with electromagnetic energy, bio-electricity, development of wireless telemetry systems, and wireless power transfer magnetic resonance coupling.
ACKNOWLEDGMENTS

I would like to thank Dr. Gianluca Lazzi for giving me the opportunity to work in his research group. His guidance and help were always unqualified, in issues related to academics, research or otherwise. My sincere gratitude to Dr. Mehmet Ozturk and Dr. Zhilin Li, who provided numerous insights and advice in the areas of fabrication and numerical mathematics, respectively. I thank Dr. Griff Bilbro for being a part of my research committee and giving critical comments and suggestions.

It was a privilege to be part of an eclectic research group of extremely talented and interesting individuals. I thank Dr. Keyoor Gosalia and Dr. Stefan Schmidt for laying the foundation on which I built this work. I would like to thank all my colleagues over the last four years, for collaboration on research projects, and providing an escape from the rigors of academics and research through stimulating discussions and other activities: Dr. Gaurav Gupta, Amit Qusba, Srinivas Jasti, Carlos Cela, Ajeet, Nitin Kwatra, Randall Barlow, Tong Zhengxin, Patrick Brown, Dr. Anand Konanur and Shruti Soora. I thank Sundar Srinivas for being a great fellow guitarist, and sports and music enthusiast; and Dr. Ajit Rajagopalan for his suggestions on everything, his 'taxi' services and for being a competitive tennis partner. I learnt a lot from them.

I would also like to acknowledge some great friends outside of my lab who were there for everything non-academic: I thank Dr. Shalini Gupta, and soon to be Drs: Radhika Shah, Vinayak Rastogi and Rakesh Ranjan.

I thank my parents for their support for all my endeavors. I thank my brother, Amit for the initial impetus for graduate studies and the resources and subsequent encouragement.

I was very lucky to have the constant unreserved support of Itisha. Her contribution was vital.
TABLE OF CONTENTS

LIST OF FIGURES ......................................................... viii

LIST OF TABLES .......................................................... xiii

1 Introduction ............................................................. 1
  1.1 Motivation .......................................................... 1
  1.2 The Eye Physiology and the Epi-Retinal Prosthesis .......... 2
    1.2.1 RP and AMD ................................................. 3
    1.2.2 Epi-Retinal Prosthesis .................................. 4
  1.3 Other Visual Prostheses .......................................... 6
    1.3.1 Sub-Retinal Prosthesis .................................. 6
    1.3.2 Optic Nerve Prosthesis .................................. 7
    1.3.3 Cortical Prosthesis ...................................... 8
  1.4 Electromagnetic and Thermal Effects ......................... 8
    1.4.1 Electromagnetic Effects ................................ 8
    1.4.2 Thermal Effects .......................................... 11
    1.4.3 Models for Compliance Testing ......................... 14
    1.4.4 Computational Methods .................................. 16
  1.5 Wireless Telemetry ............................................... 16
    1.5.1 Inductor Design Issues ................................ 18
      1.5.1.1 Dimensions and Geometry .......................... 18
      1.5.1.2 Power Transfer Efficiency ......................... 18
      1.5.1.3 Frequency of Operation ............................ 19
      1.5.1.4 Electromagnetic Effects on tissues ................ 19
    1.5.2 Integrated Inductors .................................... 20
  1.6 Overview of this Dissertation ................................. 21

2 Finite-Difference Time-Domain Methods for Maxwell’s Equations
  and the Bio-Heat Equation ........................................... 22
  2.1 Introduction ..................................................... 22
  2.2 The FDTD Method For Electromagnetics ....................... 23
    2.2.1 Yee’s Algorithm and FDTD ............................. 23
    2.2.2 The D-H Formulation .................................... 24
    2.2.3 Absorbing Boundary Conditions ......................... 27
    2.2.4 Errors and Stability .................................... 28
  2.3 Finite-Difference Methods for the Bio-Heat Equation ........ 29
6 Conclusion and Future Work .................................................. 97
  6.1 Summary of Contribution ............................................. 97
    6.1.1 Electromagnetic Effects ....................................... 97
    6.1.2 Thermal Effects .................................................. 98
    6.1.3 Fabrication of a 3-D inductor .................................. 99
  6.2 Future Work ............................................................. 99
    6.2.1 ADI-FDTD method for problems in bioelectromagnetics .. 99
    6.2.2 Hybrid explicit-ADI method for Thermal computations .. 100
    6.2.3 Electric Field Measurement using E-field Probe .......... 101

Bibliography ........................................................................ 104
LIST OF FIGURES

Figure 1.1  A Flowchart indicating the electrical/neural signal flow with a retinal prosthesis as compared to the natural signal flow ................... 3

Figure 1.2  A schematic of the epi-retinal prosthesis approach( [39]. .......... 4

Figure 1.3  The full body man model from NLM. The size of this full model at 1 mm uniform resolution is 586 × 340 × 1878 .......................... 15

Figure 2.1  The Yee Grid as used in this work. The locations of the D-field components are identical to that of the corresponding E-fields [45]. .... 26

Figure 2.2  (a) The voxel-centered computational grid. (b) The thermal resistance between adjacent voxels at (i,j,k) and (i+1,j,k). (c) The thermal RC network at a voxel node (i,j,k)................................. 31

Figure 2.3  (a) Figure of just the eye with the coil. Note that only position b (square coil in the anterior position of the eye) was considered for the comparison. The simulation model was uniform resolution at 0.5 mm.(b) Comparative time-domain plots of the temperature increase for the two methods at a position on the coil. They show very good agreement. ... 35

Figure 3.1  (a) A 3D model of the head and the approximate location of the primary coil (b)Slice of the human head indicating the extracted region for the FDTD simulations.(b) The extracted model with the implanted secondary coil. The size of the extracted region at 0.3 mm resolution was 180×240×200............................. 41

Figure 3.2  Figure (a) shows the intended position of the implant and (b) shows the discretized 3-D model of the package around the eyeball ...... 42

Figure 3.3  The different orientations of the eye and primary coil considered. 42

Figure 3.4  (a) and (b) indicate the magnitude and direction (indicated by sign on the colorbar) of y- and z- components of the current .......... 44
Figure 3.5 The Electric field magnitudes in V/m at 2 MHz along the cross-section containing the center of the spiral coil (a) Analytical, using equation 3.4.1 (b) with FDTD. Both results are without the head model .... 45

Figure 3.6 (a) The H-field distribution at a cross-section of the entire model (including head) for case A’ (shown in Figure 3.3a) (b) The variation of the H-field along the axis of the concentric loops (using formula) and the axis of the spiral (FDTD). For 5 and 10 MHz, the results are with the head model without the implant. In both the figures, the dip in the H-field magnitude can be seen at the location of the secondary coil .... 46

Figure 3.7 The E-fields, current density and the SAR along a transverse cross-section of the head model (all values are single-voxel) ................. 47

Figure 3.8 The plot gives the variation of the E-field along a line parallel to the primary coil axis and through the head. Figure (a) is for different frequencies for a primary coil current of 0.62 A. In figure (b), the field values at all frequencies are scaled to 20 MHz. It indicates that a simple frequency scaling can be performed to obtain E-field values till at least 20 MHz. 49

Figure 3.9 Variation of the peak 1-g SAR with excitation frequency (current, I=0.62 A) for case A’. ‘Er’ indicates the error due to scaling from 10 MHz ................................................................. 49

Figure 3.10 Variation of the $J_{\text{max,rms}}$ with frequency for case A’ ........... 51

Figure 4.1 Figure 1 (a) shows the intended position of the implant and 1 (b) shows the discretized 3-D model of the package around the eyeball .... 55

Figure 4.2 Discretized model of coil placed in truncated eye model .......... 55

Figure 4.3 The grid-scheme of the explicit-ADI method (shown in 2-D). The ADI method was used in the higher resolution region at 0.25 mm (model size $64 \times 82 \times 118$), and the explicit method was used for the lower resolution region at 0.5 mm. 59

Figure 4.4 Applied power versus time. ................................................. 61

Figure 4.5 Drawing of the chip package indicating the two locations of thermistors that were accurately monitored in simulations. The tip is placed in contact with the package. 61
Figure 4.6  The dimensions and shape of the thermistor. .......................... 62

Figure 4.7  Comparison of temperature increase along a cross-section through the eye and implant after 12 minutes of physical time for a package dissipating 97 mW. Top: Using ADI method, Bottom: Explicit-ADI ........ 63

Figure 4.8  Temperature reading for DOG2060L over time: experimental versus simulated. Since the power applied for the first 22 minutes is negligible as illustrated in figure 4.4 the results here are for a period of 76 minutes after the first 22 minutes. 'd' is the approximate distance between the package and the thermistor tip. The simulated values are for Case-1.... 66

Figure 4.9  Temperature gradient in tissue just outside package over the length of the thermistor (Case-1). ......................................................... 68

Figure 4.10  (a) shows the cross-sectional plane of the eye and implant over which the thermal elevation is illustrated in (b) for case-1(top) and case-2(bottom). The colorbar shows the temperature in degrees C. .......... 69

Figure 4.11  Temperature history at some locations for case-1 (Top), and case-2 (Bottom).................................................................. 70

Figure 4.12  Temperature Profile along axis (shown in figure 4.10) of chip package. ................................................................. 71

Figure 4.13  Temperature Difference for sample points in Muscle (above package) and Retina (below package).................................. 72

Figure 4.14  a) shows the cross-sectional plane of the eye and coil over which the thermal contour is illustrated in (b). The temperature increment in the sclera has been suppressed. ....................... 74

Figure 4.15  Power dissipated per electrode volume (a) for a pulse width of 1 ms (b) for a pulse width of 3 ms. Note that the biphasic pulse repetition rate is 50 Hz. ................................................................. 75

Figure 4.16  The temperature increases in a cross-section of the back of the eye, consisting of the electrodes. ................................. 76

Figure 5.1  The proposed trench design for a 3-D inductor. The planar trench region in the center could be used for a patch antenna and the back side for other electronics [?]. ................................. 79
Figure 5.2  The masks were for 6 inch wafers. (a) Locations of the windows for bulk silicon etching. The shaded region along two perpendicular diameters indicates the location of the test and alignment structures. (b) Mask for patterning of metal lines ........................................ 80

Figure 5.3  Maximum possible resolution with increasing etch depth........ 82

Figure 5.4  Schematic of the experimental setup used for electro-deposition of PEPR2400. ................................................................. 85

Figure 5.5  This figure shows the a cross-section of the Si wafer during the processes before the formation of the window for the wet Si etch....... 88

Figure 5.6  This figure illustrates the patterning of the wafers and the Si etch process. Note that the isotropic etch using KOH also etches the oxide and nitride layer, although much slower than silicon. However, etch rates should be factored in during the following nitride etch. ............ 89

Figure 5.7  (a) The figure indicates the problem with standard exposure techniques on deeply etched surfaces. (b) The result of underexposure...... 90

Figure 5.8  The result of underexposure and subsequent de-scumming. Note the thinner PEPR in the trench due to cross-exposure. ............... 91

Figure 5.9  An SEM image of isotropically etched silicon. While patterning lines near the base is relatively easy, it is virtually impossible to pattern near the top. ...................................................... 92

Figure 5.10  An illustration of the surface roughness experienced after deep isotropic etches. ................................................................. 92

Figure 5.11  An anisotropically etched wafer with a thick coating of PEPR2400 photoresist ................................................................. 93

Figure 5.12  A typical etch profile after anisotropic and isotropic etching..... 93

Figure 5.13  Illustration of poor adhesion of the photoresist. The unexposed photoresist tended to get washed away by the developer solution. .... 94

Figure 5.14  Titanium metal lines for a planar inductor on unetched silicon. The line-width here is 10 µm. ..................................................... 95

Figure 5.15  SEM images of titanium lines (25 µm) on anisotropically etched silicon. The lines are significantly thinned after the metal etch step... 96
Figure 6.1 The electric fields due to a single-turn square coil. (a) Coil model with source location. The white line indicates the cross-section along which the electric fields are shown in figures (b), (c) and (d). The ADI-FDTD simulation was performed with the $\Delta t = 8\Delta t_{\text{explicit}}$. 

Figure 6.2 An illustration of the errors in the induced electric field magnitudes of the human body at 1 mm resolution for the ADI-FDTD method ($\Delta t_{\text{ADI}} = 16\Delta t_{\text{explicit}}$) relative to the explicit FDTD method. 

Figure 6.3 (a) Equipment and setup for electric field measurements. (b) Preliminary measurements for a spiral coil along its diameter between 2 mm and 7 mm from the plane of coil.
LIST OF TABLES

Table 1.1 IEEE reference levels for an uncontrolled environment [19] ........... 11

Table 1.2 ICNIRP reference levels for General Public Exposure [20] ........... 12

Table 1.3 SAR and Current Density restrictions for general public exposure (uncontrolled environment) ................................................. 13

Table 3.1 Conductivity of key tissues at different frequencies ................. 40

Table 3.2 Peak 1-g SAR and Averaged Current Densities at 10 MHz for peak primary coil current of 0.62 A ................................................. 48

Table 3.3 Peak 1-g SAR and $J_{\text{max,rms}}$ with frequency for the worst case (A') for peak primary coil current of 0.62 A ................................................. 50

Table 4.1 Thermal properties of implanted materials ............................... 58

Table 4.2 Thermistor Locations: Experimental ................................. 60

Table 4.3 Temperature increase in canines: Experimental ..................... 60

Table 4.4 Temperature reading with open and closed eye: Experimental .... 62

Table 4.5 Maximum and Average single-voxel temperature increment due to operation of chip. (in degrees Centigrade) ......................... 64
Table 4.6 Thermal elevation after 4 minutes: Comparison between simulation and experimental (DOG2060) ................................................................. 65

Table 4.7 Thermal elevation after 50 minutes for simulation and about 12 minutes from experimental measurements. I: When the thermistor tip is approximately 0.35 mm from package. II: When the thermistor tip is approximately 0.71 mm from the package. .............................. 67

Table 4.8 Maximum single-voxel and Average temperature increment in implant materials and key tissues due to operation of chip: Simulation (all temperatures are in degrees Centigrade) ........................................... 71

Table 4.9 Thermal effects due to operation of the secondary coil ............... 73
Chapter 1

Introduction

1.1 Motivation

Any device which can assist in triggering neural activity in the human body can be referred to as a neural prosthesis (NP). Extensive research in the last few decades has led to a strong understanding of the human neural system allowing researchers to explore direct stimulation of the target neurons. The concurrent advances in material science, fabrication technology, data processing algorithms and surgical techniques have spurred collaborative biomedical research in areas once considered prohibitively complicated. It is estimated that since 1963, over 40000 NPs have been implanted to alleviate respiratory problems, restore hearing and provide bladder control [1].

The first observation of a visual neural response to electrical stimulation was made by a German neurosurgeon, Foerster, in 1929, when his patient ’saw’ a spot of light when his visual cortex was electrically stimulated. A collection of these spots of light, called phospenes, forms the image seen by a blind patient. This led to the first effort for a visual prosthesis in 1968, where the cortical region of the brain was targeted [2].

Currently, various approaches are being pursued by many research teams all over the world, with the fundamental difference being in the location of the electrical stimulus  [3, 4, 5, 6]. Specifically, the epi-retinal prosthesis targets the inner
retina and is aimed for people suffering from outer retinal degeneration. In the design considered for this work, the prosthesis is a dual-unit system with external (outside the body), and internal (inside the body) components, with an inductive wireless link for power and data transfer from the former to the latter [7, 8].

Before implanting in the body, it is imperative to verify (experimentally and via simulation of human models) that the operation of the prosthesis components does not harm the tissues in any way. In this dissertation, to facilitate a safety evaluation of the electromagnetic and thermal effects, numerical methods based on the explicit Finite-Difference Time-Domain (FDTD) and Alternating-Direction Implicit (ADI) FDTD methods have been developed to obtain time-marching solutions of the Maxwell’s equations and Pennes’ Bioheat equation [9], which can be used to compute the induced electromagnetic fields due to the wireless link, and heat generated by the prosthesis electronics, respectively. Modeling techniques have been devised to enable accurate placement of the various system components. Additionally, an investigation via photolithographic fabrication has been carried out for the development of a novel 3D inductor geometry which could offer higher inductance and compactness by enabling integration of the active devices on the same substrate. The following sections describe the epi-retinal prosthesis, and then review the issues and the advantages and disadvantages of other visual prostheses being developed.

1.2 The Eye Physiology and the Epi-Retinal Prosthesis

Light enters the eye through the cornea, passes through the lens and vitreous humor which forms the bulk volume of the eye. Once it reaches the retina, it travels through the retina thickness and reaches the photoreceptor cells located at the outer retinal region just under the pigment epithelium. The photoreceptor cells comprise of rods and cones which contain a photopigment which enables the conversion of the light signal to electrical and chemical signals (neural signals).
Figure 1.1: A Flowchart indicating the electrical/neural signal flow with a retinal prosthesis as compared to the natural signal flow

These signals travels via the retinal layers to reach the ganglion cells which links to the optic nerve, as shown in the flowchart in figure 1.1.

1.2.1 RP and AMD

Retinitis Pigmentosa (RP) and Age-Related Macular Degeneration (AMD) are two of the most common retinal diseases. Both these diseases stem from a progressive loss of the photoreceptor cells, eventually leading to complete blindness. It is estimated that at least 7 million people worldwide are affected, and the count increases by about 700,000 per year in the United States alone [8, 10].

A strong motivation towards the development of a prosthesis for people inflicted by RP and AMD is the fact that most of their visual pathway leading up to the brain is intact. Post-mortem morphometric analysis has shown that, in these diseases, while less than 5 % of the photoreceptors survive, 30-80 % of the ganglions and bipolars are functional [2]. It has been found that by injecting an appropriate pulse, the action potential of these surviving cells can be reached [4]. In fact, early clinical experiments on blind patients have elicited strong visual
Figure 1.2: A schematic of the epi-retinal prosthesis approach [39].

percepts and enabled identification of shapes of large objects [7].

1.2.2 Epi-Retinal Prosthesis

The Epi-Retinal Prosthesis (ERP) considered in this work is a dual unit device, with components internal and external to the human body [11]. The external components consist of a video camera to acquire images from the external world. These images are then processed by an chip which encodes the data in a specific format. A primary coil, powered by a battery, is used for wireless power and data transfer to the internal components. A secondary coil inside the body provides power and modulated data to the stimulator chip, which decodes the incoming data and drives the electrodes which will be located in the foveal region at the inner retinal region. Figure 1.2 illustrates the basic concept of the epi-retinal prosthesis. While the location of the electrode array is fixed, the location of the stimulator chip will be decided by factors such as thermal safety, length of the interconnect between the electrodes and the chip, and its positional stability.
Each electrode contact generally corresponds to a stimulating channel with a dedicated driver circuit in the stimulator chip. In principle, higher the number of stimulation points, better the resolution. A well established electrical-stimulus based neural prosthesis is the cochlear implant, which can provide near normal listening by using just 6 stimulating channels targeting about 50000 fibers in the auditory nerve [8]. Analogous to the auditory nerves are the axons of the ganglion cells which form the optic nerve, which number about 1.2 million. Therefore, by simple extrapolation, a significantly, if not proportionally, larger number of stimulating channels are needed to achieve a similar degree of sight. According to [8], about 600-1000 pixels/channels will be required to provide visual functions such as face recognition and reading. An immediate consequence of the increase in channels is the increased complexity of the system and more power dissipation.

Since the retina is very sensitive to heat [12], and its functioning is critical for this system, any strong heat source, like the chip, should be kept as far away as possible. However, large distances between the electrode array and the stimulator chip will necessitate long interconnects running through the tissues in and around the eye. For the secondary coil, in addition to the considerations mentioned above, it will need to be sufficiently close to the primary for maximum and stable coupling. Further, the packaging of these components should prevent any interaction of the electronics with tissues and should be bio-compatible.

Using a dual-unit system minimizes the number of internal components and keeps open the possibility of future upgrades to the system without re-entrant surgery. Another advantage of all retinal prostheses is that by being located early in the visual pathway, it uses most of the natural spatio-temporal encoding of the nervous system. However, a significant disadvantage of the ERP is that the curvature of the retina makes chronic lodging of the electrode array at the correct position difficult.
1.3 Other Visual Prostheses

The visual pathway extends from the photoreceptors to the visual cortex region in the brain, and in principle, visual percepts can be excited by providing the correct stimulus at any point along this path. The closer to the visual cortex the artificial stimulus is provided, the more visual ailments it can address. For example, the epi-retinal device can be used only for people suffering from outer retinal degeneration, while an implant at the optic nerve can serve its purpose even if the retina is completely atrophied. The positive aspect of the any retinal implant is that an implant in the retinal region leaves most of the signal processing to the functional natural cells, and therefore requires lesser knowledge of the mapping scheme and signal processing done by the neurons, leading to a relatively simpler stimulation system. In this section, a concise review of the approaches being followed by various research groups is presented. It should be mentioned that all visual prosthesis approaches face many common challenges, like biocompatibility of implanted materials, efficient stimulus parameters, and long term safety.

1.3.1 Sub-Retinal Prosthesis

Similar to the ERP, the Sub-Retinal Prosthesis (SRP) is targeted towards people suffering from AMD or RP. In one of the SRP configurations, the implantable system is a single component comprising of photodiodes to convert the incident light to electrical current, and the contact electrodes to inject charge into the retina[4, 13]. The implant is placed at the outer retina, just under the pigment epithelium, and mimics the operation of the photoreceptors of transducing incident light into electrical signals (using the photodiodes), and relaying them to the subsequent cells (using electrodes). This characteristic of the SRP, wherein it minimizes the amount of processing required by the prosthesis, best preserves the natural signal processing done by the retinal cells, and greatly simplifies the system as compared to the ERP. Other advantages of this approach include easy placement, lesser components, and, since no video camera is used, the possibility
of using the movement of the eye to locate objects [14]. In its original form, the principal drawback of this system is the inability of the photodiodes to generate strong enough currents to stimulate the adjacent neurons with just the incident light. Therefore, efforts are currently being made to use an active subretinal implant powered by a wireless inductive link [14], which will likely pose additional design and positioning issues. It should be mentioned that there is another research group which has a design very similar to the ERP discussed in section 1.2.2, but positions the electrode array in the subretinal space [15]. A major disadvantage of either SRP configuration is that the placement of the photodiode/electrode array component could damage the retina through pigment epithelium separation or obstruction of blood flow from the retina.

1.3.2 Optic Nerve Prosthesis

In the optic nerve prosthesis (ONP), a cylindrical cuff electrode array with electrodes on the inside surface is used to wrap around the optic nerve just outside the eye [3]. Similar to the ERP, this approach will require real-time image capture, subsequent processing and a telemetry link for power and data to the electrodes. By completely bypassing the retina, not only can it be used for patients with large-scale retinal degeneration, it also provides for a relatively safer and easier implantation procedure compared to the retinal approaches. According to [16], the axonal fibers in the optic nerve are bundled in a specific order in that there is a topographic correspondence between the fibers and their origin at the ganglion cell layer in the retina: the fibers close to the surface of the nerve connect to the ganglions at outer region of the retina, while the internal fibers are connected to the ganglions of the foveal/macular region (which forms the central region of the vision space in a healthy eye). Thus, percepts due to the ONP will tend to have peripheral vision. Also, this approach is relatively immature, and according to [5], the possibility of generating pattern vision by eliciting spatially adjacent phosphenes with optic nerve stimulation is still unclear.
1.3.3 Cortical Prosthesis

The Cortical region has the distinction of being the first location of interest for a visual prosthesis. The location of the Cortical Prosthesis (CP) is the visual cortex in the brain, relatively far from the eye region. After Foerster’s landmark experiment, mentioned in section 1.1, it was Krieg who first suggested a cortical visual prosthesis with multiple excitation points to form a single coherent image[17]. The first elaborate experiments were carried out in the seventies using surface electrodes in the occipital cortex of blind patients. Specifically, Dobelle used a 64-channel platinum based electrode array which allowed patients to see 6-inch characters at a distance of 5 feet [18]. The surface electrodes were soon replaced by intracortical electrodes where the electrodes were inserted into the occipital cortex, which required stimulus currents of magnitude 10-100 times smaller than that by the surface electrodes to elicit a neural response [2]. Other significant advantages of this were reduced power requirements per stimulation site, lesser phosphene interaction, absence of flicker, more predictable phosphene generation, and possibility of packing more stimulating channels thereby increasing resolution.

The most significant benefit of the intra-cortical approach is that it can be used for almost any kind of blindness, owing to its location in the final regions of the visual pathway. As mentioned in the introduction of this section, its distance from the front end of the visual pathway makes spatial mapping very complicated. Further, the convoluted surface of the brain is a problem for chronic placement of a device, and any surgical complications could lead to severe complications and even death.

1.4 Electromagnetic and Thermal Effects

1.4.1 Electromagnetic Effects

Over the last few decades, there has been a tremendous growth in wireless applications. Power transmission, numerous household applications, broadcast radio and television, automotive electronics, telecom, cellular networks, Local
Area Networks (LANs), etc; all form a complex mix of electromagnetic (EM) energy permeating regular human life covering a very wide range of frequencies from 60 Hz to a few GHz. This leads to the question whether the ubiquitous EM energy can cause harm to humans. And if it does, what kind of harm? What is the threshold?

Since the 1960s, extensive experimental, analytical, and simulation-based research have been performed on animals such as rodents, canines and monkeys and even human volunteers, to evaluate responses to exposure to EM fields. This has lead to the formation of working committees which survey recent scientific literature on biological effects of EM fields and periodically create or update standards or safety guidelines for frequencies from 0 Hz up to 300 GHz. Commonly referred and quoted standards are from the Institute of Electrical and Electronics Engineers (IEEE) [19], International Council for Non-Ionizing Radiation Protection (ICNIRP) [20] and National Radiological Protection Board (NRPB) [21]. Obviously, the stipulations from all the agencies show a strong overlap. So far, while much research has been dedicated to the investigation of indirect EM effects such as those by power lines, the only indisputably accepted effects are: neural stimulation at low frequencies (below 1 MHz) and tissue heating at higher frequencies.

The energy absorption leading to tissue heating is characterized by the Specific Absorption Rate (SAR), which is defined as the time rate of incremental energy, \( \text{SAR} = \frac{d}{dt} \frac{dW}{dm} \), dissipated in material incremental mass \( dm \), contained in a volume element \( dV \) of mass density \( \rho \). The SAR is related to the electric field by \( \text{SAR} = \frac{\sigma}{2 \rho} |E|^2 \), where \( \sigma \) is the electrical conductivity of the tissue. In essence, the SAR is the \( I^2R \) power dissipation in tissue.

The limits on induced current densities are imposed to prevent any unwanted neuronal stimulation. A human body cell membrane (which conveys the neural signals) is like a capacitor which becomes progressively less responsive with increasing frequency. Studies have shown that strong, inductive ELF fields have the capability to excite nerves directly causing undesirable physiological reaction. Further, even a below-threshold but substantial chronic stimulation values can adversely affect the activity of the central nervous system [20]. However, above
10 MHz, the exposure restrictions are dictated by tissue heating via energy deposition, rather than stimulation. As the frequency increases, there is lesser penetration of the EM energy and more absorption near the surface of the body; therefore at frequencies above 10 GHz or so, restrictions are imposed on the incident power density to ascertain surface heating.

In general, the exposure guidelines maintain a large margin of safety, for example, from experimental data on animals and humans it has been found that incident fields leading to a whole body SAR greater than 4 Watts/Kg would lead to thermal effects beyond the thermoregulatory capacity of the body [ICNIRP]. Keeping this in mind, ICNIRP imposes a further safety factor of 50 in its guidelines leading to maximum allowable whole body SAR of 0.08 Watts/Kg.

The most commonly followed standards are those stipulated by IEEE and ICNIRP. Table 1.1 and table 1.2 give the reference levels for the induced fields spatially averaged over the entire body for an uncontrolled environment for different frequencies as mandated by these agencies and table 1.3 gives the basic restrictions on averaged induced current densities and SARs. Note that the reference levels (in table 1.1 and 1.2) may be exceeded as long as the basic restrictions (in table 1.3) are complied with. A controlled environment is one when the people are aware of the kind of EM exposure they are subjected to (also called Occupational Exposure). Restrictions for an uncontrolled environment (also called General Public Exposure) are generally about 5 times stricter than that for a controlled environment.

The retinal prosthesis prototype described in this work comprises of an EM source in the wireless and telemetry link and a number of metallic implanted parts in the eye region. Before implanting in the human body, it is imperative that the implantable components of the retinal prosthesis be tested for safety of the patient in the short and long term. For the epi-retinal prosthesis, the likely frequency range in which the wireless link will be operated is between 1 to 20 MHz. According to the IEEE and ICNIRP standards (tables 1.1, 1.2, and 1.3), in this frequency range, both; the SAR and the induced current densities need to be computed for compliance testing.
Table 1.1: IEEE reference levels for an uncontrolled environment [19]

<table>
<thead>
<tr>
<th>Part A: Induced Electromagnetic Fields</th>
</tr>
</thead>
<tbody>
<tr>
<td>The exposure values in terms of the electric and magnetic field strengths are the mean values obtained by spatially averaging the squares of the fields over area equivalent to the the vertical cross section of the human body.</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Frequency range (MHz)</th>
<th>Electric field (V/m)</th>
<th>Magnetic field (A/m)</th>
<th>Power density (E-,H-)fields (mW/cm²)</th>
<th>Averaging Time</th>
<th>Averaging Time [E]² or [H]² (min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.003-0.1</td>
<td>614</td>
<td>163</td>
<td>(100,10⁶)</td>
<td>6</td>
<td>6</td>
</tr>
<tr>
<td>0.1-1.34</td>
<td>614</td>
<td>16.3/f</td>
<td>(100,10⁴/f²)</td>
<td>6</td>
<td>6</td>
</tr>
<tr>
<td>1.34-3.0</td>
<td>823.8/f</td>
<td>16.3/f</td>
<td>(180/f²,10⁴/f²)</td>
<td>f²/0.30</td>
<td>6</td>
</tr>
<tr>
<td>3.0-30</td>
<td>823.8/f</td>
<td>16.3/f</td>
<td>(180/f²,10⁴/f²)</td>
<td>30</td>
<td>6</td>
</tr>
<tr>
<td>30-100</td>
<td>27.5</td>
<td>158.3/f²</td>
<td>(.294×10⁴/f²,336)</td>
<td>30</td>
<td>.0636f²</td>
</tr>
<tr>
<td>100-300</td>
<td>27.5</td>
<td>0.0729</td>
<td>0.2</td>
<td>30</td>
<td>30</td>
</tr>
<tr>
<td>300-3000</td>
<td>–</td>
<td>–</td>
<td>f/1500</td>
<td>30</td>
<td></td>
</tr>
<tr>
<td>3000-15000</td>
<td>–</td>
<td>–</td>
<td>f/1500</td>
<td>90000/f</td>
<td></td>
</tr>
<tr>
<td>15-300 GHz</td>
<td>–</td>
<td>10</td>
<td>616000/f²</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Part B: Induced and contact RF currents</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum Current (mA)</td>
</tr>
<tr>
<td>Through both feet</td>
</tr>
<tr>
<td>Through each foot</td>
</tr>
<tr>
<td>Contact</td>
</tr>
<tr>
<td>----------------------------------------</td>
</tr>
<tr>
<td>Frequency range (MHz)</td>
</tr>
<tr>
<td>0.003-0.1</td>
</tr>
<tr>
<td>0.1-100</td>
</tr>
</tbody>
</table>

### 1.4.2 Thermal Effects

With regard to temperature increases, the above mentioned standards impose limitations on indirect thermal effects, that is, temperature increase due to EM energy absorption. Direct effects, due to the heat dissipated by passive devices, and more importantly by implanted active devices is not covered. In fact, guidelines or references specific to implant induced temperature rises is scant, probably because of the relative immaturity of the field of active bio-implantable devices.
Table 1.2: ICNIRP reference levels for General Public Exposure [20]

<table>
<thead>
<tr>
<th>Frequency range</th>
<th>E-field (V/m)</th>
<th>H-field (A/m)</th>
<th>Equivalent plane-wave power density (W/m²)</th>
<th>Averaging time (minutes)</th>
</tr>
</thead>
<tbody>
<tr>
<td>upto 1 Hz</td>
<td>–</td>
<td>3.2×10^4</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>1-8 Hz</td>
<td>1×10^4</td>
<td>3.2×10^4/f^2</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>8-25 Hz</td>
<td>1×10^4</td>
<td>4×10^4/f</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>0.025-0.8 kHz</td>
<td>250/f</td>
<td>4/f</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>0.8-3 kHz</td>
<td>250/f</td>
<td>5</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>3-150 kHz</td>
<td>87</td>
<td>5</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>0.15-1 MHz</td>
<td>87</td>
<td>0.73/f</td>
<td>–</td>
<td>6</td>
</tr>
<tr>
<td>1-10 MHz</td>
<td>87/f^{0.5}</td>
<td>0.73/f</td>
<td>–</td>
<td>6</td>
</tr>
<tr>
<td>10-400 MHz</td>
<td>28</td>
<td>0.073</td>
<td>2</td>
<td>6</td>
</tr>
<tr>
<td>400-2000 MHz</td>
<td>1.375f^{0.5}</td>
<td>0.0037f^{0.5}</td>
<td>f/200</td>
<td>6</td>
</tr>
<tr>
<td>2-300 GHz</td>
<td>61</td>
<td>0.16</td>
<td>10</td>
<td>–</td>
</tr>
</tbody>
</table>

In general, the human-core body temperature is assumed to be homeostatically regulated around 37 degrees C. According to [22], the interior of the thorax and abdomen, the brain and a part of the skeletal muscles comprise the core. The threshold for inducing physiological effects has been reported to be 1-2 degrees C for the body core [20]. In the brain, a tolerance of 4.5 degrees C (for 30 minutes) has been suggested by [23] for neuronal injury, and the threshold for retinal damage is suggested to be around 2 degrees C [24] above the normal retinal temperature [25]. According to [26, 27] thermally induced damage to the skin can occur for an increase of about 10 degrees C. For active implantable devices, the only available standard suggests that the maximum temperature of the outer surface of the implant should not exceed the normal “surrounding body temperature of 37 degrees C” by 2 degrees C [28].

The sources of heat due to the epi-retinal prosthesis is the power dissipation by the stimulator chip, the power dissipation by the current-carrying electrode array and the secondary coil, and the absorption of electromagnetic energy due to the wireless telemetry link. Amongst these, the heat generated by the chip is by the far the largest [29]. However, the heating due to the electrode array, because of
Table 1.3: SAR and Current Density restrictions for general public exposure (uncontrolled environment)

<table>
<thead>
<tr>
<th>Frequency Range</th>
<th>Current Density (Head and Trunk) mA m$^{-2}$(rms)</th>
<th>Whole Body average SAR W Kg$^{-1}$</th>
<th>Localized SAR (head and trunk)</th>
<th>Localized SAR (limbs)</th>
</tr>
</thead>
<tbody>
<tr>
<td>up to 1 Hz</td>
<td>8</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>1-4 Hz</td>
<td>8/f</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>4 Hz-1 kHz</td>
<td>2</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>1-100 kHz</td>
<td>f/500</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>100 kHz-10 MHz</td>
<td>f/500</td>
<td>0.08</td>
<td>2</td>
<td>4</td>
</tr>
<tr>
<td>10 MHz-10 GHz</td>
<td>-</td>
<td>0.08</td>
<td>2</td>
<td>4</td>
</tr>
</tbody>
</table>

(1) f is the frequency in Hz. (2) Current densities should be averaged over a cross-section of 1 cm$^{-2}$ perpendicular to the current direction. (3) Localized SAR averaging mass is any 10 g of contiguous tissue.

II. IEEE

<table>
<thead>
<tr>
<th>Frequency Range</th>
<th>Current Density SAR (in MHz)</th>
<th>Whole Body average SAR W Kg$^{-1}$</th>
<th>Localized SAR (in MHz)</th>
</tr>
</thead>
<tbody>
<tr>
<td>3 kHz-100 kHz</td>
<td>15.7 f$^*$</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>100 kHz-6 GHz</td>
<td>-</td>
<td>0.08</td>
<td>1.6$^\dagger$</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>4$^\ddagger$</td>
</tr>
</tbody>
</table>

*: f is the frequency in MHz
$^\dagger$: SAR averaged over any 1 g of tissue (over tissue volume in shape of a cube)
$^\ddagger$: SAR averaged over any 10 g of tissue (over tissue volume in shape of a cube)

its proximity to the sensitive retina, and the electromagnetic energy absorption, due to the relatively high conductivity of the retina and surrounding tissues, will also need to be considered.
1.4.3 Models for Compliance Testing

Since it is not possible to use live human heads for RF dosimetry, besides animal testing, computational methods tend to provide a very good estimate of dose quantities on actual human heads. The usage of numerical tools requires:

1. Accurate human body models which can easily be imported by the computational software.

2. Knowledge of the dielectric properties of all the tissue at the frequency of operation.

Currently, there are a few voxelized (VOlume piXEL= VOXEL) human body models that are used for computations, available from different agencies. The Utah Man from the University of Utah [30], the Visible Man from the National Library of Medicine (NLM) [31], NORMAN, NORmalized MAN [32] from the NRPB, and the Japanese male model [33] are the most well known and commonly used for computing RF doses. These anatomically correct models are generally obtained from MRI scans of human volunteers and even cadavers. Individual tissues are then identified distinguished from an analysis of the MRI 2-D scans. Finally, all the 2-D scans are merged to form a smooth 3-D model maintaining smooth boundaries between various tissue types. The Visible Man from the NLM, used in this work, is shown in figure 1.3. This model was obtained as 1 mm thick slice with a resolution of 1 mm. In this work, since only the head region was required, to save on computation resources, the model was truncated neck below, and linear interpolators have been used to further increase resolution up to 0.0625 mm.

The most comprehensive repository of frequency dependent dielectric properties of human tissues has been compiled by Gabriel et al [34] in collaboration with Brooks Air Force Base in Texas, USA. The data has been obtained via extensive experimental measurements, and consequent curve fitting using 4 Cole-Cole analysis given by:

\[ \varepsilon(\omega) = \varepsilon_\infty + \sum_{m=1}^{4} \frac{\Delta \varepsilon_m}{1 + (j\omega \tau_m)^{1-\alpha_m}} + \frac{\sigma}{j\omega \varepsilon_0} \]  \hspace{1cm} (1.4.1)
where $\omega = 2\pi f$, $j = \sqrt{-1}$, and $\epsilon_0 = 8.85 \times 10^{-12} \text{F/m}$ is the permittivity of free space. Using the parametric data, equation 1.4.1 can be written as

$$\epsilon(\omega) = \epsilon_\infty + \frac{\Delta_1}{1 + (j\omega\tau_1)^{\alpha_1}} + \frac{\Delta_2}{1 + (j\omega\tau_2)^{\alpha_2}} + \frac{\Delta_3}{1 + (j\omega\tau_3)^{\alpha_3}} + \frac{\Delta_4}{1 + (j\omega\tau_4)^{\alpha_4}} + \frac{\sigma}{j\omega\epsilon_0}$$

(1.4.2)

In the online resource compiled by Gabriel et al [34], the fitted parameters mentioned in equation 1.4.2 have been given for over 40 tissues.

A physical phantom model, which is now a worldwide standard is the Specific Anthropomorphic Mannequin (SAM), is also commonly used for compliance testing with the actual radiating device. SAM is a lossless plastic shell with dimensions taken from data corresponding to an adult male as tabulated by the US Army. For experiments, the shell is generally filled with a homogeneous fluid with the average properties of the tissue at the test frequency. Probes can be then inserted to measure the required quantities (electric field, magnetic field, temperature).
1.4.4 Computational Methods

The problems of electromagnetic interactions with biological tissue is generally too complicated to be solved analytically. Consequently, numerical methods are used which also facilitate the use of the anatomically accurate models (section 1.4.3).

The most common Computational ElectroMagnetics (CEM) software are largely based on the Finite-Difference Time Domain (FDTD) method, the Finite-Element (FE) method, and the Method of Moments (MoM). Among the three, the most accurate is the MoM, but it is also the most inefficient and toughest to implement and requires the derivation of a geometry-specific Green’s function and the solution of a dense system of linear equations. The FE method leads to sparse matrices which are relatively simpler to solve. However, the FDTD method is the simplest to implement and is a time domain solution, thus allowing the possibility of a broadband analysis of structures. Further, the FDTD method also allows the easy import of the computational voxelized biological model making it ideally suited for BEM in general. It should be noted that current state of the art in CEM includes many hybrid combinations of the above mentioned three methods for improved overall performance, time-domain implementations of the FE method, and even semi-implicit implementations of the FDTD method such as the Alternating-Direction Implicit (ADI) FDTD method or the Crank-Nicholson (CN) FDTD method.

1.5 Wireless Telemetry

Any biomedical implant consisting of an active device chronically implanted inside the human body needs constant power supply and, in some cases, real-time data from the external world for operation. In some cases, a reverse transfer of data from the implanted device may also be required to enable diagnostic checks on the operation of the implanted device. Such an arrangement requires continuous transfer of power and data to the internal components. While a wired connection
to the internal components will constitute a stable and efficient link, it has some inherent safety issues:

1. The presence of a wired connection leads to a continuous breach of skin raising the possibility of infections.

2. Tethering the implant with a wire will likely lead to a restriction in movement of the eye. Further, any strong impulsive movement could lead to device dislodging, or a breakage in the wire.

Implanting a battery is also not an option due to the requirement of regular surgeries for replenishment or re-charging.

In recent years, a commonly used method for power and data transfer for neuro-prosthetic devices has been inductive wireless telemetry using a primary coil positioned outside the body and driven by a power source (for example, a battery pack) and a secondary coil, placed inside the body, in which the energy is induced/coupled [35]. Energy transfer using inductive coupling is widely used these days in diverse applications such as transformers and RFID (Radio Frequency IDentification) tags, amongst others.

The induction mechanism is simple and is quantitatively described by Faraday’s law, wherein a current is induced in a conductor placed in a time-varying magnetic field. For strong coupling to take place, the secondary should be in the reactive near field of the source of the magnetic field. In spherical coordinates, the fields produced at any point \((r,\theta,\phi)\) by a simple current loop (magnetic dipole) in the x-y plane are given by the following equations [36]

\[
\begin{align*}
E_\phi &= \frac{60\pi I A \sin(\theta)}{\lambda} \left( \frac{\omega}{cr} - \frac{j}{r^2} \right) \quad (1.5.1) \\
H_r &= \frac{120\pi I A \cos(\theta)}{\lambda \mu_0} \left( -\frac{j}{cr^2} - \frac{1}{\omega r^3} \right) \quad (1.5.2) \\
H_\theta &= \frac{60\pi I A \sin(\theta)}{\lambda \mu_0} \left( \frac{\omega}{c^2 r} - \frac{j}{cr^2} + \frac{1}{\omega r^3} \right) \quad (1.5.3)
\end{align*}
\]

where \(E_\phi\) is an electric field component, \(H_r\) and \(H_\theta\) are magnetic field components, \(\mu_0\) is the free space permeability, \(\lambda\) is the wavelength, \(c\) is the speed of light, \(I\) is
the amplitude of the current in the loop and \( A \) is the cross-section of the loop. Clearly, for distances close to the source, the terms with \( \frac{1}{r} \) (the \( H_r \) and \( H_\theta \) terms) dominate. This region is called the near field, where maximum magnetic coupling can take place. Further, the power carrier can be modulated by the data stream.

1.5.1 Inductor Design Issues

1.5.1.1 Dimensions and Geometry

The dimensions of the coils generally depend on the environment in which they are placed. Coils implanted inside the body will likely need to be small to displace minimum tissue volume as well as adhere to strict limits on shape and size, for example, a coil placed inside the eye.

Generally, the design limitations will be far less severe for a primary coil placed outside the human body. However, certain specific characteristics of the primary can significantly affect system performance. For example, a method to improve insensitivity of relative lateral movements of the coils is to make the primary coil very large compared to the secondary. The negative aspect of this is the degradation of efficiency of power transfer. Further, the coupling can be enhanced by distributing the metal wires along the diameter of the secondary coil at the expense of the unloaded quality factor, owing to increase in wire length for the same inductance value. Coil geometries can also be made to reduce coupling variations with coil rotations. Further, 3-D stacked/multilayer coils will provide better coupling, but add to complexity of fabrication and placement and adjustment with the body tissues.

1.5.1.2 Power Transfer Efficiency

It is not preferred to have ferrite cores in inductive links for biomedical implants since it would lead to unwanted coupling with other sources such as MRI systems, power lines, etc. This could lead to large spurious currents even in the presence of low ambient H-fields. The air-cored transformers hence formed have
a very low coupling factor with typical values being between 0.1 to 0.01. Other factors that affect the coupling are coil load, operating frequency, number of turns, alignment of the primary and secondary, coil geometry and the materials used as the conductors and insulation.

1.5.1.3 Frequency of Operation

A critical figure of merit for an inductor is the quality factor, \( Q = \frac{\omega L}{R} \), where \( \omega \) is the operating frequency, \( L \) is the inductance and \( R \) is the resistive loss in the coil. Therefore, higher \( \omega \) would lead to more efficient coil performance. Further, if data transfer is also required, a higher frequency can afford higher bandwidths. However, this would lead to weaker reactive magnetic fields (Eqns. 1.5.2 and 1.5.3) for a given distance between the coils. Further, this would also lead to more tissue heating near the surface of the body and higher energy loss in tissues due to the higher induced electric fields. It is worth pointing out that the conductivity of most tissues increases with the frequency. Based on these considerations, typical biomedical telemetry links range between hundreds of kHz and about 10 MHz.

1.5.1.4 Electromagnetic Effects on tissues

The electromagnetic effects have been discussed in detail in section 1.4. The incident EM energy parameters such as power and frequency should be thoroughly evaluated for safety compliance using measurements and simulation-based experiments, and, attention must also be paid to possible high frequency harmonics that might result from the source.

In general, there are several interlinked factors that dictate the efficacy and safety of an inductive telemetry link: from dimensions, to shape, to materials and power parameters. This warrants an iterative optimization process to evaluate the design tradeoffs before fully operational implantation in patients.
1.5.2 Integrated Inductors

An inductor is a key component in RF circuits, and is typically used in matching circuits for components such as VCOs, LNAs, and mixers. With the drive towards the fabrication of fully integrated chips for wireless communications operating in the GHz regime, a critical requirement is the integration of inductors with high-Q and high resonance frequencies. To achieve this end, the key problems encountered are the resistive losses in the conductors and the substrate losses due to eddy currents.

Innovative methods utilizing MEMS techniques for 3-D inductors have been used by several research groups. A method called Plastic Deformation Magnetic Assembly (PDMA) was developed [37] which offers high inductance values with relatively smaller footprints. Yoon et. al. fabricated 3-D metal structures including suspended spiral inductors, and solenoidal inductors using a thick metal surface micro-machining technology [38]. For Biomedical telemetry systems, integrated inductors provide some significant advantages such as smaller overall size of the implanted system, fewer interconnect lines, and the possibility of a compact implantable system. Since ideal frequencies of operation are generally below 10 MHz, the high frequency effects, seen at RF frequencies, are small.

For a retinal prosthesis to have a high data bandwidth, a mutually exclusive power and data transfer via a low frequency inductive link and a high frequency microwave telemetry link, respectively is proposed in [39]. To that end, in this research, a novel 3-D inductor geometry has been considered which would allow relatively higher conductance than a planar inductor for a given surface area, and would allow the integration of a patch antenna (for microwave frequency data transfer) and the stimulator chip for data processing and driving the electrode array.
1.6 Overview of this Dissertation

This work is primarily concerned with the development and application of numerical computation methods and modeling schemes to accurately characterize the electromagnetic and thermal safety of biomedical implants, especially the 2nd generation 60-electrode epi-retinal prosthesis. The 3-D FDTD method in a D-H formulation [40] was used for the electromagnetic computations at the operating frequency of 10 MHz. An accurate FDTD source modeling scheme was developed and verified analytically. This study was extended to a number of frequencies between 2 MHz and 20 MHz to obtain the electric field trends which can be used to obtain reasonably accurate induced SAR values at any input parameters of frequency and power as long as near field conditions are valid.

The traditional ADI [42] FDTD scheme was used to compute the thermal elevation in the human head due to the operation of the retinal prostheses. Full human head models were also used to study the thermal effects of the retinal implant under the effect of MRI fields [41]. Further, a novel scheme incorporating the 3-D explicit and ADI schemes was developed which provided significant improvement in simulation times.

Finally, an experimental effort was made towards the fabrication of a novel 3-D inductor geometry which has the potential to allow the integration of the stimulator chip and an antenna (for RF data transfer) on the same substrate. A process flow including wet isotropic and anisotropic etching of silicon substrates and the application of positive tone electrophoretic photoresist, PEPR2400 was developed to enable patterning of lines on deeply etched inclines.
Chapter 2


2.1 Introduction

In this work, the Finite-Difference Time-Domain (FDTD) method has been used to solve Maxwell’s equations. It is easy to implement, offers a time-domain full-wave solution for complex structures and is readily compatible with MRI-based voxelized models of the human body. A source of error in the FDTD method is its incapability to resolve rounded corners, where voxelized discretization leads to stair-casing errors which can be significant in scenarios such as: interfaces of materials with large differences in dielectric properties; or regions near energy sources. A way to reduce these errors is to increase the resolution to adequately resolve the corners, but this will adversely affect the simulation efficiency in two ways: first, since the time step is directly proportional to the smallest dimension, the simulation time will be longer, and second, the resulting computational model will be larger. However, the benefit of problems in bio-electromagnetics (BEM), as in this work, is that the quantities of interest, such as the Specific Absorption Rate (SAR) and the induced current densities are required as averages over substantially
large volumes/areas, which tend to suppress the local maxima or minima due to the stair-casing errors. Further, variants of FDTD, such as expanding-grid, or multi-resolution methods can be used which help to locally resolve small structures without letting the problem become too large.

To solve the Bio-Heat Equation (BHE), the most common finite-difference (FD) method used is the conditionally stable explicit method since it is very easy to implement, and computations are reasonably fast for computing temperature increases due to external RF sources such as cell-phones. The unconditionally stable Alternating-Direction Implicit (ADI) method is also commonly used which can afford almost arbitrarily large time steps ($\Delta t$), and is limited by the accuracy rather than the stability.

In this chapter, first the FDTD methods used for computing electromagnetic energy spread will be described, followed by the methods for computing the heat spread. Since FDTD has been around for over three decades and has a rich literature database for many applications, only an overview will be provided, with a concentration on the D-H FDTD formulation used in this work. For the BHE, even though the mathematical model is more than 50 years old, the treatment of bio-implantable devices is relatively uncommon. A discussion of the formulation of the BHE, FD implementation, errors, and a new hybrid scheme involving the explicit and the ADI method will be presented.

### 2.2 The FDTD Method For Electromagnetics

#### 2.2.1 Yee’s Algorithm and FDTD

The fundamental FDTD algorithm was first described by Yee in 1966 [43]. However, the term Finite-Difference Time-Domain was coined by Taflove in 1970s, and he also outlined its usefulness for a multitude of electromagnetic problems [44]. The FDTD method is a direct representation of the maxwells curl equations in a finite difference form and solves for the electric and magnetic fields in space and time in a leapfrog manner. In the algorithm, the field components are not
co-located, instead they are offset in space as defined by the Yee grid (figure 2.1). Writing the Maxwell’s time-domain equations in differential form:

\[ \nabla \times \overrightarrow{H} = J_E + \frac{\partial D}{\partial t} \]  

\[ \nabla \times \overrightarrow{E} = J_M - \frac{\partial B}{\partial t} \]  

where \( J_E \) is the electric conduction current density, and \( J_M \) is the magnetic conduction current density. The central finite-difference scheme, as obtained from the Taylor series expansion, to discretize the above equations is of the form:

\[ \frac{\partial V}{\partial x} = \frac{V_{k+1}^n - V_{k-1}^n}{2(\Delta x)} + O(\Delta x^2) \]  

(2.2.3)

for the spatial partial derivatives, and

\[ \frac{\partial V}{\partial t} = \frac{V_{k+1}^{n+1} - V_{k}^{n-1}}{2(\Delta t)} + O(\Delta t^2) \]  

(2.2.4)

for the time derivative. \( V \) is the electric or magnetic field, \( \Delta x \) is the space resolution, \( \Delta t \) is the time resolution, the superscript indicates the time instant \((k,\Delta t)\) at which the field value is indicated and the subscript indicates the spatial position \((n,\Delta x)\). Both the formulations are second order accurate approximations, that is, the terms of order \((\Delta x^2)\) and \((\Delta t^2)\) are discarded from the finite difference formulations obtained from Taylor series expansions for equations 2.2.3 and 2.2.4. Higher order FDTD schemes, wherein more terms are included from the Taylor’s series expansion thus leading to smaller truncation errors but more complicated and more computation intensive schemes, have also been implemented. For a detailed description of the FDTD method, relevant finite difference equations, errors, applications, and a comprehensive literature review, the reader is referred to the book by Taflove [44].

2.2.2 The D-H Formulation

The FDTD method in its original form has as unknowns the electric fields (E) and magnetic fields (H). Sullivan modified the original FDTD method by
computing the displacement field (D-field), instead of the E-field, from the H-field \[45\]. The E-field is then computed using the D-field values. This formulation is called the D-H FDTD method. The fundamental equations for this formulation are:

\[ \frac{\partial D}{\partial t} = \nabla \times \vec{H} \]  
\[ D(\omega) = \varepsilon^{*}(\omega) \cdot E(\omega) \]  
\[ \frac{\partial \vec{H}}{\partial t} = \nabla \times \vec{E} \]

where \(\varepsilon^{*}(\omega)\) is the frequency dependent loss term for the material and factors in the permittivity and the conductivity. Since the permeability \(\mu_{0}\) and permittivity \(\epsilon_{0}\) of free space are orders apart, the E-field and H-field values in the finite-difference equation for the above set of equations differ by orders of magnitude. Therefore, the E-field, and the D-field are normalized in the following manner \[45\]:

\[ \tilde{E} = \sqrt{\frac{\epsilon_{0}}{\mu_{0}}} \cdot \vec{E} \]  
\[ \tilde{D} = \sqrt{\frac{1}{\epsilon_{0} \mu_{0}}} \cdot \vec{D} \]

Expanding for x-,y- and z-directions, we will get 6 coupled equations which can be written in finite-difference format based on the Yee grid (figure 2.1). For example, from equation 2.2.10, for the D-field along x-direction, we get:

\[ \frac{\partial \tilde{D}_{x}}{\partial t} = c \left( \frac{\partial H_{z}}{\partial y} - \frac{\partial H_{y}}{\partial z} \right) \]
Figure 2.1: The Yee Grid as used in this work. The locations of the D-field components are identical to that of the corresponding E-fields [45].

\[
\frac{D_{x(i+\frac{1}{2},j,k)}^n + D_{x(i+\frac{1}{2},j,k)}^{n-\frac{1}{2}}}{\Delta t} = c\left\{ \frac{H_{z(i+\frac{1}{2},j+\frac{1}{2},k)}^n - H_{z(i+\frac{1}{2},j-\frac{1}{2},k)}^n}{\Delta y} - \frac{H_{y(i+\frac{1}{2},j,k+\frac{1}{2})}^n - H_{y(i+\frac{1}{2},j,k-\frac{1}{2})}^n}{\Delta z} \right\} 
\]

where \( c = \frac{1}{\sqrt{\epsilon_0\mu_0}} \) is the speed of light in free space.

The D-H formulation comes at the cost of extra memory requirements and computations per time step. Its benefit lies in the ease of PML (discussed in section 2.2.3) implementation in the case of lossy media such as the human body. Detailed equations and methodology and C language implementations in 1-, 2- and 3- dimensions can be found in the book by Sullivan [45].
2.2.3 Absorbing Boundary Conditions

Most electromagnetic (EM) problems require the computation of field values in an unbounded domain. Even in bio-EM problems at very high resolutions, often the region of interest is very small. In such situations, it is prohibitively wasteful and mostly impossible to compute field values for the full domain. This is circumvented by using two distinct regions of computation: the inner maxwellian region where the normal intended material properties and maxwell’s equations are used; and an outer region, where, via analytical means or by using artificial lossy media, the outgoing EM energy is absorbed [44]. The effectiveness of the absorbing region lies in minimizing the reflected energy, thus simulating natural conditions.

In this work, the Perfectly Matched Layer (PML) first developed by Berenger [46] have been used to limit the computational space. The function of the PML is to provide a lossy medium such that the incoming EM wave gets attenuated without reflection. This is done by providing the PML with artificial electrical and magnetic conductivities. However, if an abruptly high conductivity is applied at the beginning of the layer, numerical instability could result, therefore generally a graded conductivity profile is used which increases from that of free space to its highest value. Critical parameters such as the conductivity profile, maximum conductivity values and the PML thickness are often optimized empirically.

In the first work on PML [46], a split-field approach was used where, in the PML, the fields were split into two orthogonal components, resulting in 12 components in coupled partial differential equations (PDEs). Equivalent unsplit-field PML implementations were then proposed for the conventional FDTD and the D-H FDTD [45].

For the D-H FDTD formulation used in this work, re-writing equation 2.2.13 with the PML parameters:

\[ j\omega(1 + \frac{\sigma_x(y)}{j\omega\varepsilon_0})(1 + \frac{\sigma_y(z)}{j\omega\varepsilon_0})(1 + \frac{\sigma_z(x)}{j\omega\varepsilon_0}).D_z = c.(\frac{\partial H_z}{\partial y} - \frac{\partial H_y}{\partial z}) \]  

(2.2.15)

where the time derivative \(\frac{\partial}{\partial t}\) is replaced by \(j\omega\) in frequency domain. In the propagating volume, the fictitious conductivities, that is \(\sigma_x, \sigma_y\) and \(\sigma_z\) are all
equal to zero leading to equation 2.2.13. Inside the PML, the value increases from zero to a maximum and follows a gradual profile. Note that the PML parameters:

\[
(1 + \frac{\sigma_x(x)}{j\omega\epsilon_0})^{-1}(1 + \frac{\sigma_y(y)}{j\omega\epsilon_0})(1 + \frac{\sigma_z(z)}{j\omega\epsilon_0})
\]  

are obtained based on two fundamental conditions proposed by Sacks [47]. Similar attenuating parameters can be obtained for the other 5 coupled equations.

As mentioned in section 2.2.2, the D-H FDTD formulation used in this work comes at an extra memory and computation cost. The benefit of this algorithm is in the ease of implementation of the PML in the situation where any arbitrary dielectric may be immersed in the PML. This stems from the fact that the PML is independent from any background material as can be seen from equation 2.2.15 and once the appropriate parameters are established, no change whatsoever is required for any dielectric.

2.2.4 Errors and Stability

The FDTD method has some inherent errors associated with it. First is the truncation errors stemming from the FD approximation of the PDE from the Taylor series expansion. Then, there exist numerical dispersion errors due to the departure of the numerical phase velocity from that of the physical phase velocity. Further, a traveling wave tends to travel faster along the diagonal of the computational lattice as compared to the cartesian axes [44]. The dispersion errors are also proportional to the space and time resolution. However, these errors typically affect the phase values at the observation points, and for most BEM problems, only the magnitudes of the field values are generally of interest.

The traditional FDTD method is a conditionally stable explicit method, with the solution becoming unstable if the time step is more than the upper bound, also called the Courant-Fredrich-Lewy (CFL) limit and given by

\[
\Delta t \leq \frac{1}{c} \cdot \frac{1}{\sqrt{\frac{1}{\Delta x^2} + \frac{1}{\Delta y^2} + \frac{1}{\Delta z^2}}}
\]  

where \( \Delta x, \Delta y, \Delta z \) are the spatial resolutions, and \( v \) is the phase velocity.
2.3 Finite-Difference Methods for the Bio-Heat Equation

The basic idea of the finite difference method has already been introduced earlier in this chapter, the only difference being that the variable ‘V’ in equation 2.2.3 will now represent temperature(T).

2.3.1 Bio-Heat Equation

For any body, the temperature variation with time can be described by the partial differential equation (PDE):

\[ C \rho \frac{\partial T}{\partial t} = \nabla \cdot (\kappa \nabla T) \pm S \]  

(2.3.1)

where \( C \) is the specific heat, \( \rho \) is the mass density of the material, \( \kappa \) is the thermal conductivity, \( T \) is the temperature, and \( S \) is a heat source or sink (positive or negative). To include biological effects, Pennes [9] incorporated two additional source terms modeling the effect of blood flow and the metabolic heat production. Additionally, electromagnetic energy (characterized by the Specific Absorption Rate (SAR)) induced heating and power dissipated by implanted components can also be directly included, giving:

\[ C \rho \frac{\partial T}{\partial t} = \nabla \cdot (\kappa \nabla T) + A - B(T - T_B) + \rho \text{SAR} + P_I \]  

(2.3.2)

where \( A \) is the rate of heat production due to metabolic processes, \( B \) is the blood perfusion constant which quantifies the amount of blood flow to the tissue under consideration, and \( T_B \) is the temperature of the blood which flows through the tissues. \( T_B \) is generally assumed to be \( 37^\circ \) C. \( \text{SAR} = \frac{\sigma}{\rho} |E|^2 \), where \( \sigma \) is the electrical conductivity, and \( |E| \) is the electric field magnitude. \( P_I \) is the power dissipation per unit volume by any implanted device.

2.3.2 Finite-Difference (FD) formulation of the BHE

One way of obtaining the FD scheme for equation 2.3.2 is to directly approximate the PDE. Another method to obtain the FD scheme is using the conserva-
tion law approach, which is based on the derivation from the physical laws of heat transfer. The latter approach has been followed in this work.

The formulation of the bioheat equation follows the method employed in [48], where the concept is to balance the heat flow in a computational voxel. Metabolic heat rate, blood perfusion, power dissipation, and heat transfer via conduction from adjacent volumes are considered sources (or sinks) of energy depending on their sign. Specifically, considering a voxel-centered grid such as that shown in figure 2.2a,

1. \( Q_A = A \cdot (\text{Vol}) \cdot (\Delta t) \) Joules : Heat due to metabolic heat rate.

2. \( Q_B = B \cdot (T_B - T) \cdot (\text{Vol}) \cdot (\Delta t) \) Joules: Heat due to blood perfusion of the tissue

3. \( Q_{\text{Implant}} = P_{\text{Implant}} \cdot (\text{Vol}) \cdot (\Delta t) \) Joules : Heat due to implant

where \( \text{Vol} = \Delta x \cdot \Delta y \cdot \Delta z \) is the volume of the voxel in the 3-D computational grid, and \( \Delta t \) is the time in seconds. Besides the sources within the cells, there is transfer of heat from adjacent voxels via conduction. Based on the thermal conductivities of the tissue, the conduction model can be characterized as a thermal resistance network, as shown in figure 2.2c. The capacitor C in the figure, is the thermal capacity and is an open circuit at steady state. Fourier’s law, given by

\[
(\Delta T) = Q' R_{th}
\]  

(2.3.3)
can be used to obtain \( Q_{\text{conduction}} \). Note that \( Q' = (\Delta Q_{\text{conduction}}/\Delta t) \), \( (\Delta T) \) is the temperature change in time \( \Delta t \), and \( R_{th} \) is the thermal resistance, which for two adjacent voxels (figure 2.2b,c) is given by

\[
R_{th}(i + 1) = \left( \frac{1}{2 \Delta y_{i,j,k} \cdot \Delta z_{i,j,k}} \right) \frac{\kappa_{i,j,k} \cdot \Delta x_{i+1,j,k} + \kappa_{i+1,j,k} \cdot \Delta x_{i,j,k}}{\kappa_{i,j,k} \kappa_{i+1,j,k}}
\]  

(2.3.4)

where \( \kappa_{i,j,k} \) is the thermal conductivity and \( \Delta x_{i,j,k}, \Delta y_{i,j,k}, \Delta z_{i,j,k} \) are the dimensions of the voxel centered at the grid point \((i, j, k)\).

Equating all the sources of heat with the amount of heat stored or lost by unit volume of tissue in time \( \Delta t \), we get

\[
\rho C \cdot \text{Vol} \cdot (\Delta T) = Q_A + Q_B + Q_{\text{Implant}} + Q_{\text{conduction}}
\]  

(2.3.5)
Figure 2.2: (a) The voxel-centered computational grid. (b) The thermal resistance between adjacent voxels at (i,j,k) and (i+1,j,k). (c) The thermal RC network at a voxel node (i,j,k).

Note that the $Q_{\text{conduction}}$ term has to be computed for all the 6 faces of the voxel under consideration. The tissue-air interface is modeled as a convective boundary using the formula:

$$Q_{\text{boundary}} = H_A(T_A - T).S.(\Delta t) \quad (2.3.6)$$

where $H_A$ is the convective coefficient for the tissue-air interface and $T_A$ is the ambient temperature and is assumed to be equal to 24° C, and $S$ is the surface in contact with air.
2.3.3 Numerical Methods

2.3.3.1 Explicit Method

Numerical methods to solve the bio-heat equation can be either time-marching methods (parabolic PDE’s) or iterative methods for steady state (elliptic PDE’s). Equation 2.3.2 is a parabolic equation and will become an elliptic equation if its left hand side is zero. Here, we will concentrate on time-marching methods.

Let the voxel at which the BHE is being solved be at \((i, j, k)\) for time instant \(n + 1\). Define

\[
\delta_x^{2,n} = \lambda_1 T_{i-1,j,k}^n - (\lambda_1 + \lambda_2) T_{i,j,k}^n + \lambda_2 T_{i+1,j,k}^n
\]

(2.3.7)

where \(\lambda_1 = \frac{2 \kappa_{i,j,k} \kappa_{i-1,j,k}}{\Delta x_{i,j,k}(\kappa_{i,j,k} \Delta x_{i-1,j,k} + \kappa_{i-1,j,k} \Delta x_{i,j,k})}\), and

\[
\lambda_2 = \frac{2 \kappa_{i,j,k} \kappa_{i+1,j,k}}{\Delta x_{i,j,k}(\kappa_{i,j,k} \Delta x_{i-1,j,k} + \kappa_{i+1,j,k} \Delta x_{i,j,k})}.
\]

Let \(\delta_y^{2,n}\) and \(\delta_z^{2,n}\) be similar definitions for derivatives along the \(y\)- and \(z\)-directions. In an explicit finite-difference scheme, the BHE(equation 2.3.2) can thus be written as:

\[
C \rho \frac{T_{i,j,k}^{n+1} - T_{i,j,k}^n}{\Delta t} = \delta_x^{2,n} + \delta_y^{2,n} + \delta_z^{2,n} + A_{i,j,k}(T_{i,j,k}^n - T_B) + \frac{N_{\text{air}}}{d} H_A (T_A - T_{i,j,k}^n)
\]

(2.3.8)

where \(N_{\text{air}}\) is the number of faces of the voxel at \((i, j, k)\) which are in contact with air, and \(d\) is the voxel dimension along the area vector of the surface in contact with air. Note that for faces in contact with air, the corresponding \(\lambda\) in equation 2.3.7 will be zero.

For simplicity, assuming a uniform resolution grid scheme (\(\Delta x = \Delta y = \Delta z\)), the stability analysis of the equation 2.3.8 leads to the condition:

\[
\Delta t_{\text{max}} = \frac{1}{\frac{N_{\text{int,}x}}{C \rho \Delta x} + \frac{N_{\text{ext,}A} H_A}{C \rho \Delta x} + \frac{B}{C \rho}}
\]

(2.3.9)

The explicit method is easy to implement and is very useful for computations on low resolution models and models where the the thermal conductivities are relatively small (for example the human tissues). However, to accurately model implantable components such as the retinal prosthesis and the curved retina and sclera of the eye, it is important that resolutions as high as at least 0.25 mm be
used. Using the explicit method in such cases would lead to a prohibitively large number of simulation-steps.

2.3.3.2 Alternating-Direction Implicit (ADI) Method

Implicit methods have the advantage that they usually do not have a time step($\Delta t$) constraint, and it is generally much higher than that of the explicit method even if there is a restriction. Often, the constraint is because of the accuracy rather than the stability. Amongst the most efficient methods for solving the BHE equation is the Alternating-Direction Implicit (ADI) method. It was first developed in the 1950s [42] to solve heat flow problems for an oil-exploration company. Over the years, several enhancements and modifications have been done for many different applications. However, in this work, the ADI method described by Douglas in one of the first publications’ has been closely followed for computing the thermal spread in the human head.

The basic idea behind the ADI method is to divide the computation for each time step into 3 sub-steps (for 3-D). In each sub-step, the computation for one co-ordinate direction is done implicitly while it is done explicitly for the other two. The benefit of this method is that it leads to a tri-diagonal matrix at each sub-step. For even moderately sized models, the inversion of 3 tri-diagonal matrices is more efficient than the inversion of a single fully implicit matrix.

Following the same notations as earlier in this section, the three sub-steps can be written as [42]:

SUB-STEP 1:-

$$C \rho \frac{T_{ijk}^{n+1*} - T_{ijk}^n}{\Delta t} = \frac{\delta_x^2 n+1*}{2} + \frac{\delta_x^2 n}{2} + \delta_y^2 n + \delta_z^2 n + A + B (T_{ijk}^n - T_B) + \frac{N_{air}}{d} . H_A (T_A - T_{ijk}^n)$$

(2.3.10)

SUB-STEP 2:-

$$C \rho \frac{T_{ijk}^{n+1**} - T_{ijk}^{n+1*}}{\Delta t} = \frac{\delta_y^2 n+1**}{2} - \frac{\delta_y^2 n}{2}$$

(2.3.11)

SUB-STEP 3:-

$$C \rho \frac{T_{ijk}^{n+1} - T_{ijk}^{n+1**}}{\Delta t} = \frac{\delta_z^2 n+1}{2} - \frac{\delta_z^2 n}{2}$$

(2.3.12)
where $T_{ijk}^{n+1*}$ and $T_{ijk}^{n+1**}$ are just intermediate values which are ‘corrected’ at steps 2 and 3, and not the values at intermediate physical times.

The growth factor (also called amplification factor), $g(\xi)$ can be defined as the factor by which the amplitude of each constituent frequency increases in advancing the solution by a time step. In fact, the condition for stability for the explicit method in equation 2.3.9 is obtained by evaluating $g(\xi) = 1$. For the FD scheme described in equations 2.3.10, 2.3.11 and 2.3.12, the growth factor comes out to be

$$g(\xi_x, \xi_y, \xi_z) = \frac{1 + (\mu_x + \mu_y + \mu_z) + (\mu_x\mu_y + \mu_x\mu_z + \mu_y\mu_z) - \mu_x\mu_y\mu_z}{1 - (\mu_x + \mu_y + \mu_z) + (\mu_x\mu_y + \mu_x\mu_z + \mu_y\mu_z) - \mu_x\mu_y\mu_z}$$

(2.3.13)

where

$$\mu_i = -4\eta\sin^2\left(\frac{\delta\xi_i}{2}\right)$$

(2.3.14)

if the model is homogeneous, and $i=x,y,z$; and $\eta = \frac{\lambda}{c_p}$. In this case, the growth factor, $g \leq 1$, and the scheme is unconditionally stable. If the model is heterogeneous, writing in a form similar to the previous equation for x-direction for a grid point, say (p,q,r)

$$\mu_x = -4\eta_1\sin^2\left(\frac{\delta\xi_x}{2}\right) + (\eta_1 - \eta_2)(1 - e^{j\delta\xi_x})$$

(2.3.15)

which can lead to instabilities for large time steps. However, for the inhomogeneity in tissues, it can afford very large time steps, since their thermal properties are of the same order. In fact, for relatively lower resolutions, even for medical implants inside the human body, sufficiently large time-steps can be afforded. This is illustrated in figure 2.3 for a simplistic 4-turn coil implanted in the anterior region of the eye. This also validates the ADI method.

However, in the situation of a medical implant with thermal conductivities varying over 2 orders of magnitude at a high resolution, for example, the presence of a commonly used biocompatible material, silicone($\kappa = 0.18$), adjacent to metal coil($\kappa = 72$), $\Delta t$ can be severely restricted. Therefore, while the ADI method is ideal for calculating temperature increments due to electromagnetic energy deposition for external sources such as cell phones, it is not as efficient in the case
Figure 2.3: (a) Figure of just the eye with the coil. Note that only position b (square coil in the anterior position of the eye) was considered for the comparison. The simulation model was uniform resolution at 0.5 mm. (b) Comparative time-domain plots of the temperature increase for the two methods at a position on the coil. They show very good agreement.
of implanted devices. In such cases, it is useful to have an *a priori* estimate of the maximum allowable time step. Finding an analytical solution for the growth factor can become quite tedious if it is possible; instead, it was found that in the computational model, if the largest discontinuity is known (i.e. maximum variation in $\lambda$), a reasonable estimate of the $\Delta t_{ADI}$ can be obtained using equation 2.3.13 at just one point in the computational grid. By computing the growth factor by sweeping $\theta = \xi \delta$ in the interval $[-\pi, \pi]$, the maximum $\Delta t$ for which $g \leq 1$ can be obtained quickly.

### 2.3.3.3 Hybrid explicit-ADI Method

As mentioned earlier in this section, the ADI method allows arbitrary large time steps for homogeneous and mildly heterogeneous models, but can be severely restricted for models with large discontinuities in the thermal properties. In fact, since the computation cost per time step for the ADI method is significantly larger than that of the explicit method, in some situations, the net simulation time gain of the ADI method over the explicit might be small.

To obtain the benefit of the higher time steps afforded by the ADI method and the lesser computation cost of the explicit method, a new hybrid method was developed incorporating both; where-in two computation domains were used. Regions with medical implants with thermal conductivity discontinuities were placed in the ADI domain, while the pure tissue regions were solved using the explicit method. A further advantage of using this method is that multi-resolution voxels can be used in the explicit domain. In this case, the time step is given by:

$$
\Delta t = \text{minimum}(\Delta t_{explicit}, \Delta t_{ADI}) \tag{2.3.16}
$$

In the case of a pure tissue model, since the thermal conductivity is small, $\Delta t_{explicit} > \Delta t_{ADI}$. The numerical process per time step is as follows:

1. Computation over the ADI domain

2. Update boundaries between the two domains explicitly
3. Computation over the explicit domain

Validation and results of the hybrid method are described for the 60-electrode prosthesis in chapter 4.
Chapter 3

Electromagnetic Effects in the Human Eye and Head due to the Telemetry Link of an Epiretinal Prosthesis

3.1 Introduction

In this chapter\(^1\), the fields induced in the human head by the wireless telemetry system used for Second Sight Medical Product, Inc.’s epiretinal prosthesis system are characterized for compliance testing with international safety standards using 3D finite-difference time-domain (FDTD) method in D-H formulation. The specific system under consideration utilizes an inductive link with a primary coil mounted on the subject’s eyeglasses and a secondary coil which is strapped on the eye, over the sclera. The Specific Absorption Rate (SAR) and the current density have been obtained computationally for different relative positions of the primary and secondary coils to account for the relative misalignment of the two

\(^1\)Portions of this chapter have been taken from: V. Singh et al, "Specific Absorption Rate and Current Densities in the Human Eye and Head Induced by the Telemetry Link of an Epiretinal Prosthesis", IEEE Trans. Ant.& Prop, 2009
due to the movement of the eye with the implant. For a peak normalized current of 0.62 A in the primary coil at 10 MHz, the highest peak 1-g SAR was found to be 0.45 W/Kg and the maximum root mean square (RMS) current density averaged over a $1cm^2$ area was found to be $16.05\, Am^{-2}$, both of which are within the limits imposed by IEEE and ICNIRP safety standards. Simulations between 2 MHz and 20 MHz indicated that the induced electric field values scale well with frequency, thus providing guidelines for the determination of the final frequency and input power requirements of operation for the telemetry system to meet safety standards.

A detailed discussion of the thermal effects of bio-implants can be found in [29]. Specifically, a thorough treatment of the thermal effects of the stimulator chip and the secondary coil for this particular prosthesis system will be carried out in the next chapter. The current generation of the prosthesis design, by Second Sight Medical Products (SSMP), Inc. under clinical trials is the ArgusII, a 60-electrode device. As the electrode count increases, the power requirements and the ensuing heat generated may become the primary issues for such devices. Another possible unwanted effect due to the telemetry system is parasitic neuronal stimulation.

Some earlier investigation of the electromagnetic effects (only SAR) of the retinal prosthesis was carried out in [49, 50] for a 2-D head model at 2 MHz and for a 3-D head model at 10 MHz [51]. However, in both, the primary coil was modeled as a single loop. This work differs from the previous work on the following accounts: first, the actual spiral primary coil and implanted components have been discretized and merged with the head model; second, the primary coil is mounted on the patient’s eyeglasses, and positioned on the side of the eye parallel to the forehead, while the secondary coil is strapped on the eye, over the sclera; third, induced current densities have also been computed; and fourth, different orientations and distances of the primary coil have been considered for different frequencies for a more comprehensive characterization of the electromagnetic effects of the system.

This chapter consists of 6 sections. Section II describes the head model used. This is followed by a discussion, in Section III, of the different orientations of
Table 3.1: Conductivity of key tissues at different frequencies

<table>
<thead>
<tr>
<th>Tissue</th>
<th>2 MHz</th>
<th>5 MHz</th>
<th>10 MHz</th>
<th>15 MHz</th>
<th>20 MHz</th>
</tr>
</thead>
<tbody>
<tr>
<td>Muscle</td>
<td>0.55</td>
<td>0.59</td>
<td>0.62</td>
<td>0.63</td>
<td>0.64</td>
</tr>
<tr>
<td>Sclera</td>
<td>0.69</td>
<td>0.76</td>
<td>0.8</td>
<td>0.82</td>
<td>0.83</td>
</tr>
<tr>
<td>Skin</td>
<td>0.04</td>
<td>0.11</td>
<td>0.2</td>
<td>0.25</td>
<td>0.29</td>
</tr>
<tr>
<td>Retina</td>
<td>0.69</td>
<td>0.76</td>
<td>0.8</td>
<td>0.82</td>
<td>0.83</td>
</tr>
<tr>
<td>Vitreous Humor</td>
<td>1.5</td>
<td>1.5</td>
<td>1.5</td>
<td>1.5</td>
<td>1.5</td>
</tr>
</tbody>
</table>

the primary coil with respect to the head considered in this paper. In section IV, a detailed discussion of the method of excitation is provided. The results are presented in Section V, followed by a discussion and concluding remarks in Section VI.

### 3.2 Head and Implant Models

The starting model of the human head was obtained in the form of 1-mm resolution slices from the Visible Man Project from the National Library of Medicine [31], and then re-discretized to a resolution of 0.3 mm using linear interpolators (figure 3.1(a)). Figures 3.1(b) and (c) show a cross-section of the starting head model, and the extracted region used for the simulations. The implant and primary coil models were then imported from CAD file formats, discretized and incorporated in the head model. Figure 3.2 shows just the eye with the implant positioned on it. The dielectric properties of the tissues at different frequencies were obtained from [34]. Table 3.1 lists the properties of some of the key tissues.

### 3.3 Modeling

It is envisioned that the external components including the primary coil will be placed on a pair of eyeglasses worn by the subject. Based on this, we have
Figure 3.1: (a) A 3D model of the head and the approximate location of the primary coil (b) Slice of the human head indicating the extracted region for the FDTD simulations. (b) The extracted model with the implanted secondary coil. The size of the extracted region at 0.3 mm resolution was $180 \times 240 \times 200$. 
Figure 3.2: Figure (a) shows the intended position of the implant and (b) shows the discretized 3-D model of the package around the eyeball.

Figure 3.3: The different orientations of the eye and primary coil considered investigated the energy absorption and induced currents for different orientations and distances of the primary coil with respect to the forehead near the right eye.

The primary coil position is not expected to be constant during operation, although the position of the secondary coil will be fixed with respect to the eye. Figure 3.3 shows the different orientations of the primary coil considered for the simulations. These orientations were used for two distances between the coil and the forehead.

With reference to figure 3.3, position (A) is when the primary is parallel to the forehead. Position (B) is when the primary is turned away by approximately 25° with the lower end as the pivot in the figure. This leads to a distance slightly higher than the original ‘d’. The third position (C) is when the eye is rotated by
20° (eye looking left). The distance ‘d’ was 2.07 cm, 2.8 cm, 3 cm for (A), (B), and
(C) respectively. All these three positions were repeated by moving the primary
coil along positive-x direction to be almost in contact with the forehead leading
to distances of 1.1 cm, 1.6 cm, and 1.85 cm. These are referred to as (A’), (B’)
and (C’). Two obvious inferences can be made at the outset. First, the induced
E-fields for (B & C) and (B’ & C’) will be very similar since the position of the
forehead is unchanged. Second, the induced E-fields will be greater for (A’, B’, C’)
than that for (A, B, C), because of lesser distance from the primary coil. Note
that to ensure that coil voxels are not contiguous with tissue, the location of the
primary coil had to be adjusted along y- and z- directions as well.

The diameter of the primary coil filament was about 0.9 mm and the discretiza-
tion of its CAD model at 0.3 mm resolution led to a three-cell thick primary coil.
Since rotating the model by 25 degrees would lead to disconnects in the discretized
model, instead the head model was rotated by the required angle. For the case
where the eyeball is turned, only the eye tissues and the implant is rotated.

3.4 Primary Coil: Excitation and verification

The primary coil is spiral shaped with external diameter 39 mm and internal
diameter 21.5 mm and acts as the power and data source with data modulating
the relatively low-frequency power carrier. A current can be excited in one of
several ways in the FDTD method. For example, applying a voltage difference
using the finite-gap method will force a current through the coil. However, the
problem with this excitation with respect to the coil considered in this paper is
that at this resolution (0.3 mm), the turns are not smooth, leading to possible
disconnects in the staggered grid, which will effect the integrity of the current in
the coil. Another possibility is using the H-field curl around the coil. However, this
implicitly assumes that the magnetic field is identical at the points of excitation,
which leads to an inaccurate H-field distribution.

In this work, the method used to simulate a current in the coil is a D-field
(displacement field) source at each voxel representing the current in it. Assuming
Figure 3.4: (a) and (b) indicate the magnitude and direction (indicated by sign on the colorbar) of y- and z- components of the current

the coil is in the y-z plane, and based on a normalized current, \( D_y \) and \( D_z \) values were assigned to each coil voxel based on the angle made by that cell with the center of the coil. Another reasonable assumption made was that the current magnitude through each voxel was identical. Figure 3.4 indicates the D-field magnitudes (normalized to a peak value of 1 A) in the y and z-directions.

To verify the excitation, the E-field distribution in the absence of human tissue using FDTD was compared to that obtained using the method described by [52], in which the E-field is calculated at any point \( P(\tau) \) in space due to a current element, \( dI \), using

\[
dE = \frac{\mu_0 N (dI/dt) d\overline{l}}{4\pi R}
\]  

Where \( R \) is the distance between the current element and \( P(\tau) \), \( N \) is the number of turns, and \( dI/dt \) is the rate of change of current. Since we are interested in the peak E-field distribution, \( dI/dt = I_{\text{peak}} \omega \). Figure 3.5 compares the E-field distribution in a plane through the center of the coil.

A simple well known formula for the magnetic field intensity along the axis of a concentric current-carrying loop is

\[
H = \frac{IR^2}{2\sqrt{(R^2 + x^2)^3}}
\]  

where I is the current through each filament, x is the distance from the center
Figure 3.5: The Electric field magnitudes in V/m at 2 MHz along the cross-section containing the center of the spiral coil (a) Analytical, using equation 3.4.1 (b) with FDTD. Both results are without the head model of the loop and R is the mean radius of the coil. For N concentric loops, the H field could be summed up for N single turn loops of different radii.

A comparison of the H-fields using equation 3.4.2 with that from FDTD simulations is shown in figure 3.6b. The results for 5 MHz and 10 MHz are without the implant in the head model and indicate the relative invariance of the H-field with the frequency in its reactive field and the transparency of the tissues to the magnetic fields. The results for these two frequencies show good agreement with the formula-based result. A possible reason for the difference is the fact that the formula considers concentric loops of current as opposed to the spiral in the FDTD simulation. The third plot in figure 3.6b, and the figure 3.6a illustrate the effect of the secondary coil on the magnetic fields around the implant.
Figure 3.6: (a) The H-field distribution at a cross-section of the entire model (including head) for case A’ (shown in Figure 3.3a) (b) The variation of the H-field along the axis of the concentric loops (using formula) and the axis of the spiral (FDTD). For 5 and 10 MHz, the results are with the head model without the implant. In both the figures, the dip in the H-field magnitude can be seen at the location of the secondary coil.

### 3.5 Computational Results

The maximum E-field values were obtained in the insulation region around the secondary coil which behaves like a short circuit where the fields are almost zero. The largest E-field values in the tissues were obtained in the peripheral regions of the eye and that too in the cases when the coil was closest to the forehead. Because of the discretization of the implant, the curved surfaces are jagged, which is further pronounced by subsequent rotations for the different orientations. Numerical errors due to staircasing [53] at tissue-implant and tissue-air interface, and artifacts of the 2-equation-2-unknown method are the most likely causes for some single-voxel peaks. However, these local peaks get averaged out after 1-g
calculations.

Figure 3.7 illustrates the electric field, current density and the SAR distribution for cases B and B’. The peak 1-g SAR values for the different cases are listed in table 3.2. Since it is not possible to obtain exact cubes of mass equal to 1 gram, allowances of 10 percent (by weight) were made. Further, for regions near the air-interface, where the largest field values exist, the head boundary is very irregular leading to incorporation of a large number of air voxels which might lead to an underestimation of the 1g SAR. Therefore, to reduce the effect, while including most of the interface tissue voxels, it was decided to limit the maximum number of air voxels in the cube to 20 percent. Comparison with the IEEE standards [19] indicates that the telemetry system operating at 10 MHz is within safety limits. While the 10-g SAR has not been computed, it will obviously be lower than the 1-g SAR, leading to compliance with the ICNIRP imposed restriction of 2 W/Kg over 10g of tissue as well. Since the worst case is within the safety limits at 10 MHz, we will now concentrate on analyzing the electromagnetic effects of the telemetry system for different input parameters for the worst case orientation (A’).

Figure 3.7: The E-fields, current density and the SAR along a transverse cross section of the head model (all values are single-voxel)
Table 3.2: Peak 1-g SAR and Averaged Current Densities at 10 MHz for peak primary coil current of 0.62 A

<table>
<thead>
<tr>
<th>Cases</th>
<th>peak 1-g SAR W/Kg</th>
<th>Current Density(J),Am^-2 over 1 cm^2</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>0.12</td>
<td>6.4</td>
</tr>
<tr>
<td>B</td>
<td>0.08</td>
<td>5.2</td>
</tr>
<tr>
<td>C</td>
<td>0.08</td>
<td>5.2</td>
</tr>
<tr>
<td>A'</td>
<td>0.49</td>
<td>16.05</td>
</tr>
<tr>
<td>B'</td>
<td>0.11</td>
<td>7.7</td>
</tr>
<tr>
<td>C'</td>
<td>0.10</td>
<td>7.7</td>
</tr>
</tbody>
</table>

IEEE limit for SAR: 1.6 W/Kg
ICNIRP limit for rms J: f/500=20 Am^-2

In the near reactive field of a magnetic dipole where the interaction with tissues is mainly by magnetic induction, the induced E-fields nearly scale with frequency(E~jωB). This was verified for the model with simulations up to 20 MHz (figure 3.8). Also, since SAR (ω) ~ σ(ω) |E|^2, it was verified that SAR at any frequency can be obtained with reasonable accuracy by just scaling with conductivity and frequency. Figure 3.9 illustrates this. Obviously, the computed values in table 3.3 at the closest frequency to the frequency of interest should be used for scaling, for example, with reference to figure 3.9, by scaling to 2 MHz from 5 MHz, the scaling error reduces to 10%.

The current density distribution was similar to that of the SAR. The sclera, retina, and the vitreous humor have relatively large conductivities, leading to higher current densities than outside the eye, except in the skin at the air-interface (figure 3.7).

The rms current densities were obtained as averages over a cross-section of 1 cm^2 perpendicular to the current flow, as outlined in the ICNIRP guidelines [20], and are mentioned in table 3.3. The maximum averaged rms currents (J_{max,rms}) were obtained around the skin region. Allowances and assumptions similar to
Figure 3.8: The plot gives the variation of the E-field along a line parallel to the primary coil axis and through the head. Figure (a) is for different frequencies for a primary coil current of 0.62 A. In figure (b), the field values at all frequencies are scaled to 20 MHz. It indicates that a simple frequency scaling can be performed to obtain E-field values till at least 20 MHz.

Figure 3.9: Variation of the peak 1-g SAR with excitation frequency (current, I=0.62 A) for case A’. 'Er' indicates the error due to scaling from 10 MHz.
Table 3.3: Peak 1-g SAR and $J_{\text{max}, \text{rms}}$ with frequency for the worst case (A’) for peak primary coil current of 0.62 A

<table>
<thead>
<tr>
<th>Frequency</th>
<th>peak 1-g SAR</th>
<th>$J_{\text{max}, \text{rms}}$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>W/Kg</td>
<td>Am$^{-2}$</td>
</tr>
<tr>
<td>2 MHz</td>
<td>0.005</td>
<td>3.87</td>
</tr>
<tr>
<td>5 MHz</td>
<td>0.078</td>
<td>6.86</td>
</tr>
<tr>
<td>10 MHz</td>
<td>0.49</td>
<td>16.05</td>
</tr>
<tr>
<td>15 MHz</td>
<td>1.28</td>
<td>25.89</td>
</tr>
<tr>
<td>20 MHz</td>
<td>2.59</td>
<td>39.84</td>
</tr>
</tbody>
</table>

those for the peak 1-g SAR were made.

The location of the $J_{\text{max}, \text{rms}}$ up to 10 MHz was at the same location in the muscle tissue (location M) under the skin owing to its higher conductivity (table 3.1), even though the induced E-fields are lower (relative to skin). Also, since the conductivity of muscle varies little between 2 MHz and 20 MHz (0.55 to 0.64), the variation of the $J_{\text{max}, \text{rms}}$ is almost linear (dashed curve in figure 3.10). A relatively larger increase in the conductivity of skin leads to the $J_{\text{max}, \text{rms}}$ occurring at skin at frequencies of 15 MHz and greater. However, as mentioned in the introduction, for frequencies greater than 10 MHz, induced current densities is not a safety criterion. For $J_{\text{max}, \text{rms}}$ at the simulated frequencies, the scaling errors were within 5%. An important thing to note is that in the discussion in the last few paragraphs, the tissue with the maximum contribution to the volume (for peak-1gSAR) or area (for $J_{\text{max}, \text{rms}}$) has been considered as being representative of the volume or area. This assumption would not be valid if the region being considered was highly heterogeneous.

Specifically for the retina, Dimbylow [54] had used an averaging scheme using the absolute value of the current density. In the models used in this work, the retina was not clearly defined. Since the dielectric properties of the retina are identical to that of the sclera and since the retina is about 0.5 mm thick [reference], the 2 innermost layers of the sclera have been assumed to represent the retina.
Here, at 10 MHz, the retinal single-voxel $J_{\text{max, rms}}$ was found to be around 18 $A m^{-2}$, that too at a few cells with proximity (within 3 voxel radius) to the implant. Obviously, employing an averaging scheme here would yield a significantly lower value.

In a real world scenario, the energy source will not be ideal and higher order harmonics should be present. As long as the these harmonics are within the range of the quasi-static assumptions, reasonable estimates of the SAR can be made by extrapolating from the computed results.

### 3.6 Conclusion

In this work, we have investigated, via FDTD simulations, the safety of the telemetry link aimed for SSMP’s epi-retinal prosthesis system. A simple excitation scheme was used for the multi-turn spiral primary coil, wherein ideal D-field sources were used. Different orientations and distances, based on the expected movement of the subject’s eyeball and the primary coil were considered. For a normalized peak current of 0.62 A, the peak 1g SAR for the worst-case orientation at 10 MHz was 0.49 W/Kg and the peak rms current density averaged over an area of 1 $cm^2$ was 16.05 $A m^{-2}$, which are well within safety limits specified by IEEE and ICNIRP. It was verified that for similar models, the E-field values can
be obtained for different frequencies by just multiplying with the frequency scaling factor. Additionally, between 2 MHz and 20 MHz, the key compliance factor, the peak 1-g SAR can be obtained for any frequency within an error of 10%, with just the knowledge of the conductivity of tissues. The $J_{\text{max},\text{rms}}$ can also be obtained similarly within an error of 5%. To obtain these estimates, the frequency should be low enough for near field conditions, and the dielectric properties (especially conductivity) should not vary too much over the frequency range being considered.

By interpolating, for the same current in the primary coil, the peak 1-g SAR limit of 1.6 W/Kg would be crossed around 16 MHz. Further, at 10 MHz, the $J_{\text{max},\text{rms}}$ of 20 Am$^{-2}$ would be crossed for a primary coil current of about 0.78 A. The optimum frequency will ultimately be decided factoring in parameters such as coupling between the coils, induced current densities, and energy absorption in the tissues. Using the data from this work, for modifications in frequency and input power, the electromagnetic effects of the telemetry link for an epiretinal prosthesis on the human head can be expeditiously obtained.
Chapter 4

Thermal Effects of the implanted components of a 60-channel Epi-retinal Prosthesis

4.1 Introduction

In this chapter\(^1\), the thermal elevation in the human body due to the operation of a dual-unit epiretinal prosthesis to restore partial vision to the blind affected by irreversible retinal degeneration is presented. An accurate computational model of a 60-electrode device dissipating 97 mW power, currently under clinical trials is developed and positioned in a 0.25 mm resolution, heterogeneous model of the human head to resemble actual conditions of operation of the prosthesis. A novel simple finite difference scheme combining the explicit and the Alternating-Direction Implicit (ADI) method, discussed in the second chapter, has been developed and validated with existing methods. Simulation speed improvement up to 11 times was obtained for the the head model considered in this work with very good accuracy. Using this method, solutions of the bioheat

\(^1\)Portions of this chapter have been taken from: V. Singh et al, "On the Thermal Elevation of a 60-electrode Epi-retinal Prosthesis for the Blind", IEEE Trans. Biomed CAS,pp 289-300, Dec 2008
equation were obtained for different placements of the implant. Comparison with in-vivo experimental measurements showed good agreement.

Previous studies have shown that the most significant contribution to heat is by the implanted electronics [29]. Specifically, it was shown that the maximum temperature rise due to a 4-turn secondary coil carrying a current of 150 mA was about 0.41 degrees C, and it was about 3 degrees C on the surface of a chip dissipating 49.6 mW power located in the vitreous humor. Although the location of these components will have a noticeable impact on the final temperatures, these values provide realistic trends. Consequently, we focus on the thermal effects of the microchip package, including the active device and the secondary coil, positioned on the sclera. Other electromagnetic and thermal sources are given by the wireless telemetry link achieved through the coupled coils, and the power dissipation of the electrode array.

This chapter is structured as follows: Section II provides details about the computational models of the head and retinal prosthesis system used for the numerical simulations; Section III describes the computational method used and the assumptions made in this work. A simple method combining the explicit and the ADI method is also described which significantly reduces simulation times; Section IV provides results, including experimental verification in canines (performed by Second Sight Medical Products, Inc, based in Sylmar, California); and Section V presents the conclusions of this study as well as final remarks.

4.2 Computational Models

4.2.1 Head and Eye models

The basic model of the human head was obtained in the form of 1-mm resolution slices from the Visible Man Project from the National Library of Medicine (NLM) [31]. The eye region as obtained from the NLM was relatively coarse and was classified in only three tissues. Hence, an eye model with a higher degree of accuracy was developed and merged into the original head model to replace the
eye [50, 51].

Figure 4.1: Figure 1 (a) shows the intended position of the implant and 1 (b) shows the discretized 3-D model of the package around the eyeball.

Figure 4.2: Discretized model of coil placed in truncated eye model.

The 1-mm human head model was subsequently re-sampled to a resolution of 0.25 mm and 0.125 mm using linear interpolators. Bioheat equation solutions, without any synthetic heat sources, were obtained to find the initial temperature distribution in the head model. The initial temperature for this simulation was taken to be that of air (24°C). Smaller computational models, truncated at large distances from the implant regions were then extracted from the original model. The bioheat equation was then solved again over the extracted model without any external sources in order to obtain the reference initial temperature for this finer resolution model, which is then used in the simulation of the temperature increase.
due to the implant.

4.2.2 Implant models

For the implants, two different models have been considered. The first model comprises the chip package consisting of the stimulator chip and other passive elements. The model of the chip package, in a CAD file format, is imported in the computational grid and positioned on the sclera of the human eye as shown in Figure 4.1. The broad flat portion (insulation) on the side encloses the secondary coil.

The model of the tissue region around one eye, with the implant, was discretized to a resolution of 0.25 mm, leading to a model of dimensions $202 \times 202 \times 210$ cells. Figure 4.1a and 4.1b indicate the intended physical position of the implant and the modeled position, respectively. The chip (dissipating 81 mW) and the substrate for hybrid components (dissipating 16 mW) are placed on a substrate which forms the base of the chip package (mentioned in figure 4.1(a)).

For the rest of the chapter, ‘implant’ will refer to the system comprising the electronics and coil as indicated in figure 4.1, while ‘package’ will be used to refer to the chip and its packaging. Further, the eye tissues such as the sclera, choroid, and the retina, lying adjacent to the base of the package will be referred to as lying ‘under’ the package.

A second, smaller model, extracted from the first and discretized to 0.125 mm was also developed to provide a finer resolution in the computational model and therefore allow us to describe accurately the geometry of the coil. The resulting model had dimensions of $160 \times 250 \times 220$ cells and the discretized coil in this model is shown in figure 4.2.

The base of the package and the secondary coil are covered by a bio-compatible Silicone. Silicone, with a low thermal conductivity and thermal diffusivity, also acts as a heat insulator.
4.2.3 Thermal Properties

The thermal properties of the tissues in our study have been obtained as described in [50]. The thermoregulatory mechanisms of the tissues are represented by the metabolic heat rate and the blood perfusion constant. Both of them are dependent upon the temperature [51], although it was found that their variations are negligible for the thermal elevation under consideration. Temperature dependence of the metabolic rate is given by:

\[ A(x, y, z, t) = A_0(x, y, z, T)1.1^{(T(x,y,z)-T_0(x,y,z))} \]  

(4.2.1)

where \( T_0 \) is the basal temperature and \( A_0 \) is the basal metabolic rate at the co-ordinates \((x, y, z)\). The temperature dependence of the blood perfusion is given by:

\[ B(x, y, z, t) = B_0 i f T(x, y, z) \leq 39^\circ C \]  

(4.2.2)

\[ B(x, y, z, t) = B_0[1 + S_B(T(x, y, z) - 39)] i f 39^\circ C < T(x, y, z) < 44^\circ C \]  

(4.2.3)

\[ B(x, y, z, t) = B_0[1 + 5S_B] i f T(x, y, z) > 44^\circ C \]  

(4.2.4)

\[ B(x, y, z, T) = B_0 \]

\[ i f \ T(x, y, z) \leq 39^\circ C \]  

(4.2.5)

\[ B(x, y, z, T) = B_0[1 + S_B(T(x, y, z) - 39)] \]

\[ i f \ 39^\circ C < T(x, y, z) < 44^\circ C \]  

(4.2.6)

\[ B(x, y, z, T) = B_0(1 + 5S_B) \]

\[ i f \ T(x, y, z) > 44^\circ C \]  

(4.2.7)

The blood perfusion constant accounts for the effect of blood on tissues. In the case of the retina, this effect comes by way of the adjacent choroid a highly vascu-
Table 4.1: Thermal properties of implanted materials

<table>
<thead>
<tr>
<th>Material</th>
<th>Density, $\rho$ Kg/m$^3$</th>
<th>Specific Heat, $C$ J/(Kg $^\circ$C)</th>
<th>Thermal Cond., $\kappa$ J/(m s $^\circ$C)</th>
<th>Thermal diffusivity $\eta$ m$^2$/s</th>
</tr>
</thead>
<tbody>
<tr>
<td>chip</td>
<td>2300</td>
<td>700</td>
<td>60</td>
<td>$3.7 \times 10^{-5}$</td>
</tr>
<tr>
<td>Coil</td>
<td>21400</td>
<td>134</td>
<td>71.6</td>
<td>$2.5 \times 10^{-5}$</td>
</tr>
<tr>
<td>Silicone</td>
<td>2700</td>
<td>1255</td>
<td>0.18</td>
<td>$0.5 \times 10^{-7}$</td>
</tr>
<tr>
<td>Fat tissue</td>
<td>920</td>
<td>2500</td>
<td>0.25</td>
<td>$1.1 \times 10^{-7}$</td>
</tr>
<tr>
<td>Retina</td>
<td>1039</td>
<td>3680</td>
<td>0.565</td>
<td>$1.5 \times 10^{-7}$</td>
</tr>
<tr>
<td>Humor</td>
<td>1009</td>
<td>3997</td>
<td>0.594</td>
<td>$1.5 \times 10^{-7}$</td>
</tr>
<tr>
<td>Cornea</td>
<td>1076</td>
<td>4178</td>
<td>0.58</td>
<td>$1.3 \times 10^{-7}$</td>
</tr>
<tr>
<td>Bone Marrow</td>
<td>1850</td>
<td>2700</td>
<td>0.22</td>
<td>$4.4 \times 10^{-8}$</td>
</tr>
<tr>
<td>Bone Cortical</td>
<td>1850</td>
<td>1300</td>
<td>0.4</td>
<td>$1.7 \times 10^{-7}$</td>
</tr>
<tr>
<td>Bone Cancellous</td>
<td>1850</td>
<td>1300</td>
<td>0.4</td>
<td>$1.7 \times 10^{-7}$</td>
</tr>
</tbody>
</table>

Polarized tissue assigned the properties of blood, which not only provides nourishment to the retina but also acts a heat sink for the outer retinal layers [55, 56, 57]. Therefore, in the computational models, for the retinal tissue, the blood temperature, $T_B$, is replaced by the temperature of the closest choroidal tissue. Based on the findings of [50], the choroidal temperature is allowed to vary with time, as opposed to being fixed at 37$^\circ$C even though it is modeled as blood.

Table 4.1 lists the thermal properties of some of the key materials in the models. The last column of the table lists the thermal diffusivity ($\frac{\kappa}{\rho C}$), which quantifies the ability of the material to absorb heat. The smaller it is, the more heat will be required to raise its temperature, and the longer time it will take to reach steady-state values.

4.3 Computational Methods

The computational methods used have been described in detail in chapter 2. The ADI method was used primarily for verification purposes for the hybrid explicit-ADI approach.

In this work, a dual-resolution scheme was used (figure 4.3) with volume of
dimensions $64 \times 82 \times 118$ around the chip package at uniform 0.25 mm resolution, while the rest of the volume was re-sampled to uniform 0.5 mm resolution. Comparative results with the ADI method is presented in the computational results section.

Figure 4.3: The grid-scheme of the explicit-ADI method (shown in 2-D). The ADI method was used in the higher resolution region at 0.25 mm (model size $64 \times 82 \times 118$), and the explicit method was used for the lower resolution region at 0.5 mm.

4.4 Experimental Verification Methodology

All experimental measurements performed on canines were with the approval of the Institutional Animal Care and Use Committee (IACUC). Experiments were performed with canine eyes because of the general similarity with the human eye. In the experiments, two 680 ohms resistors were used to simulate the actual resistive load of the implant and connected to the secondary coil and a voltage measurement setup through two holes in the substrate. The package also contained three capacitors, which formed the resonant tank circuit with the secondary coil for the wireless power transmission.

To measure the local temperatures, four to six thermistors were placed in the eye region and connected by wires to measurement units outside the eye. Table 4.2 gives the location of the thermistors. While four thermistors were always placed
Table 4.2: Thermistor Locations: Experimental

<table>
<thead>
<tr>
<th>Sensor</th>
<th>Location in eye</th>
</tr>
</thead>
<tbody>
<tr>
<td>T1</td>
<td>Top of the package</td>
</tr>
<tr>
<td>T2</td>
<td>Side of the package</td>
</tr>
<tr>
<td>T3</td>
<td>Internal side of coil under the package</td>
</tr>
<tr>
<td>T4</td>
<td>Coil surface close to center of coil</td>
</tr>
<tr>
<td>T5</td>
<td>Reference thermistor approx. 2 cms. away from the package</td>
</tr>
<tr>
<td>T6</td>
<td>Reference thermistor in a remote location (other eye)</td>
</tr>
</tbody>
</table>

Table 4.3: Temperature increase in canines: Experimental

<table>
<thead>
<tr>
<th>Open eye</th>
<th>DOG2060</th>
<th>DOG2060L</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sample points</td>
<td>△T</td>
<td>adjusted △T</td>
</tr>
<tr>
<td>△T1</td>
<td>2.13</td>
<td>1.5</td>
</tr>
<tr>
<td>△T2</td>
<td>2.17</td>
<td>1.54</td>
</tr>
<tr>
<td>△T3</td>
<td>0.88</td>
<td>0.25</td>
</tr>
<tr>
<td>△T4</td>
<td>1.49</td>
<td>0.86</td>
</tr>
</tbody>
</table>

close to the eye, thermistors 5 and 6 were placed far away from the implanted region to obtain baseline (reference) temperatures. Figure 4.5 shows the approximate locations of the two temperature sensors that could be accurately monitored in simulations.

Experimental results were obtained for two canines, referred to as DOG2060 and DOG2060L. Figure 4.4 illustrates the power dissipated by the load with time. Since the measured equilibrium △T values were not available at 97 mW power dissipation, the absolute and the baseline △T were obtained by interpolating between equilibrium values at 75 mW and 100 mW and 75 mW and 125 mW. Table 4.3 provides the thermal increase at the four sample points. The relative △T’s (adjusted) were obtained by subtracting the baseline increase from the absolute △T’s (unadjusted).

Experiments with open and closed eyelids in DOG2060L showed a marginally
higher temperature increase for the closed eyelid as shown in table 4.4. This is expected, since an open eyelid allows a cornea-air interface, which has a higher convective coefficient (20 W/m²·°C) than for skin-air interface (10.5 W/m²·°C), thus facilitating a higher heat transfer rate to air. Further, blood flow in the eyelid (for a closed eye) contributes to maintain a higher temperature in the cornea. A larger temperature difference will be expected in regions close to the corneal interface with air.
### Table 4.4: Temperature reading with open and closed eye: Experimental

<table>
<thead>
<tr>
<th>DOG2060L</th>
<th>Open Eye</th>
<th>Closed Eye</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>△T1</td>
<td>△T2</td>
</tr>
<tr>
<td>Unadjusted</td>
<td>2.29</td>
<td>2.41</td>
</tr>
<tr>
<td>Unadjusted</td>
<td>2.35</td>
<td>2.49</td>
</tr>
</tbody>
</table>

Figure 4.6: The dimensions and shape of the thermistor.

#### 4.5 Computational Results

For the computations, the initial temperature values were obtained by simulating the bioheat equations for the entire head with only natural heat sources, i.e., only the basal metabolic rate and the blood perfusion (assuming $T_B=37$ degrees C). It should be noted that the initial temperature at all points in the head is not 37 degrees C. Rather, it is dictated by the natural heat sources, tissue environment, and most importantly, proximity to the ambient. Therefore, tissues such as the cornea have an initial temperature around 34 degrees C. Further, Since the truncation and extraction of smaller models and subsequent smoothening by interpolation changes the original model slightly, another simulation was required to get more accurate initial temperatures. Ignoring this step will likely lead to errors which will be significant when the temperature increment due to artificial sources is not very high.

In the experimental setup, the thermistor element was encapsulated in a Polyimide tube of diameter 0.43 mm and length 4.45 mm, as shown in figure 4.6.
To obtain simulated thermal increments for a comparative study, the temperature values were averaged over the volume of the thermistor tip. However, the thermistor and the lead wires were not part of the models.

4.5.1 Numerical Validation of the explicit-ADI method

Simulations were done using the ADI method to compare with the explicit-ADI method. The time step was identical (0.015 s), and for a duration of 12 minutes of physical time, the dual resolution explicit-ADI method was about 11 times faster. Table 4.5 gives the comparative results for the two methods. The differences in the values are likely because of the resolution difference at the boundary of the two computation domains and the inherent higher numerical errors of the explicit method. Figure 4.7 illustrates the comparison between the two methods.

Figure 4.7: Comparison of temperature increase along a cross-section through the eye and implant after 12 minutes of physical time for a package dissipating 97 mW. Top: Using ADI method, Bottom: Explicit-ADI
Table 4.5: Maximum and Average single-voxel temperature increment due to operation of chip. (in degrees Centigrade)

<table>
<thead>
<tr>
<th>Material</th>
<th>ADI</th>
<th>explicit+ADI</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sclera</td>
<td>2.30</td>
<td>0.495</td>
</tr>
<tr>
<td>Choroid</td>
<td>1.74</td>
<td>0.486</td>
</tr>
<tr>
<td>Retina</td>
<td>1.55</td>
<td>0.486</td>
</tr>
<tr>
<td>Vitreous Humor</td>
<td>1.43</td>
<td>0.342</td>
</tr>
<tr>
<td>chip substrate</td>
<td>6.52</td>
<td>5.43</td>
</tr>
<tr>
<td>Package</td>
<td>5.33</td>
<td>4.53</td>
</tr>
</tbody>
</table>

4.5.2 Thermal Effects of Chip Package

As mentioned in Section II, the implant is positioned on the sclera. It is prudent to assume it does not form a continuous material with the sclera, and there is always a small, but finite gap. Since this finite gap is not a constant, two cases estimating the bounds have been considered in this research.

1. When there is negligible distance between the silicone base of the chip package and the scleral tissue. This will give the worst case estimate of the thermal elevation in the eye (referred to as case-1).

2. When there is a finite distance of approximately 0.5 mm to 1 mm between the package base and the sclera (referred to as case-2). To simulate an imperfect sclera-implant contact and more importantly, a difference between the radius of curvature of the eye and the implant, this gap is modeled as air.

For DOG2060, the power was first stepped from 0 to 97 mW and maintained for 4 minutes before further variations(figure 4.4). A comparison of the experimentally recorded values with simulated values over the first 4 minutes is given in table 4.6. The averaged temperature readings over the thermistor volumes from simulations for DOG2060L are compared with measurements in figure 4.8.
Table 4.6: Thermal elevation after 4 minutes: Comparison between simulation and experimental (DOG2060)

<table>
<thead>
<tr>
<th>Location</th>
<th>Case-1</th>
<th>Case-2</th>
<th>DOG2060</th>
</tr>
</thead>
<tbody>
<tr>
<td>△T1</td>
<td>1.64</td>
<td>1.97</td>
<td>1.50</td>
</tr>
<tr>
<td>△T2 (one side of package)</td>
<td>1.40</td>
<td>1.67</td>
<td>1.53</td>
</tr>
<tr>
<td>△T2 (other side of package)</td>
<td>1.45</td>
<td>1.75</td>
<td>-</td>
</tr>
</tbody>
</table>

Though the temperature variations are similar for the experimental and simulated values, there are two important differences. One obvious difference is in the absolute temperature values. The second difference is the fact that the experimental readings are at steady state near the time values of 14 minutes, 43 minutes, 62 minutes, and 75 minutes for applied power of about 50 mW, 51 mW, 75 mW and 100 mW, respectively; while the simulation values still show a growing trend, though small. A possible reason for this is that tissues in general have very low diffusivity (Table 4.1) leading to longer times to steady state as compared to thermistors which typically have a faster charging time.

To characterize the 60-electrode epi-retinal prosthesis under its worst-case operating conditions at 97 mW, simulations were done for longer periods to obtain near-steady state values. A comparison for 50 minutes of physical-time simulation for the two cases mentioned above is done with steady-state measured values. As mentioned in section 4.4, since the measured equilibrium △T values were not available at 97 mW power dissipation, the absolute and the baseline △T were obtained by interpolating between equilibrium values at 75 mW and 100 mW and 75 mW and 125 mW.

The experiment used a hand-wound coil, causing size inaccuracies, and its placement also introduced positional discrepancies with respect to the chip package as compared to the model of the actual implant used in the simulations.

Consequently, △T3, which depends on the relative position of the coil and the package, could not be monitored. The uncertainty in the position of △T4 and its
Figure 4.8: Temperature reading for DOG2060L over time: experimental versus simulated. Since the power applied for the first 22 minutes is negligible as illustrated in figure 4.4 the results here are for a period of 76 minutes after the first 22 minutes. 'd' is the approximate distance between the package and the thermistor tip. The simulated values are for Case-1.
Table 4.7: Thermal elevation after 50 minutes for simulation and about 12 minutes from experimental measurements. I: When the thermistor tip is approximately 0.35 mm from package. II: When the thermistor tip is approximately 0.71 mm from the package.

<table>
<thead>
<tr>
<th>Location</th>
<th>Measured</th>
<th>Simulated</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Unadjusted</td>
<td>Case-1</td>
</tr>
<tr>
<td></td>
<td>DOG2060</td>
<td>DOG2060L</td>
</tr>
<tr>
<td>△T1</td>
<td>2.13</td>
<td>2.35</td>
</tr>
<tr>
<td>△T2</td>
<td>2.17</td>
<td>2.49</td>
</tr>
<tr>
<td>△T2</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>△T4</td>
<td>1.45</td>
<td>0.67</td>
</tr>
</tbody>
</table>

variability stem from the relatively large size of the coil, and the significant temperature variation along its dimensions, respectively. Also, since △T6 is located in the other eye (which is not part of the models), only the absolute (unadjusted) temperature increments have been used for comparative study.

From tables V and VI, it can be seen that the simulated results are in good agreement with the unadjusted values of the experimental results. Two △T2’s have been reported for two diametrically opposite locations outside the package. For obtaining △T4, the thermal effects of the chip package and the coil (discussed later) were added.

The difference in temperature at the two △T2’s is expected since the power dissipation in the chip is asymmetric. As mentioned above, results in table 4.6 and 4.7 are averaged over the volume of the thermistor. However, there is a large gradient of decreasing temperature moving away from the implant, as shown in figure 4.9 for case-1.

A difference between experimental and simulation values is the fact the temperature increase is higher at the side of the package than the top, while it is vice-versa for the simulation. The likely cause of this is that, in the experiments,
the resistors dissipating all the power were closer to the sides than the top leading to higher $\Delta T_2$. In the simulations, the actual implant configuration is used, where there is a hybrid substrate (dissipating 16 mW) on the chip, making it closer to the top of the package. Further, differences in tissue properties (between human and canine head), implant positioning, measurement accuracy and accuracy of thermistor placement can play a significant role. For example, even a slight displacement by 0.25 mm of the thermistors away from the package could lead to a drop in reading by about 0.1 degrees.

Table 4.8 summarizes the overall simulation results for the implant with the maximum and average temperatures of key tissues and materials. It should be noted, however, that the mentioned maximum temperatures are observed in individual computational voxels and, therefore, they could be subjected to numerical approximations due to staircasing and linear approximation from voxel to voxel.

Barring the base of the package, which has a silicone coating, the package is in direct contact with tissue (mostly fat). For case-1, the packaging walls experience an average increase of 3.98 degrees C. Consequently, the temperatures of fat tissue at the interface are between 3.3 and 4.3 degrees C. The corresponding temperature increases for case-2 are 4.66 at the walls, and between 3.3 and 4.9
Figure 4.10: (a) shows the cross-sectional plane of the eye and implant over which the thermal elevation is illustrated in (b) for case-1(top) and case-2(bottom). The colorbar shows the temperature in degrees C.
Figure 4.11: Temperature history at some locations for case-1 (Top), and case-2 (Bottom).

for fat. However, these temperatures drop off rapidly as shown in figure 4.9. This is explains the large difference between the maximum and average temperature increases in Table 4.8, since, tissues removed from the package site experience significantly lesser heating effect.

The low thermal conductivity of the vitreous humor helps reducing the thermal effects on the eye tissues; the temperature increase at a point on the retina on the side opposite to the implant was 0.1 and 0.05 degrees C in case-1 and case-2 respectively. The positioning of the implant and the heat spread in the eye and the package is shown in the color figures of the two cases in figure 4.10. The thermal elevation in the chip package, and the regions around the eye have been suppressed.
Table 4.8: Maximum single-voxel and Average temperature increment in implant materials and key tissues due to operation of chip: Simulation (all temperatures are in degrees Centigrade)

<table>
<thead>
<tr>
<th>Material</th>
<th>Case-1</th>
<th>Case-2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sclera</td>
<td>3.65</td>
<td>0.55</td>
</tr>
<tr>
<td>Choroid</td>
<td>3.2</td>
<td>0.59</td>
</tr>
<tr>
<td>Retina</td>
<td>2.95</td>
<td>0.54</td>
</tr>
<tr>
<td>Vitreous Humor</td>
<td>2.68</td>
<td>0.56</td>
</tr>
<tr>
<td>Cornea</td>
<td>0.81</td>
<td>0.29</td>
</tr>
<tr>
<td>chip substrate</td>
<td>5.9</td>
<td>4.84</td>
</tr>
<tr>
<td>Package</td>
<td>4.79</td>
<td>3.98</td>
</tr>
</tbody>
</table>

Figure 4.12: Temperature Profile along axis (shown in figure 4.10) of chip package.
Figure 4.13: Temperature Difference for sample points in Muscle (above package) and Retina (below package).

The temperature history of some points in the tissues, for 50 minutes of simulated physical time are shown in figure 4.11 for both the cases. In both the cases, the chip gets quickly heated. At the retina underneath, the effect is lesser and more gradual due to the silicone insulation, distance from the source, and the cooling effect of the choroidal blood.

Figure 4.12 shows the temperature profile through the axis of the chip package, illustrating the thermal effects due to the variation in the position of the implant. It can be inferred from the figure and table 4.8 that, for case-1, the temperature increase is higher below the base of the package and lower above the base. This can be explained by the fact that the gap in case-2 provides a high thermal resistance path, relative to case-1, below the base resulting in more heat flow towards the upper regions of the package. This also explains why the $\Delta T_1$ and $\Delta T_2$ values were higher for case-2 (table 4.6 and 4.7).

Figure 4.13 shows the difference of temperature readings at two identical sample points for the two cases for 12 minutes of simulation time. The shaded region indicates the expected thermal increment due to an implant positioned intermediate to those considered in this work. For the retinal sample point the increment range is found to be between 1.54 degrees C and 2.25 degrees C, a variation of
Table 4.9: Thermal effects due to operation of the secondary coil

<table>
<thead>
<tr>
<th>Tissue/Material</th>
<th>Maximum Temperature Increment (degrees C)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sclera</td>
<td>0.22</td>
</tr>
<tr>
<td>Choroid</td>
<td>0.19</td>
</tr>
<tr>
<td>Retina</td>
<td>0.17</td>
</tr>
<tr>
<td>Vitreous Humor</td>
<td>0.18</td>
</tr>
<tr>
<td>Fat</td>
<td>0.23</td>
</tr>
<tr>
<td>Coil</td>
<td>0.39</td>
</tr>
<tr>
<td>Insulation(Silicone)</td>
<td>0.38</td>
</tr>
</tbody>
</table>

0.71 degrees C. This difference, however, also includes the effect of the distance from the implant, which is slightly different for the two cases.

Once again, high single-voxel temperatures obtained at tissue adjacent to the chip package could also be artifacts of the numerical method, since it treats the model as continuous, with a continuous solution. In reality, material discontinuity at the interface could lead to abrupt variations in temperatures. Therefore, the average temperature over the size of the thermistor as reported in table 4.6 and 4.7 may be the most reliable way to report temperature increases since numerical artifacts tend to eliminated by the averaging procedure.

### 4.5.3 Thermal Effects of the Secondary Coil

As mentioned earlier, for the simulation of the heat caused by the secondary coil, a higher resolution model was used. The only artificial source of heat was the current carrying coil. The power by each voxel of the discretized coil model was calculated using the formula, $\frac{I^2 \rho L}{A}$, where $I$ is the average current (75 mA), and $\rho$ is the resistivity ($24 \times 10^{-9}$). Summing over all the voxels defining the coil, the resistance and the total power dissipated were 1.83 Ω and 0.01 watts, respectively.

The heating effect of the coil in the tissues is shown in table 4.9. The maximum
temperature increase due to the current in the coil is 0.39 degrees C, obtained in the metal coil. For the tissues, the increase is relatively low, with a maximum increase in fat tissue at a point in contact with the insulation. Figure 4.14 is a transverse cross-section of the model illustrating the thermal elevation around the coil.

All the hot spots due to the coil are far away from the hot spots due to the operation of the chip. Consequently, they are not likely to affect the monitored temperatures $\Delta T_1$ and $\Delta T_2$ or the maximum tissue temperatures in table 4.8. However, the thermal effect of the package will alter the values in table 4.9. Adding the effects of the package and the coil, the maximum temperature increase at the point in retina referred to in table 4.9, is approximately 0.8 degrees C. At the top
part of the coil closest to the implant, the temperature increment is about 2.3 degrees C due to the implant and 0.3 degrees C due to the coil.

4.5.4 Thermal effects of the electrode array

The metal electrodes are responsible for injecting the required charge into the underlying inner retinal layers. For the thermal simulations, a smaller region, near the back of the eye, was extracted and its resolution increases to 0.125 mm, resulting in a model of size $76 \times 116 \times 92$. Only the electrodes, without the interconnects from the driver chip, were considered in the model. Each discrete stimulus comprises a biphasic pulse, with each pulse width being 1 ms or 3 ms. Figure 4.15 illustrates the power dissipated by each electrode. For the simulations,

![Power dissipated per electrode volume](image)

Figure 4.15: Power dissipated per electrode volume (a) for a pulse width of 1 ms (b) for a pulse width of 3 ms. Note that the biphasic pulse repetition rate is 50 Hz

corresponding to a peak current of about $620 \mu A$, the power dissipated by the each metal electrode is about $305 \times 10^{-12}$. Figure 4.16 gives the temperature increases at a cross-section of the eye consisting of the electrodes. The maximum temperature increase was about $0.005^\circ$ C, in the vitreous in contact with the electrodes, while the maximum temperature increase in the retina was about $0.003^\circ$ C.
4.6 Conclusion

We have presented a computational study of the thermal impact of Second Sight’s second generation 60-electrode retinal prosthesis implant and compared these results with experimental findings in canines. We found that the agreement between these two sets of results, conducted as a masked test, shows that numerical simulations can predict with a high degree of accuracy the thermal effects of such implants. This indicates that numerical simulations can be successfully used in determining optimal positions of the components of the retinal prosthesis system in order to minimize heat diffusion in the human tissue and prevent delicate tissues such as the retina from being harmed by the temperature increase of the implanted device. Specifically, thermal effect information of very localized hot spots, and regions where physical measurement might not be possible, can be easily monitored using these simulations.

For the chip package, based on realistic assumptions, two cases of implant position were considered to obtain the temperature elevation ‘range’ under maximum heating conditions (i.e. the power dissipation value is calculated when the chip is driving all 60 electrodes at maximum current and the current in the secondary coil is assumed constant at its maximum value). The results were in good
agreement with experimentally obtained values. Among the eye tissues, the maximum single-voxel temperature increase was obtained in the outermost sclera in the range 2.47-3.65 degrees C, while the sensitive retina experienced a single-voxel temperature increase in the range 1.72-2.92 degrees C.

The maximum temperature increases in the tissues due to the secondary coil with an average constant current of 75 mA, is about 0.23 degrees C in the fat tissue bordering the coil insulation; and 0.22 degrees C in the sclera. However, the thermal effect of the chip package dominates almost the entire eye, except for the region close to the lower tip of the coil, where $\Delta T(Due\; to\; Coil) \geq \Delta T(Due\; to\; Implant)$.

The difference between simulation and experimental results can be attributed to a number of factors, such as the inherent errors of the computational methods, the discretization of the problem space, measurement inaccuracies, discrepancies (between experiment and simulation) in thermistor placement, differences in human head model and canine tissues, and the fact that the thermistors and the lead wires were not part of the simulated model.

It is worth mentioning that the experimental measurements also included baseline measurements in the other eye ($\Delta T_6$) reflecting temperature changes due to the ambient temperature or other uncharacterized physiological action, leading to 'net' temperature increases ($\Delta T_1$ and $\Delta T_2$) of less than 2 degrees C.

In this research, the focus of the discussion has been mostly on the temperature increase rather than absolute temperatures. For the implant considered in this work, at an ambient temperature of 24 degrees C, the maximum single-voxel temperatures in retina and sclera were found when the chip-package is in perfect contact with the sclera (referred to as Case-1): in this case, final temperatures were found to be less than 38.3 degrees C in the retina and 38.9 degrees C in the sclera. For a more realistic case of a 0.5 to 1 mm gap between the base of the chip package and the sclera (termed as Case-2), temperatures in the retina and sclera drop to 37.2 and 38 degrees C, respectively.
Chapter 5

Wireless Power Telemetry

5.1 Introduction

In the previous two chapters, detailed electromagnetic and thermal modeling for a 60-channel epi-retinal prosthesis was performed. The implant comprised a hand-wound coil placed on the eyeball, with a hermetically sealed package placed on scleral strap wrapped around the eyeball. Inductive coupling is used for power and data transfer to the internal components. However, as higher resolution prostheses are pursued, data rate requirements are going to be a significant bottleneck for the inductive coupling system which need to operate below 20 MHz for reliable operation. In [39], it was proposed to have mutually exclusive power and data transfer to the internal components. Specifically, it was suggested to have a low-frequency (1-10 MHz) inductive link for power transfer and a microwave frequency (1-3 GHz) link for data transfer, with the internal antenna placed in the anterior region of the eye. Further, the intraocular patch antenna designed for data telemetry at microwave frequencies, was of dimensions 6 mm × 6 mm [39]. A novel configuration was proposed in [?], which could allow integration of the internal antenna, the secondary coil and the stimulating chip and related electronics on the same substrate.

Here, we are concerned with the fabrication of a novel inductor geometry for the secondary coil of the inductive link using silicon micro-machining and
photolithography. It was obtained using the partial inductance computational method that the proposed 3-D trench inductor design, shown in figure 5.1, could provide more coupling for the same length of inductor traces and allow more lines for the same surface area. Further, fabrication of these 3-D inductors would require two mask steps with only a single mask step for patterning the inductor lines. The region within the metal lines could be used for a patch antenna and the coil could be used to provide power to electronics on the other side of the substrate using wafer-through vias.

Figure 5.1: The proposed trench design for a 3-D inductor. The planar trench region in the center could be used for a patch antenna and the back side for other electronics [?].

5.2 Mask Design

The masks were designed in AutoCAD. Overall, this was a two mask step process, with the first step for demarcating area for silicon bulk etching, and the second for patterning lines. Figure 5.2 shows the CAD images of the masks used. The mask dimensions were 7 inches $\times$ 7 inches $\times$ 0.12 inches on sodalime chrome material. Along the two diameters (indicated in figure 5.2a) were test structures and alignment features. The line-widths of the inductors ranged from 5 $\mu$m to 25 $\mu$m.
Figure 5.2: The masks were for 6 inch wafers. (a) Locations of the windows for bulk silicon etching. The shaded region along two perpendicular diameters indicates the location of the test and alignment structures. (b) Mask for patterning of metal lines.
5.3 Fabrication Issues of proposed geometry

Before patterning of metal lines, an appropriate trench needs to be formed. Both, bulk-silicon isotropic and anisotropic etching were investigated.

The profile of an isotropically etched wafer, done by immersing in a mixture of HF+HNO$_3$ +CH$_3$COOH(HNA solution) at room temperature is circular. In anisotropic etching, sharply defined inclined walls can be obtained by etching $<100>$ Si-wafers in a 40 % KOH solution at elevated temperatures. The etch-stops are provided by different planes of the crystal lattice of silicon. Differently oriented planes ($R_{etch(<110>)} > R_{etch(<100>)} >> R_{etch(<111>)}$) offer different resistances to the etchant, with the $<111>$ plane being the etch-stop plane.

The main drawback of these walls is the steepness of the incline ($55.7^\circ$ to the horizontal), which can lead to cross-exposure on the inclines as well as at the base of the trench near the inclines. A proposed method to reduce the steepness is to make the profile more 'gradual' by performing an isotropic etch after an anisotropic etch. The fabrication results of the three different types of etches carried out is described later in this chapter.

Another issue that needs to be tackled is the loss of resolution as one goes deeper in the trench. This effect is due to Fresnel diffraction, and becomes noticeable when the distance between the mask and the exposed region becomes substantially large. The minimum line width $w_{\text{min}}$ is theoretically described by the following equation:

$$w_{\text{min}} = \frac{3}{2} \sqrt{\frac{\lambda(a + \frac{d}{2})}{}}$$

(5.3.1)

where $\lambda$ is the wavelength of the incident light, $a$ is the distance between the mask and the substrate, and $d$ is the resist thickness. Figure 5.3 shows the minimum line width at different etch depths for $\lambda = 400$ nm, $a = 400\mu$m and $d = 15\mu$m.

Before the exposure step, the main difficulty is in the patterning of lines on the walls of the inclines, for which conformal photoresist coverage is needed. Spin coated photoresists are unlikely to provide conformal coating, especially if the etch is deep. Thus, new unconventional methods of resist coatings which are immune to the structure of the surface, were considered.
Figure 5.3: Maximum possible resolution with increasing etch depth.

5.4 Electrophoretic Resists

Electrophoretic resists (such as Eagle 2100 and PEPR 2400) allow patterning of structured surfaces [59], thus allowing another degree of freedom. The major advantage of using electrophoretic photoresists is that they can be used to deposit fairly uniform resist coatings on structured substrates. The coating takes place in an electrodeposition setup (similar to electroplating) where 2 electrodes are immersed in the electrolyte (dilute resist solution). One of the electrodes is a stainless steel plate, while the other is the target surface to be coated. To facilitate conduction, the target surface needs to be metallized (Ti, Cu, Ag, Pt, etc). The coating process can be briefly described as follows: The resist emulsion consists of charged micelles (positive for negative resist, and negative for positive resist), which on application of electric potential migrate towards the opposite charged electrode (target surface). On reaching the target surface, the micelles are neutralized by hydroxide ions, destabilize and settle on the surface. Since the
deposited material is insulating, the process is self-limiting and conduction ceases once the entire conductive surface is covered. Thus, the process is independent of the shape of the surface and ideal for coating structured, 3-D terrain. The main parameters controlling the thickness are

1. Applied Voltage,
2. Temperature of resist bath,
3. Coating time
4. Conductivity of solution,
5. Percentage solids in the bath,
6. Concentration of a plasticizer chemical.

The major disadvantages of using an electrodepositable resist are related to costs, resources, and complexity of use. They generally have low shelf lives, require an elaborate setup and regular tests for coating and maintenance purposes. The resolution is lower than for spin-coated resists such as the Shipley S1813. Further, they are still not established in the IC industry, and published literature on applications and potential problems is relatively scarce.

5.5 Cleanroom Fabrication

This section will discuss the fabrication process followed, the problems encountered, the steps taken to circumvent them and some results. There are two macro steps; Trench formation by Silicon bulk-etching and PEPR 2400 processing(including metal etching and PEPR stripping).

5.5.1 Trench Formation

Since the etchants (for isotropic and anisotropic) will easily etch through any photoresist, $Si_3N_4$ is used as the mask for the etch step. Patterning of etch
windows through $Si_3N_4$ is used by masking with $SiO_2$. Isotropic etch with HNA was performed at room temperature for 25-40 minutes to get different etch depths. Anisotropic etch was carried out $80^\circ C$ for 4-8 hours with an etch rate of 0.9µm/min.

5.5.2 PEPR 2400 processing

Figure 5.4 illustrates a simple PEPR deposition setup that was used in this research. For large-scale PEPR2400 deposition, the setup can become quite complex, with an ultra-filtration setup and various maintenance sub-experiments. The initial solution consists of 50% by volume of the concentrate supplied by the manufacturer and 50% by volume of De-Ionized (DI) water. The thickness variation of deposited PEPR has a dependence on the temperature of the bath and especially the concentration of a plasticizer chemical, called the PEPR TC thickness controller. Since the primary component of the TC is 2-Octanone (97%), a 2-Octanone(OCT) solution instead of the TC was used to save on costs. The conductivity of the solution is a function of the electrodeposition process and increases with time. To limit the conductivity within prescribed limits (350 $\mu$S - 450 $\mu$S), conductivity tests were regularly conducted and DI water (about 10 $\mu$S) was added to reduce it. Also, the percentage solids (of resist) decreases with each coating. Regular tests were conducted and the percentage solids maintained within prescribed limits (9 - 11 %)

Another key parameter which decides the thickness is the voltage. For a given coating time, higher the voltage, higher the PEPR thickness.

Since we are patterning inclined walls, light rays (during exposure) will undergo reflections at various angles ($35^\circ$ for the anisotropically etched wall) on the trench walls, resulting in cross-exposure in adjacent areas, leading to loss of features due to overexposure. Further, for patterns deeper in the trenches, there will be significant loss of resolution due to diffraction. To target these two issues, it was decided to under-expose the PEPR, to allow a thin layer to remain even after developing. The wafers were then descummed in an asher in an oxygen
Figure 5.4: Schematic of the experimental setup used for electro-deposition of PEPR2400.

environment to remove unwanted resist (where it is thinner). The process steps and parameters were:

1. Metallization: Titanium metal of thicknesses 1500 A to 2000 A were deposited by thermal evaporation.

2. Surface cleaning and wetting: The target wafer was dipped in dilute HF for 30 seconds, followed by DI water. The surface was then sparged with Acetone to remove any remaining stains or particles. Finally, the wafer was washed in DI water prior to entering the resist bath.

3. Electrodeposition: The wafer and the cathode are then immersed in the resist solution and vibration is provided for 40 seconds. Then a voltage is
applied for short period (5-15 seconds). The voltage magnitude and the time depends on the resist thickness desired. Typically, 45-65 volts was applied for 8 to 10 seconds.

4. Rinsing and Drying: Post coating, the wafer was washed in DI water to remove any excess solids loosely attached to the surface, and blow dried with a nitrogen gun. It is imperative that the wafer entering the baking oven be dry.

5. Baking: The wafer was then put in a oven maintained at a temperature of 105-110°C for 12-18 minutes (typically 15 minutes). The baking step is crucial because it not only drives away moisture from the photoresist (as with most other resists), it also serves to activate a photoactive compound.

6. Alignment and Exposure: Based on literature provided by Rohm & Hass, energy dose of 450 mJ/cm² is sufficient. Based on this and the power of the MA-6 equipment that was used, a 30 second exposure was required. However, this was reduced for the etched wafers.

7. Developing: a 0.2 N NaOH solution at 30°C was used. The wafer was then rinsed in DI water and blow dried with nitrogen.

8. Descumming: was done for 300 s to remove the under-exposed resist by oxidation.

9. Bake: A 2 minute bake was done at 105°C to increase the adhesion of the resist.

10. Metal etch: The exposed metal was removed by etching in standard solutions from Transene Inc.

11. Stripping: Due to post baking, the remaining resist is difficult to remove by using the standard stripper (0.3 N NaOH at 55°C) or Acetone. It is done by stripping in asher (from MARCH Inc.) for 300s after reaching 50 % of endpoint.
An illustration of the fabrication process is given in figures 5.5–5.8.

### 5.6 Fabrication Results

Figure 5.9 shows the profile of a wafer which was etched only in HNA. As expected, the etch profile is circular. The curvature is such that it is easy to pattern lines near the base, but virtually impossible to pattern near the top (since profile is almost 90° to the horizontal) using conventional exposure methods. Further, deep isotropic etching was found to be difficult to control, and tended to heavily roughen the silicon surface (figure 5.10).

Figure 5.11 shows the profile of an anisotropically etched wafer with a coating of electrodeposited PEPR 2400 photoresist. Note the well defined inclines and the conformity of the resist over the structured surface.

Figure 5.12 shows a sample with a 5-hour KOH etch and a 10 minute HNA etch. One can see that the HNA etch serves to provide some curvature to the incline and makes it more gradual, which could make it easier to pattern the walls. However, this etching scheme was not established and further etching experiments will be required to standardize the process.

As mentioned in the earlier, baking forms a crucial step of PEPR processing. Inadequate baking not only stunts the photo-action of some compounds in the photoresist, but also weakens adhesion to wafer surface [60]. The effect of inadequate baking can be seen figure 5.13, where the PEPR lines have moved from their designated places.

It was obtained that for thicker coatings (55-65 volts for 8 seconds), a baking time of 18 minutes was required, while for relatively thinner coatings (45 volts for 8 seconds), a baking time of 15 minutes was enough. Figures 5.14, figure 5.15a and figure 5.15b show the Titanium metal lines on planar and etched surfaces.

Figure 5.14 is taken from an optical microscope. Figure 5.15a and 5.15b are Scanning Electron Microscope (SEM) images giving a better perspective of the three dimensional surface features. In figure 5.15, note that the metal lines on the incline cover more surface area than the planar lines. To get uniform metal lines,
Figure 5.5: This figure shows the cross-section of the Si wafer during the processes before the formation of the window for the wet Si etch.
Figure 5.6: This figure illustrates the patterning of the wafers and the Si etch process. Note that the isotropic etch using KOH also etches the oxide and nitride layer, although much slower than silicon. However, etch rates should be factored in during the following nitride etch.
Figure 5.7: (a) The figure indicates the problem with standard exposure techniques on deeply etched surfaces. (b) The result of underexposure.
Figure 5.8: The result of underexposure and subsequent de-scumming. Note the thinner PEPR in the trench due to cross-exposure.

the mask will need to have thinner lines over the inclines.

5.7 Conclusion

This chapter discusses the steps taken towards the fabrication of novel 3-D inductor. While there are numerous issues to be dealt with before a fully operational inductor can be fabricated, significant progress has been made, and a number of problems identified and successfully dealt with. Cross-exposure has been reduced by decreasing the exposure time, and subsequently removing the residue resist by de-scumming. It has been demonstrated that metal lines can be printed on an incline, at an angle $54.7^\circ$ to the horizontal. Further, $25\mu m$ lines were
Figure 5.9: An SEM image of isotropically etched silicon. While patterning lines near the base is relatively easy, it is virtually impossible to pattern near the top.

Figure 5.10: An illustration of the surface roughness experienced after deep isotropic etches.
Figure 5.11: An anisotropically etched wafer with a thick coating of PEPR2400 photoresist

Figure 5.12: A typical etch profile after anisotropic and isotropic etching
patterned at etch depths of up to 400 microns. While only three lines were possible on the incline, as shown in figure 5.15, a better mask design should allow more lines to be printed on the incline region. Also, it can be seen in the figure, that there is a slight misalignment between the metal lines and the trench. Special alignment marks will need to be included factoring in the etch depths desired. While anisotropic etching is slow and maintains a more or less steady etch rate of 0.9µm/min, isotropic etching is much faster (etch rate 20µm/min) and difficult to control. Consequently, to minimize the possibility of misalignment, and to have a more precise estimate of etch depths, isotropic etching should be avoided. Also, the titanium metal of thickness 0.15-0.2 µm, and oxide thickness 0.2 µm was used. Once the fabrication process is better established, better performance inductors can be made by increasing metal and oxide thicknesses and by material enhancements such as using metals like Gold or Copper.
Figure 5.14: Titanium metal lines for a planar inductor on unetched silicon. The line-width here is 10 µm.

The choice of substrate is very important to minimize substrate losses when working with RF signals. Ideally, a semi-insulating substrate like GaAs, or Quartz would be preferable for wireless telemetry. However, the need for standard and economically viable process for bulk micromachining and lithography makes Silicon the desired choice of substrate.
Figure 5.15: SEM images of titanium lines (25 μm) on anisotropically etched silicon. The lines are significantly thinned after the metal etch step.
Chapter 6

Conclusion and Future Work

6.1 Summary of Contribution

The work presented in this dissertation has dealt with various aspects of the Artificial Retina Project. A summary of the work follows:

6.1.1 Electromagnetic Effects

The main activities in characterizing the electromagnetic effects were:

1. Physical modeling of prosthesis components: Discretized models of the electronic components were obtained from CAD drawings and merged with human head models by placement at their designated positions. Resolutions up to 0.3 mm were used to ensure proper resolution of the energy source, the spiral shaped primary coil. To characterize the EM effects for a practical scenario, different orientations or the primary coil were considered.

2. FDTD Source Modeling: A simple source modeling technique was used for the spiral coil, wherein ideal D-field sources were used at the location of the coil voxels. This was verified by established analytical methods under quasi-static assumptions.

3. Safety Compliance Testing: The electric field values obtained from the
FDTD simulations were then used to compute the Specific Absorption Rate (SAR) and averaged current densities to compare with international safety standards (IEEE, ICNIRP). The study was performed for frequencies in the range 2 MHz-20 MHz and it was obtained that, for the same current in primary, the induced electric fields scale nearly linearly with the frequency in the eye region. Further, the simulations in this frequency range indicated that the main compliance factor, the peak 1-g SAR, could be obtained fairly accurately by just scaling with the frequency and conductivity. This is particular usefully for system designers needing a quick estimate of the compliance factors for a particular set of parameters (frequency and power). Preliminary measurements were made with an electric field probe which could be used to measure induced electric fields in a human phantom.

6.1.2 Thermal Effects

The main activities were:

1. Developing numerical computation codes: FORTRAN based numerical codes were developed to solve the 3-D Bio-Heat equation for the human head models implanted with medical devices. Traditional finite difference schemes for time-marching solutions, such as the explicit method and the Alternating-Direction Implicit (ADI) method, were implemented. A simple method to obtain an estimate of the maximum stable time step was used, utilizing the equation for the growth factor obtained from the Von Neumann stability analysis of the ADI scheme. Further, a novel FD scheme was developed incorporating the explicit and the ADI method which provided simulation speed-up of up to 11 times for the particular model considered in chapter 4.

2. Physical Modeling: Models of resolutions up to 0.125 mm were developed to resolve the main heat sources of the system; the stimulator chip dissipating 97 mW of power placed inside the package, and the current carrying internal coil.
3. Implants considered: The implant for a 60-channel epi-retinal prosthesis were considered including the stimulator chip package, the internal telemetry coil and the stimulating electrodes. Additionally, thermal modeling of the human head with retinal implant was done to check for safety in the presence of Magnetic Resonance Imaging (MRI) RF fields.

6.1.3 Fabrication of a 3-D inductor

The main activities were:

1. Mask Design in AutoCAD

2. Etching: Wet isotropic and anisotropic of bulk silicon. Individually, both these etching types impose structural limitations on patterning metal lines on the inclines. It was obtained that by an appropriate combination of both these etches, a more gradual incline could be obtained making it easier to pattern lines.

3. 3-D electro-depositable photoresist: A setup for handling the electro-depositable photoresist, PEPR 2400, was maintained. Tasks included obtaining optimum bake times, keeping the photoresist solution within specifications, optimum voltage application times, etc. Further, to minimize cross-exposure, additional processing steps were proposed which utilized under-exposure of the photoresist during exposure step, and subsequent removal of residue via oxidation (ashing).

6.2 Future Work

6.2.1 ADI-FDTD method for problems in bioelectromagnetics

In this work, for electromagnetic computations, the traditional explicit FDTD (exp-FDTD) has been used. The exp-FDTD is a conditionally stable method
where the maximum allowable time step is limited by the highest resolution required as prescribed by the CFL condition. In bio-electromagnetic (bio-EM) problems, often, this highest resolution is much lesser than that required to resolve the highest frequency of interest. The ADI-FDTD method theoretically is an unconditionally stable method and relatively larger time steps can be used. For the retinal prosthesis, using the ADI-FDTD method, especially in conjunction with expanding-grid methods, or even hybrid methods as done for the thermal computations, can provide a two-fold advantage:

1. Better resolution of the energy sources and scatterers such as the primary coil and the secondary coil, respectively.

2. Larger time steps leading to faster simulations.

As an example, a single turn coil was simulated using the finite-gap method in ADI-FDTD. Figure 6.1 shows the simulation results.

A concern with using the ADI-FDTD method with large time steps is the degradation of accuracy. However, in bio-EM problems, the quantities of interest are usually the induced current densities and SAR, both obtained as averages over a large number of voxels. This tends to reduce the overall errors as shown in figure 6.2.

6.2.2 Hybrid explicit-ADI method for Thermal computations

The hybrid method used in this work included the ADI computations on a volume at twice the resolution of the volume of the explicit method. Further, to simplify the problem, the entire implant was incorporated in the volume of the ADI method. The following improvements are possible to this scheme:

1. Investigating other resolution schemes where in lower resolutions are possible in regions where geometry detail is not important. Further, in the case of two energy sources separated by a distance could be part of different ADI volumes.
2. Explicit parallelization of the computation over different volumes. This will lead to faster simulations and also allow proper utilization of the system resources.

It should be noted that these ideas can also be utilized for the EM computations, although that would be far more complicated.

### 6.2.3 Electric Field Measurement using E-field Probe

To experimentally substantiate FDTD computations of Specific Absorption Rate (SAR), a commonly used methodology is the usage of field probes in hu-
man phantoms filled with tissue simulating fluids. These fluids are generally salt+sugar+water solutions with proportions depending on the tissue dielectric properties (permittivity, conductivity). These probes are typically designed for frequencies greater than few hundred MHz, which is far more than those used for inductive telemetry. However, it has been found that the probe sensors do detect electric fields in the kHz and low MHz regime, and appropriate calibration can make them useable for these frequencies.

Preliminary measurements were performed using such an electric field probe and measurement system manufactured by SPEAG inc. Figure 6.3 shows the equipment and the measurement setup used. Also shown is the field variation in a plane perpendicular to the plane of the spiral coil. To obtain the calibration (or scaling ) factor, the pattern in figure 6.3b will need to be mapped to those obtained by reliable methods, which could be using analytical or FDTD methods.
Figure 6.3: (a) Equipment and setup for electric field measurements. (b) Preliminary measurements for a spiral coil along its diameter between 2 mm and 7 mm from the plane of coil.
Bibliography


[28] Active Implantable Devices - Part 1: General requirements for safety, marking and information to be provided by the manufacturer. BS EN 45502-1:1998, European Standard, Feb. 1998


