ABSTRACT

JASTI, SRINIVAS. Modeling of RF Field Effects Due to MRI Fields in Patients with a Retinal Implant. (Under the direction of Dr. Gianluca Lazzi.)

Magnetic Resonance Imaging (MRI) is one of the most widely used clinical diagnostic procedures. Evaluation of safety due to Radio Frequency (RF) energy deposition and tissue heating in patients during MRI, especially in the presence of implantable prosthetic devices is significant for MR safety. The work presented in this thesis aims to characterize the interactions between the pulsed RF fields during MRI and biological tissues of a patient with a Retinal Prosthesis (RP) implant, in terms of Specific Absorption Rate (SAR) and temperature elevation. A logarithmically expanding grid Finite-Difference Time-Domain (FDTD) is used for computational modeling of the MR environment at 64, 128 and 256 MHz. Unlike traditional methods, expanding grid FDTD facilitates in accurate modeling of the region of the implant where a finer grid with cell sizes of the order of micrometers is used. Also, this technique greatly helps to reduce the constraints on computational memory and time. It was found that, while the RF magnetic field, \( B_1 \), homogeneity decreases with frequency; power deposition in the tissues increases slightly. However, thermal elevation resulting from the SAR distribution as well as the induced currents in the RP implant, evaluated using the bio-heat equation, is observed to be minimal at these frequencies. These results provide useful information for RF safety guidelines during MRI at high fields.
Modeling of RF Field effects due to MRI Fields in Patients with a Retinal Implant

by

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Dedication

To

My family
Biography

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Chapter 1

Introduction

Nuclear Magnetic Resonance Imaging (NMR/MRI) technology is now one of the most powerful tools in medical imaging and scientific research. With increasing applications in surgical techniques, Magnetic resonance (MR) procedures have evolved over the past 20 years with continuous technological advancements leading to MR systems with stronger static magnetic fields, faster and stronger gradient magnetic fields, and more powerful radiofrequency (RF) transmission coils [1]. Since its introduction as a diagnostic tool in clinical practices in the early 1980s, most reported cases of injuries pertaining to MRI have been due to misinformation related to the MRI safety issues of metallic implants and biomedical devices [2]. Many researchers to date studied the various physiological effects associated with abnormalities in visual, auditory, endocrine, neurological, cardiovascular, immune, reproductive, and developmental functions when human body is exposed to RF radiation [1-6].

The advent and popular use of bio-implantable devices that mimic and complement human organs adds to the concerns about adverse health effects due to RF radiation. It is
critical to understand the electromagnetic interactions of these devices with the human body. Investigation of RF deposition and heating caused by these interactions between the gradient magnetic fields and human biological tissues during MRI, especially in the presence of implants, has recently received significant attention. To date, over one thousand implants and objects have been tested for MR safety or compatibility [2]. The local RF heating caused by the absorption of energy from the gradient fields, and by the induced currents as a result of Faraday’s law form the basis of thermoregulatory safety guidelines devised by the United States Food and Drug Administration (FDA) [3].

During an MRI, the deposition and distribution of RF energy within the human body is highly non-uniform and depends on the frequency range of the incident electromagnetic radiation. The primary bio-effects associated with the RF radiation are related to the thermal effects of the electromagnetic (EM) field. The EM fields normally utilized in MRI procedures are in the frequency range of 8.5 to 340 MHz, at which high absorption occurs in the whole body [7]. The absorption of RF power in the human tissues is computed using the dosimetric term, Specific Absorption Rate of energy (SAR), which is normally measured in W/kg.

The goal of this thesis is to evaluate the interactions between the MRI RF fields and biological tissues of patient with a Retinal Prosthesis (RP) implant, in terms of SAR and temperature elevation due to the presence of the medical implant. The RP implant is a restorative device designed to mimic the function of basic photoreceptor cells in patients suffering from outer retinal degeneration caused by age related macular degeneration (AMD) or Retinitis Pigmentosa. Further details about retinal prosthesis will be explained in Section
1.6. Traditionally bio-electromagnetic analyses involving large inhomogeneous dispersive media are solved using Finite-Difference-Time-Domain (FDTD) method. However, the computational intensity and accuracy in modeling the electromagnetic system needs improved efficient algorithms. Three-dimensional expanding Grid FDTD method, [8], has been used extensively for the computational work presented in this thesis. In this chapter a brief overview of electromagnetism involved in NMR/MRI and the technical issues of Retinal Prosthesis will be discussed.

1.1 Background of MRI

Nuclear Magnetic Resonance techniques are used extensively in clinical applications. MR imaging and spectroscopy non-invasively detects the signals from various atomic nuclei in the body which can be used to create images of the body tissues with fine resolution and study their metabolism. The first MR spectrum of a human tumor was obtained by Griffiths et al. from a rhabdomyosarcoma on the dorsum of the hand [9]. The process of acquiring 2-D and 3-D images by NMR, known as magnetic resonance imaging (MRI), was first illustrated by Lauterbur who produced a 2D MR image of a phantom [10]. Since the discovery of NMR by Bloch and Purcell independently in 1946 various developments in the fields of imaging and spectroscopy led to the opening up of whole new branches of physics, chemistry, biology and medicine [12-14]. Over the last two decades, Fourier transform imaging techniques have helped in the improvement of MRI and also increased its range of applications [11].
1.2 Basics of Magnetic Resonance

Nuclei of different atoms possess different spin which can be positive or negative. All nuclei of hydrogen atoms of atomic mass 1 (proton) have a spin of $\frac{1}{2}$. A nucleus with a spin produces a magnetic field around it, which is known as its nuclear magnetic moment. The strength of the field depends on the mass, rate of spin and charge of the nucleus. Also, based on its spin state, the magnetic moment can align either parallel or anti-parallel to the field. For instance, the proton has two possible spin states, $+\frac{1}{2}$ and $-\frac{1}{2}$, which determine the nuclei’s two energy states in an external magnetic field. The positive spin state corresponds to a lower energy state where the nuclear magnetic moment is parallel to the external field and in the higher energy state with a spin of $(-\frac{1}{2})$, the magnetic moment is anti-parallel to the external magnetic field ($B_0$). Hydrogen nuclei possess the strongest magnetic moment and are in high abundance in biological material. Hence hydrogen (or proton) imaging is the most widely used MRI procedure.

The magnetic moments or spins are constrained to adopt one of the two orientations with respect to $B_0$, denoted parallel and anti-parallel. According to quantum mechanics, the alignment of the magnetic moment with the magnetic field is tilted at an angle. The angles subtended by these orientations and the direction of $B_0$ are shown in Figure 1.1a. Consequently, the nuclear magnetic moment is not static in the external field but precesses about static field’s axis. This precession is analogous to the motion of a spinning gyroscope in the presence of earth’s gravitational field (Figure 1.1b) and how fast it precesses is determined by the Larmor equation.
Figure 1.1a

**Figure 1.1** (a) Possible orientations of nuclei in the presence of an external magnetic field; (b) Magnetic Moment precessing about external magnetic field, $B_0$

The Larmor equation expresses the relationship between the strength of a magnetic field, $B_0$, and the precessional frequency, $\omega$, of an individual spin.

$$\omega = \gamma B_0$$ \hspace{1cm} (1.1)

where $\omega$ is the frequency of precession (radians/sec), and $\gamma$ is the gyromagnetic ratio (radians/sec/Tesla), $B_0$ is the magnitude of the external static magnetic field. From Equation 1.1, it can be inferred that the stronger the external magnetic field, the faster the
precession. The gyromagnetic ratio, $\gamma$, is a constant which varies with the type of nucleus. For hydrogen (1H), the gyromagnetic ratio equals to $2.675 \times 10^8$ radians/sec/Tesla.

The instantaneous magnetic moments of a group of 1H nuclei can be represented as shown in Figure 1.2. The individual components of spins perpendicular to $B_0$ cancel out, leaving only the components parallel to the external magnetic field. Since most spins align towards the parallel rather than the anti-parallel state, the net magnetization, $M$, is in the direction of the external field. Suppose the direction of $B_0$ is aligned with the z-axis of 3-D

Figure 1.2 Net Magnetization, $M$, is decided by the difference in the population density of the spin states in the presence of an external magnetic field, $B_0$
Euclidean space and the plane perpendicular to $B_0$ contains the x and y-axes. In order to detect the weak magnetization in the z-direction separated from the strong static field, $B_0$, radio frequency (RF) energy must be applied. RF energy at the Larmor frequency causes nuclear spins to switch between parallel and anti-parallel states. This has an oscillatory effect on the component of $M$ parallel to the $z$-axis. For instance, the magnetic field component of the RF energy represented by $B_1$ lies in the x-y plane. The x-y components of $M$ will be made coherent by the applied $B_1$ field resulting in a net x-y component to $M$. Hence rotation of $M$ about $B_1$, known as tipping, effectively causes $M$ to tilt from the $z$ direction into the x-y plane. This phenomenon is illustrated in Figure 1.3.
Figure 1.3 RF radiation effect on the net magnetization, $M$, results in $M_{xy}$. $M$ is rotated from its longitudinal orientation towards the transverse X-Y plane through flip angle.
The angle of deviation of the magnetization vector, \( M \), away from the z-axis is known as the flip angle. The strength and duration of \( B_1 \) determine the amount of energy available to achieve spin transitions between parallel and anti-parallel states. Thus, the flip angle is proportional to the strength and the duration of \( B_1 \). After an RF pulse is applied that causes the net magnetization vector to flip by 90 degrees, \( M \) lies in the x-y plane and begins to precess about the \( B_0 \) axis. If a receiver coil is placed in the x-y plane, the rotating \( M \) will cause a changing net magnetic flux through the coil and according to Faraday's law of magnetic induction; it induces an oscillating current in this coil. This is the principle of NMR signal detection. The induced current is the MR signal. It is from this received RF signal that a MR image can be generated.

### 1.3 Radio-Frequency MRI Signal

Radio Frequency (RF) signal is a non-ionizing electromagnetic radiation with a band comprising of radar, ultra high frequency (UHF), and very high frequency (VHF) television, AM and FM radio, and microwave communication. The RF spectrum of the EM radiation utilized for MRI examinations in clinical modalities ranges from 8.5 to 340 MHz. The protons associated with the wavelengths in this band possess insufficient energy neither to ionize the atoms of organic matter nor cause any significant damage at the cellular level [6, 7]. Hence MRI signals can not induce irreversible alterations in living systems via single-photon quantized molecular interactions, but only via multi-photon absorption, i.e. direct heating [15]. However, compared to the ionizing radiation based medical diagnostic
techniques like general radiography (x-ray), positron emission tomography (PET) and computed tomography (CT); MR imaging modality is considered to have lesser health effects.

According to the International Commission on Non-Ionizing Radiation Protection (ICNIRP) guidelines [7], based on the energy absorption properties of the human body, EM frequency spectrum can be categorized as follows:

1. From 100 kHz up to 20 MHz, the absorption in the human trunk decreases rapidly with decreasing frequency and significant absorption may occur in the neck and legs;
2. From 20 MHz up to 300 MHz, relatively high absorption can occur in the whole body, and to even higher values if partial body resonances are considered;
3. From 300 MHz up to several GHz, significant local, non uniform absorption occurs;

   Above 10 GHz, energy absorption occurs primarily at the body surface.

The EM radiation employed in MR spectroscopy is typically in the frequency range of 8.5 to 340 MHz, at which bio-effects due to high RF deposition occur in the whole body. Another distinctive characteristic of the electromagnetic waves can be made based on the distance from the source of the radiation (L). In the case of a “far-field” excitation where L is greater than wavelength of the electromagnetic wave (λ), the electromagnetic radiation may be represented as a propagating wave consisting of transverse electric (E) and magnetic (H) fields. This is referred to the plane-wave approximation where the ratio between E and H is equal to the “wave impedance” in the medium. Alternatively, if L is less than or equal to λ, “quasi-static” approximation can be used and the electric and magnetic fields are effectively
separable. This implies that the field from an excitation source in this “near-field region” is either mainly electric (E >> H) or magnetic [7, 15, 16]. The patient in an MRI examination setup experiences “near-field region” radiation. The biological effects of the RF energy are primarily caused by the magnetic field and the contribution of the electric field is relatively insignificant [6].

1.4 MRI Safety

The rapid advancements in modern technology have made the usage of clinical diagnostic imaging tools, like MRI, a common practice. However, as the possibilities of being recommended for such modality in one's lifetime is increasing, so are the concerns of potential adverse interactions with strong EM fields. During MRI spectroscopy human subject under examination and the individuals in the immediate vicinity of the equipment can be exposed to three types of magnetic fields: static magnetic field, time-varying gradient field and RF electromagnetic field.

The issue of safety under the effect of strong static magnetic field has been an interesting topic of research for more than a century [17]. Several reports state that certain biological regions such as the retina, pineal gland, and some cells in the paranasal sinuses can be affected when humans are exposed to static magnetic fields. However, literature discussed by Budinger [18] and Schenck [19] reveals that these effects are neither harmful nor carcinogenic. According to the latest guidelines from the FDA, clinical MR systems which
employed static magnetic field of intensities up to 8.0 T are considered to be of non-significant risk for human subjects [6].

Time-varying gradient magnetic fields are pulsed during and between the RF excitation pulses during MR imaging. These induce electric fields and circulating currents in the conductive tissues depending on the impedance of the biological tissue and rate of change of the magnetic field. Although induced currents cause minor sensations near the eye, stimulation of nerve cells and muscles, clinical experiments indicate no harmful effects when the rate of change of magnetic flux density is within 6 T/sec [6, 21]. Adequate research and safety standards for gradient magnetic fields associated with MR procedures ensure that there are no potential hazards causing adverse injuries in patients.

1.5 Biological effects of RF Fields

Since the invention of MRI in the early 1980s, several researchers have tried to understand the biological effects associated with the interaction of the RF electromagnetic radiation with the patient’s biological tissues during MR procedures. Until 1985, there were no reports on any quantitative information related to thermal or other physiological human responses. The first experiment on human thermal response to induced heating due to RF fields during MRI was reported by Schaefer in 1985 [23]. His work involved estimation of temperature variations and evaluation of other physiological parameters in subjects exposed to relatively high whole body average SARs of the order of 4.0 W/kg. Similar studies were conducted on volunteers exposed to whole-body average SARs ranging from approximately
0.05 W/kg to 4.0 W/kg [22, 23]. These investigations report that the thermal elevation in the body is limited to less than 0.6°C, mostly observed in the skin tissues, and no associated harmful physiological implications were observed. With the exception of the findings presented at 8 T by Kangarlu et al [24], there have been virtually no published results related to heating of biological tissues due to RF fields during MR procedure.

The importance of characterization of thermal effects in humans to determine the potential hazards with the application of high intensity MR radiation has led many researchers to carry out experimental and computational studies. The distribution of power deposition resulting from the resistive losses in the tissues and coupling of radiation depends on several factors like size, anatomical features, duration of exposure, thermal sensitivity of tissues and a number of other variables [26, 27, 28, 31]. For detailed discussions and extensive analysis of the issues of RF safety during clinical MRI practices, readers can refer to the extensive reviews in [1, 6, 27].

Certain regions of the human body are more susceptible to injuries caused by thermal elevation during exposure to radiation, particularly the testis and eye. The reason for a potential harmful effect is their reduced abilities for heat dissipation [6]. Since the primary focus of this thesis is the evaluation of thermal effects near the eye, the discussion is limited to impact of near-field RF heating to the eyes. Clinical studies show that corneal temperature variations, measured in patients during MRI of the brain using head coils, do not exceed 1.8°C for peak SAR up to 3.1 W/Kg and in the case of peak SAR values ranging from 3.3 to 8.4 W/Kg, the highest corneal temperature measured was 35.1°C [32, 33].
1.6 Implants and Prosthetic Devices

Advancements in integrated microelectronics and modern surgical techniques have enabled the field of biomedical engineering to address the issues related to impaired psycho-physiological functions of the human body. Several researchers and scientists have studied the challenges faced in employing engineering techniques in surgical procedures. The integration of the efforts of many physicians and engineers led to emergence of electronic implantable prosthetic devices to mimic the physiological functionality of human body organs [34]. For example, neuromuscular stimulation [35], voluntary control of internal organs [36], electrical stimulation for receptive sense organs of the body like eye, nose, ear, etc have been some of the aspects investigated by researchers. The performance benefits achieved by prosthetic devices like cochlear implants inspired the efforts of many researchers to develop techniques for artificial electrical stimulation to restore partial vision in the visually impaired. Engineers at NC State University and medical personnel at the Doheny Eye Institute of University of Southern California (USC) have been involved in the development of implantable retinal prosthetic devices. The work presented in this thesis is a part of the collaborative efforts in understanding the bio-safety of the developed implants.

1.6.1 Retinal Prosthesis

The iris of the human eye regulates the amount of light that enters the eye. This light travels to the lens, which focuses it to form an inverted image on the retina at the back of the eye where the information processing begins. The retina consists of 130 million
photoreceptor cells comprising of rods, associated with peripheral vision, and cones which give vivid color for the image [37]. The central region of human vision is governed by the most sensitive portion of the retina which is known as macula. The photoreceptor cells are responsible for the transformation of incident photonic energy into electrical pulses that traverse various cell layers (horizontal, bipolar and amacrine) of the retina before reaching the retinal ganglion cell layer. The axons of these ganglion cells form the optic nerve that relays visual information to the visual cortex in the brain.

Retinitis Pigmentosa (RP) and Age-related Macular Degeneration (AMD) are two of the leading causes of blindness that are responsible for affecting more than 10 million people worldwide [38]. Depletion of photoreceptors in RP causes the gradual loss of peripheral vision which eventually leads to blindness whereas AMD resulting from retinal atrophy causes the loss of central vision. In Unites States, it is estimated that approximately 400,000 people suffer from RP and about 7.6 percent of the people over the age of 40 are affected by AMD [39, 40]. However, even in the people suffering from these adverse conditions due to the loss of photoreceptors, other cell layers (ganglion, horizontal, bipolar and amacrine) retain their functionality. Clinical investigations show that artificial electrical stimulation of the remaining cell layers can restore visual perception in human patients [37, 41, 42]. Particularly, the work presented in this thesis is based on an Epi-Retinal implant used for the stimulation of retinal ganglion cells. Complex micro-electronic devices are needed to provide substantial amounts of power for image and data processing, monitoring bio-functionality of the device and performing back telemetry through inductive links.
1.6.1.1 Epi-Retinal Implant

Clinical investigations have shown that restoration of partial vision is possible by implanting an electrode array at the inner retinal membranes [42, 43]. In Epi-retinal prosthesis, the image acquisition and data processing is accomplished through an external device and thereby reduce the interference with the biological tissue. The extra-ocular unit comprises of a camera which captures the visual information whereas power and data communication between the external and internal devices is performed through a wireless telemetry link via inductive coupling between external and implanted coils [43]. The intra-ocular unit consists of electronic circuits for RF and data recovery and a current stimulator to excite the electrode array so that the visual data is transmitted to the optical nerve. The epi-retinal approach developed by researchers USC is shown in Figure 1.4.
1.7 Motivation

The behavior of human biological system in a strong electromagnetic environment has been a non-trivial field of study. The influence of EM fields plays an important role in maintaining and controlling some of the physiological functions in humans. There is no evidence of evolutionary immunity against the adverse effects caused by exposure to EM radiation and the study of thermal effects due to the transfer of energy from the external fields and the biological tissues has been an interesting area of research for several years. Particularly, investigation of safety due to Radio Frequency (RF) deposition and tissue
heating in humans during Magnetic Resonance Imaging (MRI), especially in the presence of implants, has recently received significant attention in the recent past. The interaction of the RF field with the patient’s tissue can cause significant absorption of RF energy and temperature elevation, which could be enhanced by the presence of a medical implant. Previously, significant work has been done to study this issue of RF heating due to electromagnetic (EM) interactions [44-47]. A few researchers have also addressed the concern of local RF heating due to the interaction of metallic implants with the strong EM field [48].

The MR radiation may be hazardous for patients with certain electronically activated implants, including neuro-stimulation systems, retinal implants, cochlear implants etc due to movement of implants made from metallic materials and also from RF heating and magnetic induction of electrical currents in the ferromagnetic material [49-55]. Interestingly, there were several cases of MRI related injuries in the past that have been caused by misinformation regarding the safety issues of an MRI system with metallic objects, implants, and biomedical devices. For instance, fatal incidents related to MRI of people with cardiac implants and workers with metallic particles in the eye were reported.

Proper characterization of any medical implant and thorough evaluation of the functional and operational aspects of the implants and devices is significant for MR safety. The work presented in this thesis aims to evaluate the interactions between the strong MRI fields and biological tissues of a patient with a Retinal Prosthesis (RP) implant, in terms of SAR and temperature elevation.
1.8 Thesis Organization

In this work, 3-D Expanding Grid Finite-Difference Time-Domain (FDTD) is used to characterize the EM interactions causing MR related heating, induced electric currents and artifacts. Chapter 2 reviews the FDTD formulation of Maxwell’s curl equations employed for the computational analysis. Simulations are performed at three different frequencies, 64MHz, 128 MHz and 256 MHz to evaluate the SAR distribution and extent of thermal elevation in the human head during exposure to MR radiation. This analysis is presented in Chapter 3. Chapter 4 discusses the results obtained from FDTD analysis and concludes with suggestions about future work.
Chapter 2

Computational Methods

With the growing applications of implantable microelectronic devices, the analysis of interactions between the biological tissues and electromagnetic fields enhanced by the presence of an implant has become critical. Efficient computational methods play a significant role in the design prototyping and evaluation of the behavior of the 3-D implanted biomedical devices by computing the interaction of the fields with dielectric media. Various numerical techniques to solve bio-electromagnetic systems using Maxwell’s coupled curl partial differential equations are employed in the field of Computational Electromagnetics (CEM). Among these techniques, Finite-Difference (FD) based methods, Finite-Element Method (FEM), and Method of Moments (MoM) [56-61] have been used to solve complex electromagnetic problems at high frequencies where the dimensions of the model are comparable to the wavelength of the EM sources. Until 1985, frequency domain analyses like MoM and FEM were widely used. However, both these computational methods are implicit techniques with larger memory requirements. The efforts of researchers like Yee and Taflove [56-58] resulted in the development of alternate numerical algorithms which could
be efficiently used for solving bio-electromagnetic problems involving time-evolving EM fields. Also, rapid advancements in high-performance computational infrastructure resulting in computer workstations with large memory help in solving complex systems with arbitrarily heterogeneous and dispersive materials at moderate budgets.

The Finite-Difference-Time-Domain (FDTD) method was first proposed by Yee in 1966 [58]. Over the last three decades, FDTD has been used extensively solving problems involving biological systems. Unlike FEM and MoM, this technique is an explicit scheme and it does not impose upper bound to the number of unknowns it can solve. Another important factor is that the computational time required in the case of FEM is proportional to $N^{>1.5}$ where $N$ is the number of unknowns, whereas in the case of FDTD, the computational time is proportional to $N^{1.33}$ [45]. FDTD allows accurate modeling of large CEM problems with inhomogeneous media. Hence, it is useful in the comprehensive analysis of bio-electromagnetic systems and characterization of the interaction of the EM fields with the biological tissues. The following section discusses the basic FDTD theory and its implementation.

### 2.1 Finite-Difference-Time-Domain Method

Bio-electromagnetic problems involving inhomogeneous dispersive media are traditionally solved using the Finite-Difference Time-Domain (FDTD) method. The FDTD method is based on direct discretization of the Maxwell’s curl equations in time and space.
Using the D-H formulation of the 3-D explicit FDTD method [62], Maxwell’s equations in linear, isotropic, non-dispersive materials can be written as:

\[
\frac{\partial \vec{D}}{\partial t} = c_0 \left( \nabla \times \vec{H} - \vec{J} \right) \tag{2.1}
\]

\[
\frac{\partial \vec{H}}{\partial t} = -\frac{c_0}{\varepsilon_r} \left( \nabla \times \vec{D} \right) \tag{2.2}
\]

The implementation of the finite-difference algorithm follows a time-marching scheme which uses central differences by discretizing the space into number of cells known as the Yee cells [58]. One of the advantages of this approach is that it can be parallelized on high performance computing architectures [63]. Several studies have enabled the development of the FDTD method in terms of identification of stability conditions, modeling of accurate absorbing boundary conditions and frequency dispersive media [64-68]. Taflove [56, 57], Kunz and Luebbers [69] summarize the extensions and development of some of the FDTD techniques.

FDTD analysis of a CEM model involves modeling the region under consideration based on the dielectric properties of the materials. The spatial discretization of any electromagnetic problem and the Courant-Friedrichs-Lewy (CFL) stability bound [57, 58] are determined by very fine geometric details of the modeled objects. The problem space is discretized in the form of cubes (or cuboids) with each grid point (i,j,k) defined as (i\Delta x, j\Delta y, k\Delta z) where \Delta x, \Delta y and \Delta z are the spatial increments in the x, y and z directions respectively. The simulation time is divided into time steps of interval \Delta t. The modeled physical space is
terminated by absorbing boundary conditions in order to prevent the EM fields from reflecting.

Figure 2.1 $D$-field and $H$-field components in a Yee cell
In the Cartesian coordinate system, equations [2.1-2.2] are discretized in time and space using second-order accurate central differences on the Yee grid, shown in Figure 2.1. It can be seen that each \( \vec{D} \)-field component is surrounded by four circulating \( \vec{H} \)-field vector components, and vice versa. Using the six equations resulting from the discretization, \( \vec{D} \) and \( \vec{H} \) field components are determined at staggered spatial and temporal locations [70]. The field quantities are solved using a “leap-frog” scheme where the solutions of the electric and magnetic fields are separated by a half time step. This implementation facilitates the use of central finite difference approximations and the integral form of the Faraday's law and the Ampere's law.

The electric and magnetic field components evaluated at every node of the Yee’s grid can be defined as a function of space and time.

\[
 f \bigg|_{i,j,k}^n = f \left(i \Delta x, j \Delta y, k \Delta z, n \Delta t \right) \tag{2.3}
\]

where, \( n \) is an integer. The spatial and temporal central difference approximations of \( f \) will be of the form given by,

\[
 \frac{\partial f}{\partial t} \bigg|_{i,j,k}^n = \frac{f \bigg|_{i,j,k}^{n+1/2} - f \bigg|_{i,j,k}^{n-1/2}}{\Delta t} + O\left(\Delta t^2\right) \tag{2.4}
\]

\[
 \frac{\partial f}{\partial x} \bigg|_{i,j,k}^n = \frac{f \bigg|_{i+1/2,j,k}^n - f \bigg|_{i-1/2,j,k}^n}{\Delta x} + O\left(\Delta x^2\right) \tag{2.5}
\]

Using the above equations, \( \vec{D} - \vec{H} \) method gives the following equations for electric field, \( \vec{D} \), formulation:
\[ D_x^{n+1}_{i, j+\frac{1}{2}, k} = D_x^n_{i, j, k} + \frac{c_0 \Delta t}{\Delta y} \left( H_z^n_{i, j+\frac{1}{2}, k} - H_z^n_{i, j, k+\frac{1}{2}} \right) - \frac{c_0 \Delta t}{\Delta z} \left( H_y^n_{i, j, k, \frac{1}{2}} - H_y^n_{i, j+\frac{1}{2}, k} \right) \]  

(2.6)

\[ D_y^{n+1}_{i, j, k+\frac{1}{2}} = D_y^n_{i, j, k} + \frac{c_0 \Delta t}{\Delta z} \left( H_x^n_{i, j, k+\frac{1}{2}} - H_x^n_{i, j+\frac{1}{2}, k} \right) - \frac{c_0 \Delta t}{\Delta x} \left( H_z^n_{i, j+\frac{1}{2}, k} - H_z^n_{i, j, k, \frac{1}{2}} \right) \]  

(2.7)

\[ D_z^{n+1}_{i, j, k} = D_z^n_{i, j, k} + \frac{c_0 \Delta t}{\Delta x} \left( H_y^n_{i, j, k, \frac{1}{2}} - H_y^n_{i, j+\frac{1}{2}, k} \right) - \frac{c_0 \Delta t}{\Delta y} \left( H_x^n_{i, j+\frac{1}{2}, k} - H_x^n_{i, j, \frac{1}{2}, k} \right) \]  

(2.8)

The discrete equations for magnetic field, \( H \), formulation are:

\[ H_x^{n+1}_{i, j+\frac{1}{2}, k+\frac{1}{2}} = H_x^n_{i, j+\frac{1}{2}, k+\frac{1}{2}} + \frac{c_0 \Delta t}{\Delta z} \left( D_y^n_{i, j+\frac{1}{2}, k+1} - D_y^n_{i, j+\frac{1}{2}, k} \right) - \frac{c_0 \Delta t}{\Delta y} \left( D_z^n_{i, j+1, k+\frac{1}{2}} - D_z^n_{i, j, k+\frac{1}{2}} \right) \]  

(2.9)
The 3-D explicit FDTD method is explained in detail in [62]. At any time instance, \( n\Delta t \), the \( \mathbf{D} \) field components at any spatial location \((x_i,y_i,z_i)\), where \( i \) is the \( i\)-th cell in the spatial grid, are computed using the \( \mathbf{D} \) field values at time-step \((n-1)\Delta t\) and the \( \mathbf{H} \) field at time instance \((n-1/2)\Delta t\). Similarly, at any spatial location, the \( \mathbf{H} \) field components at time-step \((n+1/2)\Delta t\) are computed using the \( \mathbf{D} \) field values just calculated at time-step \( n\Delta t \) and the previous \( \mathbf{H} \) field values at time instance \((n-1/2)\Delta t\). This “leap-frog” scheme provides unique solution when the values of electric and magnetic fields at time instance \( t=0 \) are specified over the entire space. Also, tangential value of the field components at the boundary for time \( t>0 \) must be specified so that field component at any spatial point can be computed.
2.2 Stability Criterion

The accuracy of the FDTD calculations is governed by the size of the Yee cell. The central difference approach in time and space used in the algorithm is second-order accurate and the error in the approximation is proportional to the square of the increment in time as well as minimum cell size of the grid. In general, Yee cell dimensions are chosen to be less than or equal to one-tenth of the wavelength to achieve desired accuracy. In the case of conventional 3-D uniform FDTD analysis, Courant-Friedrichs-Lewy (CFL) stability bound which limits the time-step, \( \Delta t \), for stability is given as follows [56]:

\[
\Delta t < \frac{1}{C_0 \sqrt{\left(\Delta x\right)^2 + \left(\Delta y\right)^2 + \left(\Delta z\right)^2}}
\]  

(2.12)

where \( C_0 \) is the maximum velocity of light in the region. However, in the case of an expanding grid FDTD algorithm the stability limit governing the time step is:

\[
\Delta t < \frac{1}{C_0 \sqrt{\left(\Delta x_{min}\right)^2 + \left(\Delta y_{min}\right)^2 + \left(\Delta z_{min}\right)^2}}
\]  

(2.13)

It is interesting to note that the minimum time step required to ensure accurate analysis and stability is not smaller than that required in the case of a uniform grid method.
2.3 Absorbing Boundary Conditions

Computational models in bio-electromagnetic problems need to be truncated so as to conform to the available resources to solve them. Absorbing boundary conditions (ABCs) are employed to effectively simulate an unbounded region. The entire computational region consists of an interior region containing the biological model under investigation and an exterior region which is governed by the ABCs. Several approaches have been used for the implementation of ABC [56-57]. In broad sense, two main approaches used so far are – Analytical ABCs, which are employed by approximating the wave equation at the boundary of the model and Material ABCs, which incorporate a complex lossy medium to absorb the incident wave. An approach similar to the latter, the Perfectly Matched Layer (PML) developed by Berenger [66] is one of the most efficient ways of implementing an ABC. The D-H-formulation, which was discussed in Section 2.1, facilitates a flexible way of implementation of split field components PML absorbing boundary condition that is independent of the materials modeled in the FDTD space. The work presented in this thesis employs Berenger’s PML which is discussed in detail in [62, 66].

2.4 Expanding Grid FDTD

Extensive application of FDTD method in the study of bio-electromagnetic systems to evaluate the safety limits can be attributed to its computational efficiency and the necessity to accurately model present miniaturized biomedical implants at fine resolutions. Typically, the resolution of the implant to be modeled is in the order of micrometers. Unfortunately,
traditional explicit FDTD method using uniform grid resolution would need around trillion cells in order to model such a biological system. However, this problem can be resolved by use of efficient modification of traditional FDTD algorithm by implementing a logarithmically expanding grid technique. The work presented in this thesis is based on an extensive use of this method, presented in [71].

The advantage of using an expanding grid FDTD algorithm is two-fold. First, computer memory requirements are greatly reduced. For instance, in order to solve the MRI system with the Retinal Implant under consideration, it would have required more than ten trillion cells to model the biological system using uniform grid. This is clearly indicates that it is unsolvable. However, using logarithmic grid expansion the model size can be easily reduced by a factor of 100 and can be modeled using high performance computer systems with gigabytes of memory. Second, more accurate modeling of the region of the implant is possible where a finer grid with cell sizes of the order of micrometers is used. This helps to model the RF birdcage coil and the human head with the implant as a single system and simulate the EM interactions of the tissues.

In the 3-D expanding grid FDTD algorithm, the equations determining the electric and magnetic field components differ from those discussed earlier. The field at each point in the grid depends on the cell-cell expansion factor, $\alpha$, and the minimum grid resolution, $\Delta_{\text{min}}$. The discretized differential equations for $E - H$ formulation are given by [71]:

\[ \text{Equation not provided here.} \]
\[ C_z\left( i, j, k + \frac{1}{2} \right) = \left[ \frac{\mu_o \Delta z_{\text{min}}}{2 \Delta t} \left( \sigma_z\left( i, j, k + \frac{1}{2} \right) + \varepsilon_z\left( i, j, k + \frac{1}{2} \right) \right) \right]^{-1} \]  

\[ D_z\left( i, j, k + \frac{1}{2} \right) = \left[ \frac{\mu_o \Delta z_{\text{min}}}{2 \Delta t} \left( \sigma_z\left( i, j, k + \frac{1}{2} \right) - \varepsilon_z\left( i, j, k + \frac{1}{2} \right) \right) \right] \]  

\[ E_z^{n+1}\left( i, j, k + \frac{1}{2} \right) = C_z\left( i, j, k + \frac{1}{2} \right)^* \left[ -E_z^n\left( i, j, k + \frac{1}{2} \right)^* D_z\left( i, j, k + \frac{1}{2} \right) \right. 
\left. + \frac{\Delta z_{\text{min}}}{\Delta x_\text{h}\left( i \right)} \left\{ \hat{H}_y^{n+1}\left( i, j + \frac{1}{2}, k + \frac{1}{2} \right) - \hat{H}_y^n\left( i, j - \frac{1}{2}, k + \frac{1}{2} \right) \right\} \right. 
\left. - \frac{\Delta z_{\text{min}}}{\Delta y_\text{h}\left( j \right)} \left\{ \hat{H}_x^{n+1}\left( i + \frac{1}{2}, j, k + \frac{1}{2} \right) - \hat{H}_x^n\left( i - \frac{1}{2}, j, k + \frac{1}{2} \right) \right\} \right] \]  

\[ \hat{H}_z^{n+1}\left( i + \frac{1}{2}, j + \frac{1}{2}, k \right) = \hat{H}_z^n\left( i + \frac{1}{2}, j + \frac{1}{2}, k \right) \right. 
\left. + \frac{\Delta z_{\text{min}}}{\Delta y_\text{c}\left( j \right)} \left\{ E_x^n\left( i + 1, j + \frac{1}{2}, k \right) - E_x^n\left( i, j + \frac{1}{2}, k \right) \right\} \right. 
\left. - \frac{\Delta z_{\text{min}}}{\Delta x_\text{c}\left( i \right)} \left\{ E_y^n\left( i + \frac{1}{2}, j + 1, k \right) - E_y^n\left( i + \frac{1}{2}, j, k \right) \right\} \]  

where, \( \mu_o \) is the permeability in free space, \( \sigma \) is the conductivity of the material and \( \varepsilon \) is the permittivity of the medium.
\[ \Delta x_e(i) = x(i+1) - x(i) \]
\[ \Delta x_h(i) = x\left(i + \frac{1}{2}\right) - x\left(i - \frac{1}{2}\right) \] (2.18)

\[ \mathcal{H}_0 = \sqrt{\frac{\mu_0}{\varepsilon_0}} H \] (2.19)

\[ \Delta t = \frac{\Delta_{\min}}{2C_0} \] (2.20)

The minimum time step required to ensure simulation stability is not smaller than that required in the case of a uniform grid model, as can be seen in equation 2.13. For the simulation results to be valid over the entire region of interest it is important to ensure that the cell size of the human tissue modeled is within the order of \( \lambda_{\varepsilon_r}/10 \), where \( \varepsilon_r \) is the relative permittivity of the tissue. Higher relative permittivity of free space allows the required minimum cell size to be larger than the size within the tissue. Also, number of cells can be significantly reduced by modeling regions which are of lesser significance with coarser resolution.
Chapter 3

Introduction

The importance of characterization of thermal effects in humans to determine the potential hazards with the application of high intensity MR radiation has led many researchers to carry out experimental and computational studies. During MRI spectroscopy human subject under examination can be exposed to three types of magnetic fields: static magnetic field, time-varying gradient field and RF electromagnetic field. The RF spectroscopy used is typically in the frequency range of 8.5 to 340 MHz, at which bio-effects due to high RF deposition occur in the whole body. Radio frequency heating of the human tissues resulting from the interactions with EM fields is primarily caused by ohmic losses and resistive losses in the tissues.

Advances in bioelectronics and surgical procedures have led to emergence of electronic implantable prosthetic devices to mimic the physiological functionality of human body organs. However, evaluation of safety due to RF deposition and tissue heating in patients during MRI, especially in the presence of these implants, has recently received significant attention. Proper characterization of any medical implant and thorough evaluation
of the functional and operational aspects of the implants and devices is significant for MR safety. The work presented in the following sections aims to evaluate the interactions between the strong MRI fields and biological tissues of a patient with a Retinal Prosthesis (RP) implant, in terms of SAR and temperature elevation.

### 3.1 MRI Birdcage Coil

The pulsed radio frequency field is only present during an MRI scan. It can be produced by a variety of transmission RF coils such as a whole body transmit coil or a transmit/receive head coil. However, the adverse biological affects like excessive tissue heating and thermal lesions can be reduced by proper choice of the RF coil. Hence, a transmit/receive head coil is used as the other RF coils can expose more of the biological system to RF energy at the Larmor frequency. Most effective way to couple energy into the nuclei and vice-versa is by using circularly polarized fields that rotate at Larmor frequency.

The RF coil, when used as a transmit coil, must be able to produce a homogenous $B_1$-field in the volume of interest at the Larmor frequency. Also, when used a receive coil, it must have high SNR and must be able to pick up RF signals with the same gain at any point in the volume of interest. Some of the coils used for this purpose are volume coils like Helmholtz coils, saddle coils, High Pass & Low Pass Bird-Cage coils; Surface coils like single loop and multi-loop coils. Of these, the Bird-Cage Coil (BCC) design provides the best field homogeneity. One advantage of using BCC design is that it is easier to generate a uniform $B_1$ RF field over most of the coil's volume, which would result in images with a high
degree of uniformity. Second advantage is that nodes with zero voltage occur 90° away from the driven part of the coil, thus making it possible to use a second signal in quadrature, which produces a circularly polarized field. Third advantage is that the cylindrical symmetry of BCC facilitates quadrature drive which reduces RF power requirements and also improves SNR. For the purposes of the simulations presented in the later sections a structure similar to that of a low-pass Bird-Cage Coil shown in Figure 3.1.

![Capacitor](image)

**Figure 3.1** Schematic of a Low-pass Bird-Cage Coil

### 3.1.1 Coil Excitation

The direction of the RF field, $B_1$, is perpendicular to the direction of the external static magnetic field ($B_0$). There are two different methods employed for the excitation of the coil. In the case of a linearly polarized $B_1$-field, the RF signal is fed at a single point. This
produces an E-field polarized in the direction parallel to the coil’s axis and varies linearly along the vertical direction and is constant in the horizontal direction. However, in order to generate a homogenous RF field circular polarization is the preferred mode of excitation. This is achieved when the coil is fed at two points which are separated spatially by 90 degrees and the signal at both the points is in quadrature. In this case E-field varies linearly in radial direction with minimum at the center of the coil and remains constant in angular direction. The magnetic field is circularly polarized in this case.

A theoretically simplified low-pass Magnetic Resonance birdcage coil, without capacitors, is used for the calculation of SAR distributions presented in this work. The length and the diameter of the coil are chosen as 38 cm and 28.8 cm respectively in order to represent the size of a typical 1.5 T birdcage coil [45]. The birdcage coil with 16 equally spaced rungs or legs is terminated at each end with a circular wire. However, BCC coil model is truncated in the FDTD simulations so as to reduce the computational space. The RF magnetic field is induced by exciting each rung with an axially directed electric field which varies sinusoidally with the azimuthal angle. The phase of the RF sources, with an input impedance of zero Ohms, is shifted progressively around the cage resulting in a uniform, circularly polarized $B_1$ field inside the coil while the magnitude is kept constant. Figure 3.2 depicts circularly polarized instantaneous electric field at every $1/16^{th}$ of a time cycle. The currents in the legs obtained from the simulation, shown in Figure 3.3, are in accordance with excitation.
Figure 3.2 Instantaneous electric field observed at an axial plane through human head model during an MRI.
Figure 3.3 Current distribution in the legs of the Bird Cage coil during the simulation, obtained using a ramped sinusoidal D-field excitation of the rungs

### 3.2 Computational Method

Three-dimensional explicit Finite Difference Time Domain (FDTD) method is employed to quantify the power deposition in terms of SAR and thermal elevation when a human is subjected to MRI radiation at frequencies 64, 128 and 256 MHz which correspond to static field strength of 1.5, 3.0 and 4.7 T respectively. It is critical to estimate the risk of
power deposition by verifying if the SAR values and temperature increase are within the thermoregulatory limits devised by the United States Food and Drug Administration (FDA) [75].

3.2.1 Human Head Model

The entire human head model is developed from a 1mm resolution cross-sections obtained from the National Library of Medicine (NLM) “Visible Man Project”. The dielectric properties of 25 types of tissues at different frequencies of interest are obtained from [72]. The axial, coronal and sagittal views of a cross-section of the human head model can be seen in Figure 3.4.
Figure 3.4 Different slices of Human Head model: (a) Sagittal view; (b) Coronal view; (c) Axial view

The human head model along with the 16-leg low-pass Bird-Cage Coil is modeled as a single system at different frequencies. In the efforts to model the Retinal Implant embedded in the system, the actual computational models are truncated so as to reduce the FDTD space. An axial view of the human head inside the MRI BCC system can be seen in Figure 3.5.
3.2.2 Retinal Prosthesis Implant

The implant used for computational modeling is similar to that used in Epi-Retinal prosthesis. The intra-ocular receiving antenna, which is also referred as the internal coil, is placed in the anterior region of the eye. The implanted micro-chip is placed in the middle of the vitreous cavity. For simulation purposes, size of the chip is taken to be 4mm by 4mm by 0.5mm (exclusive of the 0.5mm insulating material). Figure 3.6 shows a slice of the head model with the implant modeled.
**Figure 3.6** Axial view of human head model with the RP implant showing regions of the micro-chip and receiving antenna (coil).

### 3.2.3 RF Field Homogeneity

One of the critical aspects in the design of MRI Birdcage Coil is to ensure that the RF magnetic field, $B_1$, is reasonably homogenous. The presence of the human head within the resonator can alter the magnetic field homogeneity due to interactions of the RF fields with the biological tissues. The non-uniformities in the RF magnetic field, resulting from the induced additional non-uniform currents, cause inhomogeneous intensity distribution in MR
imaging. Also, the RF electric field, which can be potentially inhomogeneous too, can expose biological tissues to an excessive RF power deposition with induced local heating.

A study of $B_1$-field homogeneity is performed on the Bird Cage Coil in three different conditions, (i) without the head model, i.e., an unloaded BCC system, (ii) when loaded with the human head model without the RP implant embedded and (iii) when loaded with human head model containing the RP implant, at the three frequencies of interest using expanding grid FDTD.

All the field values are normalized such that the absolute value of the $B_1$-field ($|B_1|$), is sufficient to produce a flip angle of 90 degrees. The normalization factor, $N_F$, is determined so that $N_F|B_1| = \frac{\alpha}{(\gamma \tau)} = 1.957 \mu T$ at the coil center, where $\gamma$ is the gyromagnetic ratio for Hydrogen and $\alpha$ (flip angle) is $90^\circ$. This value of $B_1$-field, produced by a rectangular excitation RF pulse with duration ($\tau$) of 3 ms on the axial plane at the center of the coil in a normal MRI system, causes the net magnetization vector to flip by 90 degrees [76].
Case (i): Without Human head Model – Uniform Grid FDTD Analysis

**Figure 3.7** Magnetic field (A/m) distribution in the axial slice for an unloaded BCC system:

(a) 64 MHz; (b) 128 MHz; (c) 256 MHz
Case (ii): Loaded with Human head Model – Uniform Grid FDTD Analysis

Figure 3.8 Magnetic field (A/m) distribution in the axial slice for a BCC system loaded with human head model: (a) 64 MHz; (b) 128 MHz; (c) 256 MHz
Case (iii): Loaded with Human head Model embedded with RP Implant – Expanding

Grid FDTD Analysis

![Figure 3.9](image)

**Figure 3.9** Magnetic field (A/m) distribution in the axial slice for a BCC system loaded with human head model embedded with RP implant: (a) 64 MHz; (b) 128 MHz; (c) 256 MHz

The $B_1$ field distribution is found to be considerably homogenous inside the coil at lower frequencies with slight variation due to the effect of induced currents in the human head. However, the inhomogeneity of the RF field increases with frequency which is
apparent from the results shown. In order to achieve a better uniformity of the magnetic field, magnitude of the excitation signal should be increased as frequency increases. However, this would lead to increase in the power deposition and heating of the biological tissue. In the presence of implant, additional non-uniform currents induced in the implant further deteriorate the magnetic field homogeneity. This is evident from the simulation results shown in Figure 3.9.

### 3.2.4 Expanding Grid FDTD

A 3-D logarithmically expanding grid FDTD method has been used for computational modeling required for the work presented in this thesis. Expanding grid FDTD technique, discussed in detail in Section 2.2, facilitates the use of a finer grid with cell sizes of the order of micrometers which allows the accurate modeling of the implant region. This in turn helps to model the RF birdcage coil and the human head with the implant as a single system and simulate the EM interactions of the tissues. Also, this technique enables to model the implant and its internal coil to capture the effect of RF heating due to the interaction of the surrounding tissues with the EM fields and the influence of currents induced in the internal coil on the temperature elevation, which would be impossible to estimate using uniform grid simulations due to the number of computational cells required.

For the simulation results to be valid over the entire region of interest it is important to ensure that the cell size of the human tissue modeled is within the order of $\lambda_{\text{eff}}/10$, where $\varepsilon_r$ is the relative permittivity of the tissue. Higher relative permittivity of free space allows the
minimum required cell size to be larger than the size within the tissue. The algorithm is employed with the same cell-to-cell expansion factor along the three axes. A finer representation of the region of the eye with the implant at 0.12mm resolution is possible by using an expansion factor of 1.095. The choice of this factor depends on the trade-off between reducing the FDTD simulation space and minimizing the dispersion effects by ensuring smooth transition between cells. An example of one such grid profiles is shown in Figure 3.10. The perfectly matched layer (PML) is used as an absorbing boundary condition. Sixteen layers of PML and a separation of about 15 cells between the PML surface and closest point on the MRI system are used for simulation purposes.

Figure 3.10 Example of Expanding Grid Variation profile
3.3 SAR Evaluation

Over the past 30 years, various computational and experimental techniques have been implemented to understand the interaction of the RF fields with the patient’s tissue. Thermal effects associated with the absorption of RF energy in the tissues are characterized in terms of Specific Absorption Rate of energy (SAR) and temperature elevation. Several techniques have been developed for determination of SAR distributions or induced electric fields and current densities in anatomically based models of the human body. Previously, significant work has been done to study this issue of RF heating due to electromagnetic (EM) interactions [44-47]. A few researchers have also addressed the concern of local RF heating due to the interaction of metallic implants with the strong EM field [48]. In this work, the human head model is implanted with the RP implant and exposed to RF magnetic field in the birdcage coil. The SAR distribution in the head model is calculated to understand the RF heating in the tissues surrounding the metallic implant.

3.3.1 SAR as Dosimeter

Specific absorption rate (SAR), represents the current national and international dosimetric term used to characterize the thermogenic aspects of RF electromagnetic fields [73-75]. As per the definition given by the International Electrotechnical Commission (IEC) standard [73], it is the amount of RF power absorbed per unit of mass of an object indicated in W/kg. Peak 1-voxel SAR patterns are computed using
\[ SAR = \frac{\sigma |\vec{E}|^2}{2\rho} \] (3.1)

\( \sigma (\text{S/m}) \), \( \rho (\text{Kg/m}^3) \) and \( |\vec{E}| (\text{V/m}) \) are the conductivity, the mass density of the human tissue and magnitude of the peak electric field respectively. The primary health concern is the temperature rise produced by the power deposition along with the RF “hot spots”. Thermal elevation is calculated from the SAR distribution and the induced currents in the internal coil of the implant. The electric field \( |\vec{E}|, \text{V/m} \) and SAR values are normalized pertaining to the explanation given in Section 3.2.3. The following sections present results from the simulations performed using expanding grid FDTD analysis for SAR distribution and the thermal elevation at 64, 128 and 256 MHz.

3.3.2 Average 1-g SAR Distribution

The RF energy deposition in human head during exposure to MR radiation is characterized and quantized in terms of 1g SAR so as to compare with the guidelines by IEEE/ANSI [77]. As a part of this evaluation, initially simulations are performed to estimate the SAR levels when human head model does not contain the RP implant, shown in Figure 3.11. An increase in the values of SAR is observed as RF frequency increases. This is due to higher interactions of the RF fields with biological tissues.
Figure 3.11 Peak 1g SAR (W/Kg) distribution in the axial slice for a BCC system loaded with human head model: (a) 64 MHz; (b) 128 MHz; (c) 256 MHz

Usually, biomedical devices are excluded from SAR evaluation because it is assumed that people with implants are under constant clinical observation. Also, the benefits received from the prosthetic devices are likely to outweigh the possible risks due to exposure to MR radiation. However, it is necessary to understand the extent of harmful effects arising from
RF energy deposition. Hence, peak 1-voxel and peak 1g SAR distributions are computed when head model is embedded with RP implant.

Figure 3.12 Electric Field (V/m) distribution in the axial slice for a BCC system loaded with human head model embedded with RP implant: (a) 64 MHz; (b) 128 MHz; (c) 256 MHz

The presence of the implant causes slightly more non-uniformities in the RF field distribution. Also, slightly higher values of SAR are observed near the regions surrounding
the implant. However, these local peaks are averaged when we consider 1g SAR distribution. Hence, the effect of the RP implant is not very pronounced.

![Figure 3.13](image)

**Figure 3.13** Peak 1g SAR (W/Kg) distribution in the axial slice for a BCC system loaded with human head model embedded with RP implant: (a) 64 MHz; (b) 128 MHz; (c) 256 MHz
<table>
<thead>
<tr>
<th>Frequency (in MHz)</th>
<th>Peak 1g SAR (in W/Kg)</th>
<th>Location (Tissue Type)</th>
<th>Induced Current in the Implant (in mA)</th>
</tr>
</thead>
<tbody>
<tr>
<td>64</td>
<td>0.33</td>
<td>Nose (Muscle)</td>
<td>1.63</td>
</tr>
<tr>
<td>128</td>
<td>1.12</td>
<td>Nose (Muscle, Cortical Bone)</td>
<td>3.13</td>
</tr>
<tr>
<td>256</td>
<td>2.27</td>
<td>Nose (Muscle, Cortical Bone)</td>
<td>5.35</td>
</tr>
</tbody>
</table>

In the regions near the nose and brain, more power deposition is noted. The influence of metallic implant increases with frequency as SAR values observed near the regions surrounding the implant increase at higher frequencies. Although highest values of SAR are not found near these regions, there is almost a two-fold increase in the SAR values computed in the tissues around the implant. However, this rise is inconsequential as it is within the safety limit of 8 W/Kg, devised by FDA for MRI diagnostic devices for general population [75].

### 3.4 Thermal Elevation Analysis

In the field of Bio-Electromagnetics, characterization of temperature distribution as a result of electromagnetic power deposition is critical. Certain regions of the human body are more susceptible to adverse affects due to unwanted thermal elevation. Understanding the temperature increase in the tissues of human eye, under the influence of strong...
electromagnetic fields, has been an interesting topic of research because of the low blood flow through the eye tissues [78-79]. Some researchers in the past studied the potential hazards due to temperature increase during an MRI, particularly in the ocular region [32-34]. However, there have been no reports about the influence of the metallic implant which can possibly have a detrimental affect.

Temperature increase evaluation has been implemented using a finite-difference computational method for the solution of the bio-heat equation [80]. Numerical formulation of this equation, including the affect of implant, has been described in detail in [81]. With the consideration of the external thermal sources, the bio-heat equation can be written as

\[
\frac{C_p}{\rho} \frac{\partial T}{\partial t} = \nabla \cdot (K \nabla T) + A - B(T - T_B) + \rho^* \text{SAR} \left[ \frac{W}{m^3} \right]
\]  

(3.2)

where left-hand side denotes the thermal capacitance per unit volume multiplied with temperature increase per unit time. Thermal capacitance is the product of the specific heat of the biological tissue \((C, \text{[J/(Kg.°C)]})\) and mass density of that tissue \((\rho, \text{[Kg/m}^3\text{]}\)). In the right-hand side expression, the first term stands for thermal spatial diffusion signifying heat transfer through conduction (where \(K\) is the thermal conductivity). \(A\) is the metabolic heat production rate of human tissue; \(B\) is the blood perfusion constant. \(T_B\) is the temperature of the blood which is assumed to be 37 degrees for computational purposes. The last term represents the contribution due to the induced RF power deposition. Also, the thermal spatial diffusion term accounts for additional heat transfer between adjacent cells resulting from conduction. This is modeled using a thermal resistance network which is based on the tissue
conductivity properties. The thermal resistance, $R_{th}$, for two adjacent cells of different dimensions, $\Delta x_1$ and $\Delta x_2$, and thermal conductivities, $K_1$ and $K_2$, is given by,

$$R_{th} = \left( \frac{1}{2\Delta x_1 \Delta x_2} \right) \ast \left( \frac{K_1 \Delta x_2 + K_2 \Delta x_1}{K_1 K_2} \right)$$  \hspace{1cm} (3.3)

Alternating Direction Implicit (ADI) method explained in [82] has been used for the computational analysis of thermal elevation.

3.4.1 Temperature Rise Results

SAR distributions obtained using expanding grid FDTD analysis at 64, 128 and 256 MHz are used to evaluate the thermal affects of the RF power deposition. The human head model as well as the RP implant used in this case is similar to those discussed in the earlier sections of this chapter. To estimate the contribution of temperature increase due to the presence of implanted device, the natural base temperature distribution in the biological model in the absence of such external heat sources is computed. The initial temperature for the human head model and the external ambient temperature are taken as 24°C.

It is observed from thermal analysis that there is no significant increase in the temperature due to the presence of the metallic implant. For implantable devices, the guidelines given in [83] state that the maximum temperature of the outer surface of the implant should not be over the normal surrounding body temperature of 37°C by 2°C. Although there is a thermal elevation in the tissue regions surrounding the implant, shown in Figure 3.12, this is within the stipulated temperature rise mentioned.
Figure 3.14 Temperature increase profile in the axial slice for a BCC system loaded with human head model embedded with RP implant: (a) 64 MHz; (b) 128 MHz; (c) 256 MHz
As can be seen from the Figure 3.14, maximum thermal increase is observed in the region close to the position of the implant, in the vitreous humor. However, the vitreous cavity behaves more like a heat sink and the rise in temperature in tissues away from the eye is very small.
Chapter 4

Conclusion

The popular use of MRI in today’s clinical practices makes it a very common imaging modality to undergo in one’s lifetime. It is critical to evaluate the risk of health hazards when a person is subjected to strong MR radiation, particularly if he/she has a metallic implant. Apart from highlighting the necessity to evaluate the affect of conductive implants, neither the International Electrotechnical Commission (IEC) nor the Food and Drug Administration (FDA) document standards about evaluating MR radiation affects on patients with implants. Nevertheless, this work characterizes the electromagnetic energy deposition due to the interaction of RF fields with the biological system during an MRI (at 64 MHz, 128 MHz and 256 MHz). A 3-D explicit Expanding Grid FDTD was used to evaluate the power deposition in terms of Specific Absorption Rate (SAR) distribution within a human head and specifically near the regions surrounding the implant used for Retinal Prosthesis (RP). Expanding grid FDTD method helps to accurately model the detailed anatomical human head model along with RP implant with a significantly reduced requirement of computational resources. A simplified 16-leg low-pass Bird Cage coil is excited such that circularly
polarized RF field is produced in the region within the coil. It was found that, as the frequency increases, homogeneity of the RF magnetic field decreases due to enhanced electromagnetic interactions with the human tissues as well as higher induced currents in the RP implant. Peak 1-voxel and peak 1-g SAR in the human head model are computed. About two-fold increase in the SAR levels is observed in tissues surrounding the implant. However, it is observed that peak 1-g SAR values computed at 64 MHz, 128 MHz and 256 MHz are well within the limits given by FDA for MRI diagnostic devices’ significant risk in general population. Hence, it can be stated that the effect of RP implant is not significant. Usually, patients with implants are under constant clinical observation and in comparison to the benefits of biomedical devices any minor thermal effects arising due to exposure to MR radiation become insignificant.

Temperature elevation resulting from the SAR distribution as well as the induced currents in the RP implant was evaluated using the bio-heat equation. The results obtained from these simulations provide guidelines to estimate the RF energy levels absorbed in the biological system and hence evaluate MRI safety. As the frequency increases, temperature rise is more since the RF absorption increases. Maximum thermal increase is observed in the region close to the position of the implant, vitreous humor. However, this temperature rise is observed to be inconsequential. Thus a comprehensive analysis of RF effects due to EM fields during MR studies is performed.
4.1 Future Work

The computational analysis presented in this work addresses the issue of RF heating during an MRI. Nonetheless increased heating of tissues due to the presence of the metallic implant depended on the dimensions, orientation, shape, and location of the metallic implant in the patient. Also, movement of the eye during examination changes the orientation of the internal coils in the RF field. These are some of the factors that need to be considered in future simulations.

The implanted system consists of an implanted secondary coil, current stimulator chips and an electrode array consisting of micro-electrodes to provide actual excitation to the retinal cells. The resolution of the image that can be viewed by the patient depends on number of electrodes used for stimulation. Hence to comprehensively evaluate the MRI safety in the case of further advancements of the retinal prosthetic system electrode array needs to be modeled accurately along with the chip.
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