EVALUATION OF SAR AND TEMPERATURE ELEVATION IN A
MULTI-LAYERED HUMAN HEAD MODEL EXPOSED TO RF
RADIATION

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July, 2010
DEDICATION

To my lovely family.
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List of Symbols

E  Electric field strength vector (v/m)
H  Magnetic field strength vector (A/m)
\( C_o \) Speed of light in vacuum \((\approx 3 \times 10^8 \text{m/s})\)
f  Frequency (Hz)
\( \varepsilon \)  Permittivity (F/m)
\( \varepsilon_r \) Relative permittivity
\( \varepsilon_o \) Permittivity in vacuum\((\approx 8.85 \times 10^{-12})\)
\( \mu \) Permeability (H/m)
\( \mu_r \) Relative permeability
\( \mu_o \) Permeability in vacuum\((= 4\pi \times 10^{-7})\)
\( \omega \) Angular frequency \((2\pi f)\)
\( \rho \) Density \((\text{kg/m}^3)\)
\( \sigma \) Electric conductivity \((\text{S/m})\)
\( \Delta u \) Cell width in \( u \) direction
\( n \) Time step index
\( \Delta t \) Time step
\( j \) Imaginary unit
\( \gamma \) Propagation constant \((\text{m}^{-1})\)
\( \alpha \) Attenuation constant \((\text{Neper/m})\)
\( \eta \) Intrinsic impedance \((\Omega)\)
\( \eta_o \) Intrinsic impedance in vacuum (\( \approx 377 \Omega \))

**Abbreviations**

**ABC** Absorbing Boundary Condition

**ANSI** American National Standard Institution

**BHE** Bioheat Equation

**CPML** Convolutional Perfectly Matched Layer

**EM** Electromagnetic

**FDTD** Finite Difference Time Domain

**ICNIRP** The International Commission on Non-Ionizing Radiation Protection

**IEEE** Institution of Electrical and Electronic Engineering

**RF** Radio Frequency

**SAR** Specific Absorption Rate

**WLAN** Wireless Local Area Network
In recent years, the wireless communication operators have been using more and more systems based on the transmission and reception of Electromagnetic (EM) waves. This has prompted the public’s concern of the health effects of the use of such wireless systems. To date, the most prominent and scientifically verifiable biological effect of EM waves is the heating effect. The purpose of this thesis is to investigate the effect of Radio Frequency (RF) wave radiated from cellular phones and Wireless Local Area Network (WLAN) antennas on human head. A multi-layered head model is used to evaluate the Specific Absorption Rate (SAR) distribution and temperature elevation. This simple model will enable us to investigate the effect of thin layers on SAR and temperature elevation distributions without the need of high computational resources. The Bioheat equation with suitable boundary conditions will be solved using the Finite Difference Time Domain (FDTD) method in order to find temperature elevation and exposure time effect. Moreover, using phones inside enclosed environments will be tested. Effect of the distance between the head model and the source is studied as well. Aging effect on dielectric properties of tissues will be discussed too. Also, a smaller head size will be used to represent the head of a 5-years old child. An oblique incidence plane wave will be used to excite the head model. Both perpendicular and parallel polarizations are studied. In order to obtain an overview of the consequences of using the cellular phones by people who have a metallic implant inside their heads, a simple model of a homogeneous sphere that has the same properties as the brain will be adopted. This model will be used to verify the results obtained using the planar multi-layered head model. The obtained results confirm the importance of performing a thermal analysis along with the dosimetric one because the relationship between SAR and temperature elevation is not linear. It will be shown that the induced temperature elevation in the brain region, in all the examined conditions, never exceeds 0.4°C. This value is well below the threshold for the induction of adverse thermal effects to the neurons.
Chapter One: Introduction

1.1 Motivation

With the recent rapid increase in the use of portable telephones and wireless local area networks (WLANs), public concern regarding potential health hazards due to the absorption of electromagnetic (EM) energy emitted by these applications has been growing. Safety guidelines for protecting the human body from Radio Frequency (RF) exposure have been issued in various countries.

The most widely acceptable standards are those developed by ANSI/IEEE (American National Standard Institution / Institution of Electrical and Electronic Engineering) and ICNIRP (the International Commission on Non-Ionizing Radiation Protection). These safety guidelines are based on the findings from animal experiments. Sets of maximum tolerable values of specific absorption rate (SAR) have been proposed. SAR parameter has been widely used to determine the possibility of health hazards in the human head due to RF radiation. It is defined as the transferred power divided by the mass of the object. But, very little is known about possible biological effects of localized SAR and the SAR distribution effect based on physiological ground is still unclear.

Since the SAR is a physical quantity, which causes tissue heating due to RF exposure, the safety guidelines on localized SAR for wireless applications should be determined in relation to temperature rise in the head, especially in the brain, which includes the central part governing the body temperature regulation function. This is due to the fact that the biological hazards are mainly due to temperature rise, caused by exposure to RF waves, in the tissue.
From the above point-of-view, in this thesis, the SAR in the human head will be determined inside a simple planar multi-layered human head model. Moreover, the temperature rise in the human head, due to emissions from wireless applications, will be computed by solving the bioheat equation numerically using the Finite Difference Time Domain (FDTD) method with appropriate boundary conditions.

1.2 Proposed Research

This thesis aims to calculate the electric field intensity, SAR and temperature elevation in the head when exposed to RF radiation. The effect of age on dielectric properties of tissues for SAR and temperature elevation will be investigated too. A planar multi-layered model of the human head will be used. This model was originally proposed by Abdalla and Teoh in [1] and consists of 6 planar layers of tissue, as shown in Figure 1.1. It was used in [2] to find SAR distribution due to obliquely incident plane waves. This model is chosen because it does not need extensive numerical techniques for the electric field and SAR calculations, like other complicated models described in Section 1.3. In addition, it will be demonstrated that this model will produce SAR and temperature elevation values that are very close to those obtained using other time consuming and more elaborate models.

1.3 Head Models

Different head models were considered in the literature in studying the effect of EM waves on human head. The simplest model used a semi-infinite homogeneous plane of tissue as a head model [3]. Another model is a homogeneous sphere that represents the brain properties [3]. A simplified physical model of the human head of the user of hand held radio transceiver has been proposed by IEEE and IEC standards for compliance
testing and is called the Specific Anthropomorphic Mannequin (SAM) [4]. SAM consists of a homogeneous head shape sphere. It has a lossless plastic shell and an ear spacer. Khalatbari et al. [5] considered a 3-layered sphere model that represents skin, bone and brain, while Bernardi et al. [6] used an inhomogeneous human head model with 32 tissues and 3 mm cell size.

![Image]

**Figure 1.1:** Planar multi-layered model of human head.

Two problems are faced in these models, first the long execution CPU time needed in order to get the results; in the best case it requires few hours on an ordinary PC. The second problem is that the cell size used in all these models is at least 2.5 mm which is considered low resolution segmentation for head models and thus the effect of thin layers such as skin, fat and Dura is neglected.

Since the number of existing anatomical computer models is small, and their quality is often not sufficient to study the effects of tissues with small thicknesses, a planar six-layer head model will be used in this thesis. This model consists of skin, fat, bone, Dura, CSF and brain (Figure 1.1). This stratified model will enable us to study the effects of variations of tissue thicknesses due to different ages.
1.4 Electromagnetic Regulations

Dosimetry is the science of predicting the dose of the EM field present at any point inside or outside of the body. It is used to determine the amount of power, fields, and current occurring in various parts of the body from different field exposure. Determining if these are safe or not requires an understanding of the biological effects of these fields. These effects have been used to define allowable guidelines or regulations for electromagnetic field exposure. In addition, the allowable exposure strengths are generally specified in the guidelines, as well as allowed frequencies.

Limits on whole-body absorbed power (as defined by SAR) were established by the Institute of Electrical and Electronics Engineers (IEEE C95.1) in 1991 and adopted by the American National Standards Institute (ANSI) in 1992. ICNIRP also issued basic restriction limits, below which there will be no adverse health effects. The numerical values of the basic restriction limits are presented in Table 1.1 [7, 8]. There are limits for both occupational and general public exposure, the latter is more restrictive. The European Council recommendations are the same as the ICNIRP limits for general public exposure.

Table 1.1: SAR basic restriction limits in the frequency range (300 KHz to 300 GHz) used by wireless applications [7, 8].

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<th>Limbs Localized SAR (W/Kg)</th>
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<td>ICNIRP, Occupational</td>
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<tr>
<td>IEEE, General public</td>
<td>1.6</td>
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<tr>
<td>IEEE, Occupational</td>
<td>8</td>
<td>20</td>
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1.5 Thermal Effects

Probably, the best way to understand the electromagnetic effect on the body is to investigate the thermal interactions with the body. Tissue heating is caused by deposition of power from the ionic conductivity of the tissue as well as losses associated with motion of the molecules caused by the time variation of the electromagnetic field. Power deposition is measured by specific absorption rate (SAR). The temperature rise in the tissue is determined by the rate of electromagnetic power deposition in the tissue and the metabolic heating rate, as well as the thermal dissipation by conduction and blood flow. The temperature rise is usually predicted by solving the bioheat equation, which will be presented in section 2.7.

1.6 Numerical Methods

Many numerical methods have been used to evaluate the power absorption into biological bodies. One simple method uses the analogy between the layered media and a uniform transmission line, where each layer is replaced by an equivalent transmission line of the same impedance [1]. Other methods include the Method of Moments (MoM) [9], the Finite Element Method (FEM) [10], and the Finite Difference Time Domain (FDTD) method [11-14]. Whereas in the few works that studied temperature elevation, FDTD method was used to solve the bioheat equation in [15-17], while in [3], temperature rise was estimated by finite element solution of the bioheat equation using PDEase software [18].

The FDTD method was first proposed by Yee in 1966 [19]. It is based on the substitution of each partial derivative in Maxwell’s equations in time domain with its finite difference representation. This substitution leads to a set of equations where each field component of a cell is evaluated at a time step as a function of the adjacent cells’ components which were evaluated in the preceding time steps. To this end, space and time
are divided into discrete intervals in which the electromagnetic field is supposed constant. With reference to space, this leads to the definition of a unit cell (referred to as Yee’s cell) in which the electromagnetic field is supposed constant.

1.7 Literature Review

Abdallah and Teoh [1] investigated SAR effect of perpendicular polarized plane wave, in the frequency range of 100 MHz to 300 GHz in stratified human head model. The solution was computed using the analogy between the layered media and a uniform transmission line, where each layer was replaced with the equivalent transmission line of the same impedance. Results highlighted the dependence of SAR on frequency and dielectric parameters.

Omar et al. [2] investigated SAR distribution due to obliquely incident plane wave exciting a multi-layered human head model at 900 MHz and 1800 MHz. Exact formulas for the electric field within the structure were derived. The maximum power dissipation and its dependence on the frequency and dielectric parameters were presented. Using these results, the local Specific Absorption Rate (SAR) was calculated.

Riu and Foster [3] suggested a uniform homogeneous sphere and a uniform semi-infinite homogeneous plane of tissue as a head model. SAR calculation was made using a commercial FDTD program (XFDTD) [20] and the resulting temperature rise was estimated by the finite element solution of the bioheat equation using PDEase software. The effect of the distance between the RF source and the head model on SAR distribution was investigated. It was also found that temperature elevation reached its steady state value after several minutes of exposure.

Khalatbari et al. [5] calculated SAR in two models of the head. One is a homogeneous sphere with the dielectric properties of brain and the other is a 3-layered sphere that represents skin, bone and brain. Calculations were done using FDTD
simulation software with 2.5 mm cell size [21]. This research confirmed that in most cases the homogeneous sphere resulted in larger SAR values than the layered sphere, which is due to taking the effect of the covering tissues in consideration such as skin and skull.

Bernardi et al. [6] used the FDTD method to evaluate the effect of dual band phones on SAR and temperature elevation in a human head model with 3 mm cell size which is considered a low resolution segmentation and, thus, the effect of thin layers was neglected. Exposure time effect was not investigated. Although peak SAR value was higher at 900 MHz than 1800 MHz, the peak temperature increase was higher at 1800 MHz. The obtained results confirmed the importance of performing a thermal analysis together with the dosimetric one.

Gandhi et al. [16] studied the thermal implications of the SAR limits for the occupational exposures suggested in the commonly used safety guidelines at 835 MHz and 1900 MHz and the head model was segmented with 3 mm cell resolution. Such SAR rates would lead to temperature elevation up to 0.5°C in the brain tissue.

Wang and Fujiwara [17] found that an exposure time of 6-7 minutes yields a temperature rise of approximately 90% of the steady state value. In this paper, the FDTD method with 2.5 mm cell resolution was used to compute the average temperature increase in a Japanese adult head model exposed to 900 MHz and 1.5 GHz RF waves.

Hsing-Yi and Han-Peng [22] stated that little is known about temperature increase in human head and investigated the effect of exposure time on temperature increase due to different models of antennas by solving the bioheat equation. An FDTD software was used with 2 mm cell resolution and second order Mur absorbing boundary condition was applied. The temperature increase in the model reached its steady state value after an exposure time of about 20 minutes.
Zygirdis and Tsiboukis [23] solved the bioheat equation assuming that thermally steady state is reached by using an FDTD simulation software for 2-D head model. In [24], Yioultsis et al. presented an analysis of the impact of various mobile phones and WLAN antennas on a human being. The computational analysis was based on the FDTD method. The study focused on the steady state thermal rise due to power absorption with cell size of 3 mm. Although both SAR and steady state thermal rise in the case of WLAN are one or two orders of magnitude lower than cell phones, the issue of prolonged exposure was raised, since it was found that the safety limits for long exposure are marginally violated.

Bernardi et al. [25] made an electromagnetic and thermal analysis on a three dimensional human head using FDTD with a resolution of 2 mm cell size. Results showed that the steady state temperature increase in brain vary from 0.08°C to 0.19°C and is reached after about fifty minutes of exposure. Thermoregulation mechanism and metabolic increase were neglected.

Gasmelseed and Yunus [26] evaluated SAR in two dimensional human head model at 900 MHz. The head model was exposed to a normal incident plane wave toward coronal and sagittal planes. The FDTD method was implemented using LABVIEW software with a cell resolution of 2 mm [27]. The results showed that the tissue’s average SAR is considerably higher for the sagittal incidence case.

Cooper and Hombach in [28] investigated the effect of metallic passive element on SAR distribution inside a homogeneous spherical head model using MAFIA software [29]. In the worst case, SAR is expected to be doubled by an implement inside the head.

1.8 Thesis Outline

This thesis is organized as follows:

- Chapter 2 introduces wave propagation of normal incident plane wave. Finite difference time domain is introduced, and the used model is presented. The electric fields, magnetic
field, absorbed power, specific absorption rate and temperature elevation, by solving the bioheat equation combined with the appropriate thermal boundary conditions, are calculated in the different layers of the head model. Using the cellular phones inside enclosed environments and testing the effect of distance between the head model and the antenna will be studied also. Moreover, the effect of age on dielectric properties of tissues for SAR and temperature elevation will be tested. Variation of layers thickness due to age is discussed too.

- Chapter 3 presents wave propagation of obliquely incident plane wave with parallel and perpendicular polarizations. The SAR and temperature increase distributions will be evaluated for both types of polarizations.

- Chapter 4 studies the effect of metallic implants inside the human head by using a spherical head model that represents the brain with a metallic implant present inside the model. This model will be used to verify the results obtained using the multi-layered planar head model.

- Chapter 5 concludes the findings and addresses the recommended future work.
Chapter Two: Normal Incidence

2.1 Introduction

In this chapter, the multi-layered model of a human head irradiated by an electromagnetic plane wave with normal incidence is studied. First, electric and magnetic field interactions with living tissues will be discussed. After that, a description of the used head model and the electric properties of the tissues will be presented. Then, the Finite Difference Time Domain (FDTD) method will be described. Specific Absorption Rate (SAR) will be introduced as well as temperature elevation and modeling its distribution using the bioheat equation. The SAR in each layer is calculated. The exposure time effect and the steady state thermal elevation distribution will be evaluated. Many curves are presented showing the electromagnetic (EM) wave behavior in each layer. The achieved SAR distribution results will be verified using the method presented in [2]. Thermal elevation distribution will be compared to the results obtained using another elaborate human head model. It will be shown that the results are very close to each other, with small differences due to the different cell resolution used.

2.2 Electric and Magnetic Field Interactions with Living Tissues

One of the most important aspects of bio-electromagnetics is how electromagnetic fields interact with materials, for example, how the electric (E) and magnetic (H) fields affect the human body. Because E and H were defined to account for forces among
charges, the fundamental interaction of $E$ and $H$ with materials is that they exert forces on the charges in the materials.

Biological materials are lossy, and this loss changes the way the wave interacts with the material and its propagation behavior. A material is lossy if the conductivity $\sigma \neq 0$. Power will be dissipated in the lossy material as a wave passes through it, thus causing loss to the propagating wave. If power is dissipated in the material, the material will heat up, and this is what raises the concerns of RF waves’ effect on human tissues [30].

In many electromagnetic field interactions, energy transfer is of prime consideration and concern. For example, in hyperthermia for cancer therapy, the electric field is transformed in the body into heat, which is the desired outcome of the therapy. For cell phones, the energy transfer must be below some predefined regulations. The $E$ field can transfer energy to electric charges through the forces it exerts on them, but the $H$ field does not transmit energy to charges. $H$ effect is not prominent in EM biological interactions. For steady-state EM fields, the electric power density is given by

$$p = \sigma \frac{E^2}{2} \ (W/m^3)$$  \hspace{1cm} (2.1)

2.3 Description of the Proposed Model

The human head model considered in this thesis is a stratified medium that consists of six layers as shown in Figure 2.1. The construction of the layered media consists of the skin of thickness 0.07 cm, fat of thickness 0.16 cm, bone of thickness 2.05 cm, Dura of thickness 0.05 cm, cerebrospinal fluid (CSF) of thickness 0.2 cm, and the brain of infinite thickness [1].
2.4 Electrical Properties of the Human Head

The electrical properties of human tissues (relative permittivity $\varepsilon_r$ and conductivity $\sigma$) control the propagation, reflection, attenuation of electromagnetic fields in the body. These properties depend strongly on the tissue type and the frequency of interest. As the water content in the tissue increases, the conductivity increases. Figure 2.2 [31] shows the electrical properties of skin tissue as a function of frequency. As the frequency increases, the conductivity increases and the permittivity decreases.

Table 2.1 shows the electrical properties of the different tissues in the head model at 900, 1800 and 2400 MHz, which are the commonly used frequencies for cellular phones and wireless local area networks [31]. It is worth mentioning that the body is so weakly magnetic such that, generally, the relative permeability ($\mu_r$) is assumed to be 1.
2.5 Finite Difference Time Domain (FDTD) Method

2.5.1 Introduction

Computational electromagnetics plays a vital role in assisting engineers and researchers to investigate the behavior of various structures and the interaction of fields...
with dielectric media (which can possess material inhomogeneities). Numerical methods are used to calculate the internal fields. These numerical methods consist of solving Maxwell’s equations using some kind of computer technique.

The existing numerical techniques solve Maxwell's coupled curl partial differential equations involving electric and magnetic fields. Amongst the numerous techniques available for numerically solving partial differential equations, three have been widely used for solving Maxwell's equations at high frequencies: the Method of Moments (MoM) [9], the Finite Element Method (FEM) [10] and the Finite Difference Time Domain Method (FDTD) [11-14].

FDTD method is the most prevalent method for bio-electromagnetic dosimetry today. Both MoM and FEM are frequency domain methods and were used widely until about 1985, but lately, because of their implicit computational technique, the memory limitations have restricted their use to antenna and microstrip problems. In contrast, the FDTD is an explicit scheme and there is no intrinsic upper bound to the number of unknowns it can solve. Also, being a time domain technique, it treats impulsive and non-linear behavior and allows investigating the time of exposure on the SAR and temperature elevation distributions. FDTD allows accurate analysis of configurations of inhomogeneous media. Hence, FDTD is essential in bio-electromagnetics and in the characterization of the interaction of EM fields with the tissues of the human body. This section presents the basic theory about the FDTD method and discusses the salient features of the Absorbing Boundary Conditions (ABCs).

**2.5.2 Finite Difference Time Domain Method** [11-14]

Mathematically, FDTD is just a direct discretization and implementation of Maxwell's curl equations in time and space. It is capable of predicting broadband response
since it is in the time domain and can also analyze lossy and anisotropic materials, ferrites
and plasmas. FDTD simulation of any problem includes dividing the region under
consideration into different material properties, by modeling the proper structure to be
analyzed, and finally the unbounded region is terminated by absorbing boundary
conditions to prevent the wave from reflecting back. Next, the physical space of the
problem is discretized in the form of spatial cells and the time of simulation is also divided
into $\Delta t$ interval steps. The source of EM energy is then introduced into the physical
domain and allowed to interact with the structure. After that, the time domain information
is processed to determine the electromagnetic behavior of the structure. Maxwell's curl
equations involving $\mathbf{E}$ and $\mathbf{H}$ for any linear, isotropic, source-free, media with material
constants $\varepsilon, \mu$ and $\sigma$; are given as:

$$\nabla \times \mathbf{H} = \sigma \mathbf{E} + \varepsilon \frac{\partial \mathbf{E}}{\partial t}$$ (2.2)

$$\nabla \times \mathbf{E} = -\mu \frac{\partial \mathbf{H}}{\partial t}$$ (2.3)

These equations can have unique solutions when the following conditions are
specified:

(i) The value of the fields at $t = 0$ must be specified in the whole domain.

(ii) The tangential components of $\mathbf{E}$ and $\mathbf{H}$ on the boundary of the domain
must be given for all $t > 0$.

$\mathbf{E}$ and $\mathbf{H}$ are vectors in three dimensions. For one-dimensional case, the problem
geometry and field distribution are constants in two of the three dimensions. Figure 2.3
illustrates incidence and reflection of a plane wave that impinges normally on the interface
of human head model. The incident plane wave is represented by $\mathbf{E}_i$, $\mathbf{H}_i$, and $\mathbf{k}_i$, which are
the electric field intensity, magnetic field intensity, and propagation vector, respectively,
of a plane wave propagating in free space. The wave is said to be normally incident
because the direction of propagation $k_i$ is normal to the model interface, which makes the $E_i$ and $H_i$ vectors tangential to this surface.

Figure 2.3: A plane wave normally incident on the interface of a human head model.

In this model for normal incidence, it is assumed that only $E_x$ and $H_y$ components exist, so equations (2.2) and (2.3) become:

$$\frac{\partial E_x}{\partial t} = -\frac{1}{\varepsilon_0\varepsilon_r} \frac{\partial H_y}{\partial z} - \frac{\sigma}{\varepsilon_0\varepsilon_r} E_x \quad (2.4)$$

$$\frac{\partial H_y}{\partial t} = -\frac{1}{\mu} \frac{\partial E_x}{\partial z} \quad (2.5)$$

These are the equations of a plane wave with the electric field oriented in the $x$-direction, the magnetic field oriented in the $y$-direction, both traveling in the $z$-direction.

Finite difference approximations are used to solve the above mentioned partial differential equations. For higher accuracy, a central difference scheme is utilized:

$$\frac{\partial E}{\partial t} \bigg|_{u_0} = \frac{f(u_0 + \frac{\Delta u}{2}) - f(u_0 - \frac{\Delta u}{2})}{\Delta u} \bigg|_{\Delta u \to 0} + O(\Delta u)^2 \quad (2.6)$$

FDTD is implemented by discretizing the space into a number of cells. Discretization means that both $E$ and $H$ should be determined only at discrete spatial locations $(x_i, y_i, z_i)$ where $i$ is the $i^{th}$ cell in the spatial lattice. As shown in the Figure 2.4, both $E$ and $H$ are
interleaved such that their components’ positions satisfy the differential form of Maxwell's equations. Also, both the $\mathbf{E}$ and $\mathbf{H}$ field components are off by half a time step leading to a leap-frog scheme. Therefore, the electric field components are calculated at integer time steps, and magnetic field components are calculated at half-integer time steps, and they are offset from each other by $\frac{\Delta t}{2}$ as shown in Figure 2.4.

Equations (2.4) and (2.5) are discretized in time and space leading to the following equations [11]:

$$\frac{E_x^{n+1}(k) - E_x^n(k)}{\Delta t} = -\frac{1}{\varepsilon_0\varepsilon_r} \frac{H_y^{n+\frac{3}{2}}(k+\frac{1}{2}) - H_y^{n-\frac{1}{2}}(k-\frac{1}{2})}{\Delta z} - \frac{\sigma}{\varepsilon_0\varepsilon_r} \frac{E_x^{n+1}(k) - E_x^n(k)}{2}$$  \hspace{1cm} (2.7)

$$\frac{H_y^{n+\frac{1}{2}}(k+\frac{1}{2}) - H_y^{n-\frac{1}{2}}(k+\frac{1}{2})}{\Delta t} = -\frac{1}{\mu} \frac{E_x^{n+1}(k+1) - E_x^n(k)}{\Delta z}$$  \hspace{1cm} (2.8)

In the previous two equations, time is specified by the superscript, where $n$ actually means a time $t=n\Delta t$. While the indexes define the positions of the field nodes such that $z = k\Delta z$. Equations (2.7) and (2.8) can be rearranged in an iterative algorithm:

$$E_x^{n+1}(k) = E_x^n(k) \left[ \frac{2\varepsilon_0\varepsilon_r - \Delta t \sigma}{2\varepsilon_0\varepsilon_r + \Delta t \sigma} \right] \left[ H_y^{n+\frac{1}{2}}(k + \frac{1}{2}) - H_y^{n-\frac{1}{2}}(k - \frac{1}{2}) \right]$$  \hspace{1cm} (2.9)

$$H_y^{n+\frac{1}{2}}\left( k + \frac{1}{2} \right) = H_y^{n-\frac{1}{2}}\left( k + \frac{1}{2} \right) - \frac{\Delta t}{\Delta x \mu} \left[ E_x^n(k+1) - E_x^n(k) \right]$$  \hspace{1cm} (2.10)

Time is implicit in the FDTD method, so in programming, the superscripts that represent time, are gone and time is represented by the FOR loop iterations. Position, however, is explicit. The only difference is that $k + \frac{1}{2}$ and $k - \frac{1}{2}$ are rounded off to $k$ and $k + 1$ in order to specify position as a vector in the Matlab program.
2.5.3 Cell Size Determination

Choosing the cell size to be used in an FDTD formulation is similar to any approximation procedure: enough sampling points must be taken to ensure that an adequate representation is made. A good rule of thumb is to use 10 cells per wavelength. Literature has shown this to be adequate, with inaccuracies appearing as soon as the sampling drops below this rate.

Naturally, the worst-case scenario must be assumed, and this will involve looking at the highest frequencies that will be simulated and determining the corresponding wavelength. For instance, suppose we are running simulations at 2.4 GHz. In free space, EM energy will propagate at the wavelength of:

$$\lambda = \frac{c_0}{f} = 0.125 \text{ m}$$  \hspace{1cm} (2.11)

where $c_0$ is the light speed in free space. Then, the spatial step can be chosen to be:

$$\Delta z \leq \frac{\lambda}{10} = 1.25 \text{ cm}$$  \hspace{1cm} (2.12)
But, since the tissues in the head have thin thicknesses, steps must be chosen to be small enough to represent all tissues with very small averaging. Thus, a cell size of 0.1 mm will be used.

2.5.4 Stability Criterion of the Time Step

The numerical algorithm is second order accurate in both space and time and being an explicit method, it requires an upper bound on the time step $\Delta t$ for stability. If $\Delta t$ exceeds this bound, the algorithm tends to compute results which are non-physical and increase without limit. An electromagnetic wave propagating in free space cannot go faster than the speed of light. To propagate a distance of one cell requires a minimum time of $\Delta t = \Delta z/C_0$. When getting to two-dimensional simulation, the propagation in the diagonal direction has to be allowed, which brings the time requirement to $\Delta t = \Delta z/(\sqrt{2}C_0)$, assuming that discretization is equal in both directions, i.e., $\Delta z = \Delta x$. Obviously, three-dimensional simulation requires $\Delta t = \Delta z/(\sqrt{3}C_0)$, assuming that discretization is the same in the three dimensions $\Delta z = \Delta x = \Delta y$. This is summarized by the well-known "Courant Condition":

$$\Delta t \leq \frac{\Delta z}{\sqrt{n}C_0}$$

(2.13)

where $n$ is the number of dimensions considered in the problem. Throughout this thesis, $\Delta t$ will be determined by:

$$\Delta t = \frac{\Delta z}{2C_0}$$

(2.14)

which achieves the Courant stability condition.

2.5.5 Absorbing Boundary Conditions (ABC’s)

It is necessary to terminate the computational domain to conform to the available computational resources. Most electromagnetic problems entail a structure whose behavior
is to be studied in an unbounded domain. Since it is impossible to simulate infinite physical domains, the computational space is terminated by absorbing boundary conditions which effectively simulate an unbounded region by preventing reflections from the edges.

ABCs can be achieved in several ways and are classified into either analytical ABCs or material ABCs. Analytical ABCs are simulated by approximating the wave equation at the boundary while the material ABCs incorporate a lossy medium to physically absorb the incident wave. In this research, for one dimensional problem, simple analytical ABC will be employed. For three dimensional problem, Convolutional Perfectly Matched Layer (CPML) ABC will be used.

Normally, in calculating the electric field, knowing the surrounding $H$ values is essential; this is a fundamental assumption of the FDTD method. At the edge of the problem space the value at one side is not available. However, it is known that there are no sources outside the problem space. Therefore, the fields at the edge must be propagating outward. These two facts will be used to estimate the value at the end by using the value next to it [32].

Suppose we are looking for a boundary condition at the beginning of the head model, where $k = 0$. If a wave is going toward a boundary in free space, it is traveling at a speed $C_0$, the speed of light. So, in one time step of the FDTD algorithm, it travels

$$distance = C_0 \Delta t = C_0 \frac{\Delta x}{2C_0} = \frac{\Delta x}{2}$$  \hspace{0.5cm} (2.15)

This equation basically explains that it takes two time steps for a wave front to cross one cell. So, a common sense approach tells us that an acceptable boundary condition could be
\[ E_x^n(0) = E_x^{n-2}(1) \]  \hspace{1cm} (2.16)

It is relatively easy to implement this condition. Simply, the value of \( E_{x}(1) \) is stored for two time steps, and then used in \( E_{x}(0) \). Boundary conditions such as these have been implemented at the left end for \( E_{x} \), while for the right end, the dielectric properties of the last layer (brain) in consideration have to be taken into account:

\[
\text{distance} = V \Delta t = \frac{c_0 \Delta z}{\sqrt{\varepsilon_r} \Delta z} = \frac{\Delta z}{2 \sqrt{\varepsilon_r}} \approx \frac{\Delta z}{S}
\]  \hspace{1cm} (2.17)

where \( V \) is the speed of the wave inside the tissue and \( S \) is equal to or higher than two. Distance that the wave has travelled depends on the dielectric properties of the medium. At the right boundary, the brain, whose dielectric properties vary with frequency, exists. So, the number of time steps, needed for the wave to travel one cell, will vary according to the frequency of interest. This equation basically explains that it takes \( S \) time steps for a wave front to cross one cell. So, a common sense approach tells us that an acceptable boundary condition could be

\[ E_x^n(KE) = E_x^{n-S}(KE - 1) \]  \hspace{1cm} (2.18)

where \( KE \) is the cell at the end of the model. The value of \( E_{x}(KE-1) \) has to be stored for \( S \) time steps, and then used in the calculation of \( E_{x}(KE) \).

2.5.6 Excitation Source

The excitation signal can be a sinusoidal signal or a pulse which has a finite duration. For pulse excitation, the approach used can be summarized as follows:

1- Excite the problem using a Gaussian pulse which has the following expression:

\[ E_x = E_o e^{-\left(\frac{(r-r_0)}{r_{\text{decay}}}\right)^2} \]  \hspace{1cm} (2.19)
where \( n_{\text{decay}} \) is the standard deviation of the Gaussian pulse, and the maximum frequency is

\[
f_{\text{max}} = \frac{1}{2\Delta t n_{\text{decay}}} \tag{2.20}
\]

It should be insured that the frequency of interest is within the Gaussian pulse band. The time waveform of the pulse is centered at time-step \( n_0 \). If a smooth transition from zero into the Gaussian pulse is required, \( n_0 \) should be taken at least \( 3n_{\text{decay}} \). \( E_0 \) is chosen in order to give us the required amplitude of the EM wave in frequency domain at the frequency of interest. For instance, an amplitude of 300 V/m is needed. It was found that the values that give this amplitude are: center of the pulse \( (n_0) \) equal 400, standard deviation \( (n_{\text{decay}}) \) of 120 and \( E_0 = 1.54 \) at 900 MHz, \( E_0 = 1.94 \) at 1800 MHz and \( E_0 = 2.5 \) at 2.4 GHz. Figure 2.5 shows a typical Gaussian pulse in time domain.

![Figure 2.5: Gaussian pulse in time domain \((n_0=400, n_{\text{decay}}=120, E_0 =1.54)\).](image)

2- Record the desired electric field and magnetic field at every time step, \( n=1,\ldots, N_{\text{steps}} \).

3- Calculate the discrete Fourier transform for the desired quantity, which is calculated using the following formula:
\[ E = \sum_{n=1}^{N_{\text{steps}}} E^n e^{-j2\pi f n \Delta t} \]  

(2.21)

Figure 2.6 shows the Fourier transform of the Gaussian pulse shown in Figure 2.5.

Figure 2.6: Fourier transform of the Gaussian pulse of Figure 2.5 (\(\Delta t = 0.166\) Pico sec).

In theory, the results for a wide range of frequencies can be acquired by just one simulation. For human SAR computations, however, wide frequency range has limited practical benefits since the human tissues parameters are frequency dependent, which would be a great computational burden. So, the Gaussian excitation results will be correct only at a single frequency where the tissues parameters are defined. Both types of excitation sources (sinusoidal signal and Gaussian pulse) have been used and gave exactly the same results.

Antennas are assumed to be isotropic. For practical considerations and assuming the worst case, the total power radiated by the cellular phone antenna is assumed to be 0.6 W, and the distance between the head and the phone is assumed to be 2 cm. From this data, the electric field is calculated and found to be 300 V/m using the following formula:
\[ P = \frac{R^2}{2\eta_o} (4\pi R^2) \]  

(2.22)

where \( P \) is the total transmitted power, \( R \) is the distance between the phone and the head, \( \eta_o = 377\Omega \) is the intrinsic impedance of free space.

### 2.6 Specific Absorption Rate (SAR)

The specific absorption rate (SAR) is defined as the dissipated power divided by the mass of the object. SAR is the basic parameter that institutions take into consideration for the evaluation of the exposure hazards in the RF and microwave range. “Specific” refers to the normalization to mass, and “absorption rate” refers to the rate of energy absorbed by the object.

The maximum temperature generally occurs in the tissue region with high heat deposition. However, one should note that SAR and temperature distribution may not have the same profile, since temperature distribution can also be affected by the environment or imposed boundary conditions. Several methods can be used to determine the SAR distribution induced by various heating applicators [33].

One method is the experimental determination of the SAR distribution based on the heat conduction equation. The experiment is generally performed on a tissue equivalent phantom gel. The applicability of the SAR and temperature elevation distributions measured in the phantom gel (to that in the living tissue) depends on the electrical properties of the phantom gel. The electrical properties depend on the electromagnetic wave frequency and water content of the tissue. The ingredients of the gel can be selected to achieve the same electrical characteristics of the tissue for a specific electromagnetic wavelength.

The basic ingredients of the gel are water, formaldehyde solution, gelatin, and NaCl. Water was used to achieve similar water content as the tissue. Formaldehyde and
gelatin were the solidification agents. NaCl was added to obtain the desired electrical conductivity of tissue at that frequency.

The simplest experimental approach to determine the SAR distribution is from the temperature change when power source is ON. In this approach, temperature sensors are placed at different spatial locations within the gel. Before the experiment, the gel is allowed to establish a uniform temperature distribution within itself. As soon as the initial heating power level is applied, the gel temperature is elevated and the temperatures at all sensor locations are measured and recorded by a computer. The transient temperature field in the gel can be described by the heat conduction equation:

\[
C_p(z)\rho(z) \frac{\partial T(z,t)}{\partial t} = K(z)\nabla^2 T(z,t) + \rho(z)\text{SAR}(z), \quad t = 0
\]  

(2.23)

where \( T \) is the temperature of the tissue, \( K \) is the thermal conductivity of the tissue (W/(K.m)), \( C_p \) is the specific heat of the tissue (J/(K.Kg)) and \( \rho \) is the tissues density (Kg/m³). Within a very short period after the heating power is ON, heat conduction can be negligible if the phantom gel is allowed to reach equilibrium with the environment before the heating. Thus, the SAR can be determined by the slope of the initial temperature rise, i.e.,

\[
\text{SAR} = C_p \left( \frac{\partial T}{\partial t} \right) \bigg|_{t=0}
\]  

(2.24)

Neglecting the heat conduction and assuming that SAR at each spatial location is constant during the heating, the temperature rise at each location is expected to increase linearly. It should be noted that this method is an approximate method that assumes a linear relationship between SAR and temperature elevation. Some other assumptions are also made that makes this method inaccurate.
Another method to find SAR distribution is by deriving it from Maxwell’s equations. \( \mathbf{E} \) and \( \mathbf{H} \) are first determined analytically or numerically from Maxwell’s equations. The SAR (W/kg) is then calculated by the following equation:

\[
SAR_i = \frac{\sigma_i |E_i|^2}{2 \rho_i}
\] (2.25)

where \( \rho_i \) is the \( i^{th} \) tissue density (Kg/m\(^3\)). Since the electric fields are now available, the dissipated power density in each layer can also be calculated using the following equation.

\[
P_i = \frac{\sigma_i |E_i|^2}{2} = \rho_i \cdot SAR_i
\] (2.26)

Equation (2.25) is a point relation, so it is often called the local SAR. The space-average SAR for a body or a part of the body is obtained by calculating the local SAR at each point in the body and averaging over the part of the body being considered. This method is feasible when the derivation of the electromagnetic field is not very difficult. It generally requires a rather large computational resource and a long calculation time when using an ordinary PC. The values of \( \rho_i \) for the different tissues of the model are given in Table 2.2 [34]. It is noticed that SAR varies directly with the conductivity. Generally speaking, the tissues with higher water content, such as skin and CSF, are more lossy for a given electric field magnitude than drier tissues, such as bone and fat, as can be seen from Table 2.1.

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Skin</th>
<th>Fat</th>
<th>Bone</th>
<th>Dura</th>
<th>CSF</th>
<th>Brain</th>
</tr>
</thead>
<tbody>
<tr>
<td>( \rho ) (Kg/m(^3))</td>
<td>1100</td>
<td>920</td>
<td>1850</td>
<td>1050</td>
<td>1060</td>
<td>1030</td>
</tr>
</tbody>
</table>

Electromagnetic dosimetry is the science of predicting the dose of the electromagnetic field present at any point inside or outside the body. For instance, it is
used to predict the strength of fields in the head from cell phones to determine if a particular design meets regulatory guidelines. Dosimetry consists of two main parts: First, the incident $E$ and $H$ fields must be defined; second, the $E$ and $H$ fields inside the tissues must be determined. The relationship between the incident EM fields and the internal EM fields is a function of the frequency of the incident fields, and the electromagnetic properties of the body. However, the relationship between the incident fields and the internal fields is very complicated.

2.7 Temperature Elevation in the Tissues [30, 33, 35]

2.7.1 Introduction

Over the past 100 years, the understanding of the thermal properties of human tissues and the physics that governs biological processes have been greatly advanced by the utilization of fundamental engineering principles in the analysis of many biological heat and mass transfer applications. During the past decade, there has been an increasingly intense interest in bioheat transfer phenomena, with particular emphasis on diagnostic and new technology applications effect on living tissues. Relying on advanced computational techniques, the development of mathematical models has greatly enhanced the ability to analyze the bioheat transfer process. The collaborations among physiologists, physicians, and engineers in the bioheat transfer field have resulted in improved treatment, preservation, and protection techniques for biological systems, including the protection of humans from extreme environmental conditions.

In this section, the fundamental aspects of bioheat transfer will be introduced, as well as the required boundary conditions. The bioheat equation will be solved using the FDTD method.
2.7.2 Fundamental Aspects of Bioheat Transfer

One of the remarkable features of the human thermoregulatory system is that a core temperature near 37°C over a wide range of environmental conditions and during thermal stress, such as exposure to RF waves, can be maintained. The amount of blood flow to the body varies over a wide range depending upon the need for its functions.

As for the function of heat transfer for systematic thermoregulation, blood is known to have a dual influence on the thermal energy balance. It can be a heat source or sink, depending on the local tissue temperature. During winter time, blood is transported from the heart to warm the rest of the body. Theoretical study has shown that during exercise, our body temperature would typically rise 12°C in one hour if no heat was lost by the blood flow.

Blood perfusion rate is defined as the amount of blood supplied to a certain tissue region per minute per 100 g tissue weight. In most situations, it is representing the nutrient need in that tissue area. High blood perfusion is also associated with heat dissipation during exercise or thermal stress.

Knowledge of thermal properties of biological tissues is fundamental to understanding the heat transfer processes in the biological system. This knowledge has increased due to its importance in viewing the concerns for radiological safety with microwave irradiation.

Tissue temperature distributions during exposure to RF waves can be determined by solving the bioheat transfer equation, which considers the contributions of heat conduction, blood perfusion, and external heating. In addition to geometrical parameters and thermal properties, the SAR distribution induced by the external heating device should be determined first. All this information, with appropriate boundary and initial conditions, allows one to calculate the temperature distribution of the tissue.
2.7.3 Pennes Model of the Bioheat Equation (BHE)

Pennes [36] published the basic work on developing a quantitative basis for describing the thermal interaction between tissues and perfused blood. His work consisted of a series of experiments to measure temperature distribution as a function of radial position in the forearm of nine human subjects. A butt-junction thermocouple was passed completely through the arm via a needle inserted as a temporary guide way, with the two leads exiting on opposite sides of the arm. He measured the temperature as a function of radial position within the interior of the arm. The environment in the experimental suite was kept thermally neutral during experiments.

Pennes proposed a model to describe the effects of metabolism and blood perfusion on the energy balance within the tissue. These two effects were incorporated into the standard thermal diffusion equation, which is written in the following form:

\[ C_p(z) \rho(z) \frac{\partial T(z, t)}{\partial t} = \nabla \cdot (K(z) \nabla T(z, t)) + \rho(z) \text{SAR}(z) + Q(z) - B(z)[T(z, t) - T_b] \]  

(2.27)

where \( T_b \) is the temperature of the blood, \( Q \) is the metabolic heat generation, and \( B \) is the term associated with blood perfusion.

The temperature elevation due to handset antennas can be considered as sufficiently small not to activate the thermoregulatory response; including the activation of sweating mechanism. Thus, this response is neglected in our study. Additionally, the blood temperature is assumed to be spatially and timely constant, since the EM power absorption due to antennas (output power of less than 1 W) is much smaller than the metabolic heat generation of an adult (approximately 100 W). Then, Equation (2.27) is simplified as follows:

\[ C_p(z) \rho(z) \frac{\partial T(z, t)}{\partial t} = K(z) \nabla^2 T(z, t) + \rho(z) \text{SAR}(z) - B(z)(T(z, t) - T_b) \]  

(2.28)
Next, boundary conditions are needed to account for the heat exchange between the body surface (namely, the skin) and the external environment, in the formulation of the BHE as represented by the following equation

\[
K(z_{\text{min}}) \frac{\partial \tau(z,t)}{\partial z} = -h(T(z,t) - T_a) + SW(z,T(z,t))
\]

where \( h \), \( T \), and \( T_a \) denote, respectively, the heat transfer coefficient, the surface temperature of the tissue, and the temperature of the air. The last term \( SW(z,T(z,t)) \) represents the sweating effect, which is neglected in this study, since the temperature elevation due to the handset is sufficiently small to activate the sweating response.

The thermal parameters used in this study are given in Table 2.3, which are the same as in [37]. They were taken from physiological textbooks. The convective heat transfer coefficient \( (h) \) between the model surface and air was set to 10.5 W/(°C.m²), which is the typical value at room temperature. The ambient temperature \( (T_a) \) and the blood temperature \( (T_b) \) were set to be 20 °C and 37 °C, respectively.

Table 2.3: The thermal parameters of the head tissues [37].

<table>
<thead>
<tr>
<th>Tissue</th>
<th>( K ) W/(K.m)</th>
<th>( C_p ) J/(K.kg)</th>
<th>( B ) W/(K.m³)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Skin</td>
<td>0.42</td>
<td>3600</td>
<td>9100</td>
</tr>
<tr>
<td>Fat</td>
<td>0.25</td>
<td>3000</td>
<td>1700</td>
</tr>
<tr>
<td>Bone</td>
<td>0.39</td>
<td>3100</td>
<td>1850</td>
</tr>
<tr>
<td>Dura</td>
<td>0.5</td>
<td>3600</td>
<td>1125</td>
</tr>
<tr>
<td>CSF</td>
<td>0.62</td>
<td>4000</td>
<td>0</td>
</tr>
<tr>
<td>Brain</td>
<td>0.535</td>
<td>3650</td>
<td>40000</td>
</tr>
</tbody>
</table>
The temperature rise due to the RF exposure from a portable telephone was obtained from the difference between the temperature $T(z,t)$ and $T(z,0)$, where $T(z,0)$ is the normal temperature distribution in the unexposed head, i.e., with SAR=0 at thermal equilibrium.

2.7.4 Discretization of the Bioheat Equation

The discretization of the bioheat equation follows that of the FDTD cell used to determine the SAR. For a continuous function of space and time $F(z,t)$, its discretized form at $m^{th}$ time step can be written as $F^m(z) = F(k\Delta z,m\Delta t)$, where $\Delta z$ is the cell size in the finite difference representation, and $\Delta t$ is the incremental time step, which should be chosen according to a condition that ensures the numerical stability. By expanding the bioheat equation in its finite difference approximation, (2.28) and (2.29) can be written as follows:

$$T^{m+1}(i) = T^m(i) + \frac{\Delta t}{C_p(i)}\text{SAR}(i) - \frac{\Delta t B(i)}{C_p(i)\rho(i)}(T^m(i) - T_b)$$

$$+ \frac{\Delta t k(i)}{C_p(i)\rho(i)\Delta z} [T^m(i+1) + T^m(i-1) - 2T^m(i)]$$

(2.30)

$$T^{m+1}(i_{\text{min}}) = \frac{K(i_{\text{min}})T^m(i_{\text{min}}+1)}{K(i_{\text{min}})+h\Delta z} + \frac{T_a h\Delta z}{K(i_{\text{min}})+h\Delta z}$$

(2.31)

2.7.5 Time Step Determination

A stability criterion similar to the time-step criterion used to calculate electric field and magnetic field using the FDTD method (which is known as Courant condition) is needed in order to make sure that the heat transfer distribution is correct. Of course, the standard FDTD time step, used in the electromagnetic field computation, is not proper for a heat transfer simulation because it is incredibly small (close to nanoseconds) and would
take a very long time to do a proper heat transfer simulation. Another reason is that heat does not move with the speed of light. Therefore, a new time step is required.

The numerical stability condition for $\Delta t$ and $\Delta z$ in solving the bioheat equation in its finite-difference form can be derived by expressing the solution using Fourier series and by checking the variation of the amplitude of each Fourier component. In order to ensure the numerical stability, $\Delta t$ was chosen to satisfy:

$$\Delta t \leq \frac{2C_p(i)\rho(i)\Delta z^2K(i)}{T^2K(i) + B(i)\Delta z^2}$$

(2.32)

which was derived from Von Neumann’s condition [17].

2.8 Numerical Results and Discussion

Matlab programs, that solve Maxwell’s curl equations, were written using the FDTD method so that the unknown total electric and magnetic fields in each layer are determined. After that, the electric fields were used to calculate the dissipated power density in each layer at three different frequencies using equation (2.26).

Figures 2.7, 2.8 and 2.9 show the computed electric field intensity in the layers at 900 MHz, 1800 MHz and 2.4 GHz, respectively. Accordingly, the dissipated power density in each layer at 900 MHz, 1800 MHz and 2.4 GHz is calculated and illustrated in Figures 2.10, 2.11 and 2.12, respectively with cell size of 0.1 mm and time step of 0.166 Pico seconds.
**Figure 2.7:** Induced field in the head model at 900 MHz for normal incidence ($E_0=1.54$).

**Figure 2.8:** Induced field in the head model at 1800 MHz for normal incidence ($E_0=1.94$).
Figure 2.9: Induced field in the head model at 2.4 GHz for normal incidence ($E_0=2.5$).

Figure 2.10: Dissipated power density in the head model at 900 MHz.
Figure 2.11: Dissipated power density in the head model at 1800 MHz.

Figure 2.12: Dissipated power density in the head model at 2.4 GHz.
SAR can be calculated in the different layers of the head by applying (2.25). In order to verify the results obtained using the FDTD method for the electric field, dissipated power and SAR distributions, the technique presented in [2] was used. In this technique, the electric and magnetic fields in each layer are expressed as the superposition of forward and backward traveling waves, each with specific unknown amplitude. Then, appropriate boundary conditions (the continuity of tangential electric and magnetic fields components) are enforced at the interfaces between the layers resulting in a set of equations to be solved for the unknown amplitudes. As the values of incident and reflected waves in each layer are found, fields in the different layers can be calculated using:

\[ E_{Total_i} = E_{incident_i} + E_{reflected_i} \]  (2.33)

SAR distributions for the three frequencies are found using the FDTD method and the method described above. Results are presented in Figures 2.13, 2.14 and 2.15. It is clear that the FDTD method gives results that are very close to the exact results using matrix inversion of boundary conditions equations. The difference between the two methods is due to that the FDTD method is not a closed form solution like the matrix inversion method. Moreover, The FDTD method has a second order accuracy.

Generally speaking, the distributions of the electric field and SAR have different shapes at different frequencies. This results from the fact that the dielectric properties are frequency dependent. In SAR distributions, it is noted that the peaks appear in skin and CSF layers. The international level for the SAR (2 W/kg) is exceeded in these two regions. This is due to the fact that these tissues have relatively high conductivity values, which means higher losses within them. This causes SAR peaks, since the higher conductivity implies a higher SAR. Table 2.4 shows the peak SAR values in skin, CSF and brain at the three frequencies of interest.
Table 2.4: Peak SAR values in skin, CSF and brain.

<table>
<thead>
<tr>
<th>Tissue</th>
<th>900 MHz</th>
<th>1800 MHz</th>
<th>2.4 GHz</th>
</tr>
</thead>
<tbody>
<tr>
<td>Skin</td>
<td>14.6</td>
<td>4.8</td>
<td>11.8</td>
</tr>
<tr>
<td>CSF</td>
<td>6</td>
<td>2.79</td>
<td>3.3</td>
</tr>
<tr>
<td>Brain</td>
<td>1.76</td>
<td>1.04</td>
<td>1.27</td>
</tr>
</tbody>
</table>

Compared with the results obtained in [6, 26], the SAR values found here are very close to the FDTD simulation results of other complicated head models. Calculated values of SAR for the head model are all well below the safety standards except in skin and CSF tissues. In the literature, the highest cell resolution used was 2 mm, which means that the actual SAR value in the tissues that have small thickness were averaged or even neglected and this is the reason behind the fact that the peak values found in this work are higher than those in [26].

Figure 2.13: SAR distribution using FDTD and matrix inversion methods at 900 MHz.
Figure 2.14: SAR distribution using FDTD and matrix inversion methods at 1800 MHz.

Figure 2.15: SAR distribution using FDTD and matrix inversion methods at 2.4 GHz.

It is informative to consider how quickly the temperature in the head is elevated because the steady state temperature rise may not give a realistic picture of the
temperature rise distribution within the head since a telephone call is not so long to reach steady state in most calls. Figures 2.16, 2.17 and 2.18 show the short-term peak temperature rise in the skin, fat, bone and brain tissues at the different considered frequencies.

It is found that the peak temperature increases exponentially over the first 7–8 minutes, then the rate of temperature rise slows down. The steady state is reached after about 25 minutes of exposure. These results agree with what was found in [17] and [22]. The peak temperature rise in the head occurs in the skull and it is up to 0.227°C at 900 MHz, 0.268°C at 1800 MHz, and 0.385°C at 2.4 GHz.

![Diagram showing the short-term peak temperature rise at 900 MHz.

**Figure 2.16:** Short-term peak temperature rise at 900 MHz.
Figure 2.17: Short-term peak temperature rise at 1800 MHz.

Figure 2.18: Short-term peak temperature at 2.4 GHz.
The steady-state temperature computation provides information on the maximum temperature rise within the human head exposed to the RF fields from wireless applications. Figures 2.19, 2.20 and 2.21 show the temperature-rise distribution in the steady state (after fifty minutes of exposure). It should be pointed out that the peak SAR occurs at the skin tissue, while the peak temperature rise occurs within the internal bone tissue rather than skin tissue. This is due to the fact that the temperature distribution is affected by the environment, imposed boundary conditions, thermal conductivity and blood perfusion rate inside the living tissues. Moreover, bone has a low blood perfusion rate along with a high thermal conductivity (as shown in Table 2.3). This means that the dissipated power by the skull is not lost by the blood flow like the skin.

Also, it is found that with an antenna output power of 0.6 W, the maximum temperature rise in the brain is up to 0.13°C at 900 MHz, 0.097°C at 1800 MHz, and 0.14°C at 2.4 GHz. Because the brain has the largest blood-flow rate, the temperature rise in it slows down rapidly. However, since the normal active heat transfer is very effective in regulating temperature in the brain, a temperature rise of up to 3.5°C in the brain is harmless and does not cause any physiological damage [17]. This value is approximately 27 times the computed maximum temperature rise at 900 MHz, 36 times the computed maximum temperature rise at 1800 MHz and 25 times the computed maximum temperature rise at 2.4 GHz. From the obtained results, it is found that temperature elevations are not directly proportional to the local SAR values.
Figure 2.19: The steady state temperature-rise distribution at 900 MHz.

Figure 2.20: The steady state temperature-rise distribution at 1800 MHz.
2.9 Effect of Using Mobile Phones in Enclosed Areas

With the rapid increase in the use of mobile phones in enclosed environments, such as elevators, public anxiety over the possibility of RF exposures in these environments exceeding the international restrictions has been growing. The arguments are based on the well known fact that conductive structures tend to reflect electromagnetic fields that may result in an increase of SAR and temperature elevation.

Temperature elevation is the most important parameter in reaching any conclusion regarding any added effect of using cellular phones inside enclosed environments. For this purpose, the multilayered human head model is adopted as shown in Figure 2.22. The distance between the wall and the head model ($d$) is varied.

Firstly; we investigate the effect of using a mobile phone inside an elevator. In this case, an infinite metallic wall is simulated since the elevator wall is usually made of iron with conductivity $10^7$ S/m. Figures 2.23 and 2.24 show the temperature elevation.
distributions for the head model in free-space (without walls) and with a metallic wall existing behind the plane wave source.

**Figure 2.22:** A plane wave normally incident on the interface of a human head model with a wall behind the source.

**Figure 2.23:** Effect of having an iron wall behind the plane wave source on the steady-state temperature elevation in the head. The head model is set at \( \frac{\lambda}{4} \) and \( \frac{\lambda}{2} \) away from the wall at 900 MHz.
Secondly; the effect of having a concrete wall beside the mobile phone is investigated. Concrete is essentially a non-magnetic material with a relative permeability of unity. The principal factor influencing the concrete permittivity is the amount of moisture contained within the concrete pores, which often varies with increasing depth below the surface. The relative permittivity of concrete used in the simulation is 6 for naturally dry concrete and 12 for saturated concrete [38]. Conductivity of concrete varies with water content as well. The used conductivity of concrete is 10 mS/m for naturally dry concrete and 20 mS/m for saturated concrete [39]. Moreover, Figures 2.25 and 2.26 show temperature elevation distributions for the head model with a concrete wall existing
behind the plane wave source. In these figures, the head model is set at \( \frac{\lambda}{4} \) and \( \frac{\lambda}{2} \) away from the wall, for 900 MHz and 1800 MHz.

The difference between the temperature elevation distribution while the head is in free space and beside the conductive walls can be easily observed in these figures. It is very clear that the effect of these walls depends mainly on the distance between the wall and the head model. Furthermore, it is found that the peaks of the excitation source occur at the distance from the metallic wall given approximately by the equation

\[
d_{z}^{peak} = (2n + 1)\frac{\lambda}{4}
\]

(2.34)

where \( n = 0, 1, 2 \ldots \) and \( \lambda \) is the free-space wavelength at a given frequency.

**Figure 2.25:** Effect of having a concrete wall behind the plane wave source on the temperature elevation in the head. The head model is set at \( \frac{\lambda}{4} \) and \( \frac{\lambda}{2} \) away from the wall at 900 MHz.
Figure 2.26: Effect of having a concrete wall behind the plane wave source on the temperature elevation in the head. The head model is set at $\frac{\lambda}{4}$ and $\frac{\lambda}{2}$ away from the wall at 1800 MHz.

In the case of metallic walls, the maximum temperature elevation obtained at 900 and 1800 MHz are 0.37°C and 0.3°C, respectively, corresponding to an increase of 64% and 15.3% over those of the human model in free space. It has been also found that for maximum SAR values, there was an increase of 67% and 40% over those found in free space at 900 and 1800 MHz. Our SAR results are in good agreement with the measured ones at 900 MHz presented in [40].

For the case of wet concrete walls, the maximum temperature elevation obtained at 900 and 1800 MHz were larger than those found in free space case by 26.4% and 3.8%, respectively. It has been also found that for maximum SAR values, there was an increase of 56.5% and 27% over those found in free space at 900 and 1800 MHz. For the case of dry concrete wall, SAR values and temperature elevation are found to be a little bit smaller.
than the wet concrete walls results. This is due to the fact that the conductivity of dry concrete is smaller than that for wet concrete.

Another interesting fact is that, at some distances from the wall, there is a destructive interference between the EM fields due to multi-reflections in the enclosed environments. This causes the electric field to be almost zero at some definite distances from the walls. At these distances, the temperature elevation in the head is lower than the free space case. The minimum values of the peak SAR, as well as temperature elevation, are obtained when the distance is given approximately by the following equation:

\[ d_{2}^{\text{min}} = n \frac{\lambda}{2} \] (2.35)

The SAR and temperature elevation distributions were found to depend on the head model position relative to the wall. For the other type of enclosed environments, concrete walls, a substantial enhancement due to multi-reflections was observed. These finding are true regardless of the type of the human model used. Moreover, their effect is smaller than that observed in an elevator because concrete conductivity is smaller than that of iron. This would result in less reflections from the walls.

2.10 Effect of Distance between the Head Model and the Source

In all of the above results, the power radiated from cell phones was set to be 0.6 Watt. Figure 2.27 illustrates the peak SAR when the distance between the head model and the source is variable at different frequencies. Moreover, Figure 2.28 illustrates the peak temperature elevation in the head model when the distance between the head model and the source is variable at different frequencies. It is obvious from these two figures that there is an inversely proportional relationship between these distributions and the distance between the head model and the excitation source. This observation agrees with that found in [5, 41]. For instance, increasing the distance between the source and the head model
from 1 cm to 2 cm causes the SAR to decrease by almost 75% at the frequencies of interest. Moreover, it causes the temperature elevation to decrease by 74% at 900 MHz, 75% at 1800 MHz and 70% at 2.4 GHz.

Figure 2.27: Peak SAR when the distance between head model and the source is variable.

Figure 2.28: Peak temperature elevation at different distances between head model and the source.
2.11 Effect of Age on SAR Distribution

The level and distribution of radio frequency energy absorbed in a child’s head during the use of a mobile phone has been a controversial issue in recent years. When investigating this problem, the dielectric properties of biological tissues for adults are so far being used due to the lack of the ones of children which raised the public concern on whether or not the children heads absorb more electromagnetic energy or allow deeper electromagnetic penetration than adult ones.

In this section, the age effect of dielectric properties on the spatial peak SAR will be examined by employing a 7-year-old and a 3-year-old head’s tissues properties, which are presented in Table 2.5 at a frequency of 900 MHz. They are taken from [42] in which an empirical formula was derived according to Lichtenecker’s exponential law, for the complex permittivity of various tissues as a function of the hydrated rate or the Total Body Water (TBW). The children’s tissues properties were found by means of the relationship between the TBW and the age.

Table 2.5: Dielectric properties of children and adult at 900 MHz [42].

<table>
<thead>
<tr>
<th>Tissues</th>
<th>Adult</th>
<th>3 years old</th>
<th>7 years old</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Relative permittivity</td>
<td>Conductivity (S/m)</td>
<td>Relative permittivity</td>
</tr>
<tr>
<td>Skin</td>
<td>41.4</td>
<td>0.87</td>
<td>43.48</td>
</tr>
<tr>
<td>Fat</td>
<td>11.33</td>
<td>0.11</td>
<td>13.26</td>
</tr>
<tr>
<td>Bone</td>
<td>11.27</td>
<td>0.23</td>
<td>23.12</td>
</tr>
<tr>
<td>Dura</td>
<td>44.43</td>
<td>0.96</td>
<td>46.38</td>
</tr>
<tr>
<td>CSF</td>
<td>68.64</td>
<td>2.41</td>
<td>69.1</td>
</tr>
<tr>
<td>Brain</td>
<td>45.8</td>
<td>0.765</td>
<td>47.65</td>
</tr>
</tbody>
</table>
Figure 2.29 shows the short-term peak temperature rise in the brain tissue for an adult, 3 years and 7 years old head models. The results show a very small difference in temperature elevation due to age. For children, temperature increase is less than adults’ temperature increase value by approximately 2%. This means that the change in dielectric properties due to age does not significantly affect the temperature elevation. This could be explained as a cancellation of the increased conductivity and decreased electric field penetrating into the tissue. This is due to the fact that the conductivity and permittivity increase by the same amount.

![Figure 2.29](image-url)

**Figure 2.29:** Short-term peak temperature rise in the brain tissue for an adult, 3 years and 7 years old head models at 900 MHz.

Another issue of concern regarding the effect of wireless applications on children is their small head size and whether this fact has an effect on SAR and temperature elevation in the brain or not. To investigate this, a 5-year old child head model is obtained by linear scaling of the adult head model using a scaling factor of 0.693 in the horizontal
plane [43]. Figure 2.30 shows the SAR and steady-state temperature distributions for the 5-year old child head model at 900 MHz.

**Figure 2.30**: SAR and temperature distributions for 5-years old head size at 900 MHz.

The results show that the SAR values increase in the CSF and brain tissues. Moreover, SAR values exceed the international limit level at the beginning of the brain tissue. But, compared to the results of an adult head, the steady-state temperature elevation (as well as its peak value) decreases in the whole head model. The peak of the temperature increase in an adult head model is 0.227°C (Figure 2.19), while it is 0.143°C in the head model of a 5-years old child.
Chapter Three: Oblique Incidence

3.1 Introduction

In this chapter, the electric field intensity, SAR and temperature elevation distributions in the multilayered model of human head irradiated by oblique incidence plane wave are evaluated. The SAR distribution in all layers will be calculated. Moreover, the steady state thermal elevation distribution will be found by solving the BHE equation using the FDTD method as described in Chapter 2. Obliquely incident plane wave can be represented by a linear combination of two special cases, parallel polarization and perpendicular polarization. Both polarization types will be discussed in this chapter. It is worth mentioning that the calculation of the electric field for oblique incidence is based on field matching at the boundaries of the layers [2].

3.2 Oblique Incidence; Perpendicular Polarization (TE Polarization)

The used head model is shown in Figure 3.1 where a uniform plane wave is obliquely incident onto the interface between the air and the skin layer. The plane of incidence containing the propagation vector and the normal to the interface is the x-z plane. The electric field of the incident wave is perpendicular on the plane of incidence. Because of that, this wave orientation is defined as perpendicular polarization (or TE Polarization). $\theta_i$ is the angle of incidence, which is the angle between the direction of propagation and a line that is normal to the surface (the z axis in this case). The reflection angle is shown as $\theta_r$ at which the reflected wave will propagate away from the interface.
Assuming that the $x$ coordinate is set to zero, for simplicity, the forward traveling electric fields in the different layers of the head model follow the following expression:

$$E_{i_i} = E_{i_f} \hat{a}_y e^{-\gamma_i x \cos \theta_{i_i}}$$  \hspace{1cm} (3.1)

Similarly, the backward traveling electric field in the $i^{th}$ layer of the head model is:

$$E_{r_i} = E_{b_i} \hat{a}_y e^{\gamma_i x \cos \theta_{r_i}}$$  \hspace{1cm} (3.2)

where $E_{i_f}$ and $E_{b_i}$ are the amplitudes of the incident and reflected fields in the $i^{th}$ layer, respectively, $\theta_{i_i}$ is the angle of incidence in the $i^{th}$ layer, and $\theta_{r_i}$ is the angle of reflection in the $i^{th}$ layer. The parameters $\gamma_i$ and $\eta_i$ are the propagation constant and the intrinsic impedance of the $i^{th}$ layer, respectively ($i=0, 1, \ldots, 6$), and are given by [44]:

$$\gamma_i = \sqrt{j \omega \mu_i (\sigma_i + j \omega \varepsilon_i)}$$  \hspace{1cm} (3.3)

$$\eta_i = \frac{j \omega \mu_i}{\sqrt{\sigma_i + j \omega \varepsilon_i}}$$  \hspace{1cm} (3.4)

where $\sigma_i$ and $\varepsilon_i$ are the conductivity and permittivity of the $i^{th}$ layer, respectively.

In a similar manner, the forward traveling magnetic fields in the different layers of the head model follow the following expression:

$$H_{i_i} = \frac{E_{i_f}}{\eta_i} (-\cos \theta_{i_i} \hat{a}_x + \sin \theta_{i_i} \hat{a}_z) e^{-\gamma_i x \cos \theta_{i_i}}$$  \hspace{1cm} (3.5)
Similarly, the backward traveling magnetic field in the $i^{th}$ layer of the head model is:

$$H_{ri} = \frac{E_{bi}}{\eta_i} (\cos \theta_{ri} \hat{a}_x + \sin \theta_{ri} \hat{a}_z) e^{y_i z \cos \theta_{ri}}$$  \hspace{1cm} (3.6)

In order to find the angles of incidence and reflections at each interface, Snell’s law of refraction must be applied along with the fact of $\theta_i = \theta_r$. This gives

$$\sin \theta_{ri} = \frac{\gamma_o}{\gamma_i} \sin \theta_{ro}$$  \hspace{1cm} (3.7)

where $\theta_{ri}$ and $\gamma_o$ are the angle of incidence and the propagation constant in the first layer (air), respectively. Using the following relationship:

$$\cos \theta = \sqrt{1 - \sin^2 \theta}$$  \hspace{1cm} (3.8)

Equation (3.7) can also be written as follows:

$$\cos \theta_{ri} = \sqrt{1 - \left(\frac{\gamma_o}{\gamma_i} \sin \theta_{ro}\right)^2}$$  \hspace{1cm} (3.9)

The unknown incident and reflected field constants can be found by applying the boundary conditions at the interfaces. The tangential components of $E$ and $H$ ($E_y$ and $H_x$) at the interface $z = d_i$ are continuous. $E_y$ at the left of interface $z = d_i$ is equal to that at the right of interface $d_i$ which can be written in the following equation:

$$E_{fi} e^{-\gamma_i d_i \cos \theta_{ri}} + E_{bi} e^{\gamma_i d_i \cos \theta_{ri}} =$$

$$E_{fi+1} e^{-\gamma_{i+1} d_i \cos \theta_{ri+1}} + E_{bi+1} e^{\gamma_{i+1} d_i \cos \theta_{ri+1}}$$  \hspace{1cm} (3.10)

Moreover, the $x$-component of the magnetic field should be continuous across the interfaces and this equality is given by the following expression:

$$\frac{-\cos \theta_{ri}}{\eta_i} (E_{fi} e^{-\gamma_i d_i \cos \theta_{ri}} - E_{bi} e^{\gamma_i d_i \cos \theta_{ri}}) =$$

$$\frac{-\cos \theta_{ri+1}}{\eta_{i+1}} (E_{fi+1} e^{-\gamma_{i+1} d_i \cos \theta_{ri+1}} - E_{bi+1} e^{\gamma_{i+1} d_i \cos \theta_{ri+1}})$$  \hspace{1cm} (3.11)
Applying the boundary conditions at \( z = d_o, d_1, ..., d_5 \) leads to a set of twelve equations [2]. Solving this set of equations yields the values of the incident and reflected electric fields at the interfaces. The total electric field in the \( i^{th} \) layer can be calculated using the following equation:

\[
E_{t_i} = E_{i_i} + E_{r_i}
\]  
(3.12)

### 3.3 Oblique Incidence; Parallel Polarization (TM Polarization)

In this section, we will study the TM polarization where the electric field of the incident wave lies in the plane of incidence such that this wave orientation is defined as parallel polarization. Figure 3.2 shows the head model with the parallel polarized plane wave. Assuming that the \( x \) coordinate is set to zero, for simplicity, the forward traveling electric fields in the different layers of the head model follow the following expression:

\[
E_{t_i} = E_{f_i}(\cos \theta_i \hat{a}_x - \sin \theta_i \hat{a}_z)e^{-\gamma_i z \cos \theta_i}
\]  
(3.13)

Similarly, the backward traveling electric field in the \( i^{th} \) layer of the head model is:

\[
E_{r_i} = E_{b_i}(\cos \theta_i \hat{a}_x + \sin \theta_i \hat{a}_z)e^{\gamma_i z \cos \theta_i}
\]  
(3.14)

where \( E_{f_i} \) and \( E_{b_i} \) are the amplitudes of the incident and reflected fields in the \( i^{th} \) layer, respectively, and \( \theta_i \) is the angle of incidence in the \( i^{th} \) layer. The forward traveling magnetic fields in the different layers of the head model follow the following expression:

\[
H_{t_i} = \begin{pmatrix} E_{f_i} e^{-\gamma_i z \cos \theta_i} \end{pmatrix} \hat{a}_y
\]  
(3.15)

Similarly, the backward traveling magnetic field in the \( i^{th} \) layer of the head model is:

\[
H_{r_i} = -\begin{pmatrix} E_{b_i} e^{\gamma_i z \cos \theta_i} \end{pmatrix} \hat{a}_y
\]  
(3.16)
Figure 3.2: Oblique incidence of plane wave on the head model (parallel polarization)

The unknown incident and reflected field constants can be found by applying the boundary conditions at the interfaces. The tangential components of $\mathbf{E}$ and $\mathbf{H}$ ($E_x$ and $H_y$) at the interface $z = d_i$ are continuous. $E_x$ at the left of interface $z = d_i$ is equal to that at the right of interface $d_i$ which can be written in the following equation:

\[
(E_f \, e^{-\gamma_i d_i \cos \theta_{r_i}} + E_b \, e^{\gamma_i d_i \cos \theta_{r_i}}) \cos \theta_{i+1} = \\
(E_f^{i+1} \, e^{-\gamma_{i+1} d_i \cos \theta_{r_{i+1}}} + E_b^{i+1} \, e^{\gamma_{i+1} d_i \cos \theta_{r_{i+1}}}) \cos \theta_{i+1}
\]  

Moreover, the $y$-component of the magnetic field should be continuous across the interfaces and this equality is given by the following expression:

\[
\frac{1}{\eta_i} (E_f \, e^{-\gamma_i d_i \cos \theta_{r_i}} - E_b \, e^{\gamma_i d_i \cos \theta_{r_i}}) = \\
\frac{1}{\eta_{i+1}} (E_f^{i+1} \, e^{-\gamma_{i+1} d_i \cos \theta_{r_{i+1}}} - E_b^{i+1} \, e^{\gamma_{i+1} d_i \cos \theta_{r_{i+1}}})
\]  

Applying the boundary conditions at $z = d_0, d_1, ..., d_5$ leads to a set of twelve equations. Solving this set of equations yields the values of incident and reflected electric fields at the interfaces.
3.4 Results

The total electric field intensity has been calculated in each layer of the multi-layered model at different incidence angles (0°, 30°, 60°) and is shown in Figures 3.3 and 3.4 at frequencies of 900 MHz, 1800 MHz and 2.4 GHz for perpendicular and parallel polarizations, respectively. These figures show that for the same incidence angle, the electric field inside the human head in the case of parallel polarization is larger than that in the case of perpendicular polarization. This may be due to a higher field reflection by the skin layer in the perpendicular polarization case as compared to the parallel polarization case [2].

It is worth mentioning that the electric field in perpendicular polarization has one component only which is tangential to the interface. Thus, the electric field must be continuous across the interfaces between different layers. On the other hand, for parallel polarization, the electric field has two components; one of these two components is tangential to the interface while the other is not. So, the total electric field is not continuous across the interfaces for the parallel polarization case.
Figure 3.3: Induced total electric field intensity in the head model for perpendicular polarization ($E_{fo} = 300 \text{ V/m}$) at (a) 900 MHz, (b) 1800 MHz and (c) 2.4 GHz.
Figure 3.4: Induced total electric field intensity in the head model for parallel polarization ($E_{fo} = 300 \text{ V/m}$) at (a) 900 MHz, (b) 1800 MHz and (c) 2.4 GHz.
Once the field distribution is known, the SAR distribution in the different layers of the head model can be calculated. Figures 3.5 and 3.6 represent the SAR distributions for different angles of incidence at 900 MHz, 1800 MHz and 2.4 GHz for perpendicular and parallel polarizations, respectively. SAR distribution has its own shape at each frequency which is due to the fact that the electric properties of living tissues depend mainly on the frequency of the wave. Moreover, as the angle of incidence increases, the SAR values decrease. It is noted that the SAR peaks occur in the skin and CSF layers. The international accepted level for the SAR is 2 W/Kg. For perpendicular polarization, this value is exceeded in the CSF layer for the 0° (normal incidence) and 30° incidence angles, while it is exceeded in the skin layer for the three angles of incidence. Moreover, for parallel polarization, this value is exceeded in both the skin and CSF layers for the three angles of incidence as shown in Figure 3.6. It has been found that for perpendicular polarization, at 45° incidence angle, the SAR peak occurring in the CSF tissue will be less than the international limit. However, the SAR peak occurring in the skin tissue becomes less than the international limit at 71.5° angle of incidence. On the other hand, for parallel polarization, the peak of the SAR occurring in the CSF tissue becomes less than the international limit at an incidence angle of 65°, while the SAR peak occurring in the skin becomes less than the international limit at an angle of 78°.

Compared with the results obtained in [6] and [26], the obtained SAR values are very close to the FDTD simulation of other complicated head models. In the literature, the cell resolutions used are inadequate to find the actual SAR value in very thin tissues and it is averaged or even neglected. This is the reason behind the fact that the obtained peak values in this work are higher than those found in [26].

The steady-state temperature computation provides information on the maximum temperature rise within the human head exposed to the RF fields from wireless
applications. Figures 3.7 and 3.8 show the temperature-rise distribution at the different frequencies of interest and different angles of incidence for the perpendicular and parallel polarizations, respectively. It should be pointed out that even though the peak SAR occurs at the skin tissue, the peak temperature rise occurs within the internal bone tissue rather than the skin tissue.
Figure 3.5: SAR distribution in the head model for perpendicular polarization at (a) 900 MHz, (b) 1800 MHz and (c) 2.4 GHz.
Figure 3.6: SAR distribution in the head model for parallel polarization at (a) 900 MHz, (b) 1800 MHz and (c) 2.4 GHz.

It is found that the peak temperature rise increases exponentially over the first 9-10 minutes, then the rate of temperature rise slows down. The steady state is reached after about 25 minutes of exposure. Our results agree with those presented in [17] and [22].

It is clear from Figures 3.7 and 3.8 that as the angle of incidence increases, the temperature elevation decreases. In order to compare the temperature elevation at the different frequencies, Tables 3.1 and 3.2 show the peak temperature elevation at different frequencies and angles of incidence at the beginning of the brain tissue and in the skull for perpendicular and parallel polarizations, respectively.
**Table 3.1:** Peak temperature elevation at different frequencies and incidence angles (perpendicular polarization).

<table>
<thead>
<tr>
<th>Angle of incidence</th>
<th>Skull Tissue</th>
<th>Brain Tissue</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>900 MHz</td>
<td>1800 MHz</td>
</tr>
<tr>
<td>0°</td>
<td>0.227</td>
<td>0.268</td>
</tr>
<tr>
<td>30°</td>
<td>0.186</td>
<td>0.21</td>
</tr>
<tr>
<td>60°</td>
<td>0.082</td>
<td>0.087</td>
</tr>
</tbody>
</table>

**Table 3.2:** Peak temperature elevation at different frequencies and incidence angles (parallel polarization).

<table>
<thead>
<tr>
<th>Angle of incidence</th>
<th>Skull Tissue</th>
<th>Brain Tissue</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>900 MHz</td>
<td>1800 MHz</td>
</tr>
<tr>
<td>0°</td>
<td>0.227</td>
<td>0.268</td>
</tr>
<tr>
<td>30°</td>
<td>0.20</td>
<td>0.253</td>
</tr>
<tr>
<td>60°</td>
<td>0.13</td>
<td>0.195</td>
</tr>
</tbody>
</table>
Figure 3.7: The steady state temperature-rise distribution model for perpendicular polarization at (a) 900 MHz, (b) 1800 MHz and (c) 2.4 GHz.
Figure 3.8: The steady state temperature-rise distribution model for parallel polarization at (a) 900 MHz, (b) 1800 MHz and (c) 2.4 GHz.
The highest temperature elevation is 0.37°C which occurs in the skull at 2.4 GHz with a zero angle of incidence (i.e., normal incidence case). Moreover, it is known that the brain has the largest blood-flow rate in the head, so that temperature rise slows down rapidly. However, since the normal active heat transfer is very effective in regulating the temperature in the brain, a temperature rise up to 3.5°C in the brain is harmless and does not cause any physiological damage [17]. Figure 3.9 compares the peak temperature elevation in the brain tissue versus the angle of incidence for both polarizations at 2.4 GHz.

![Figure 3.9: Peak temperature rise in the brain tissue at different angles of incidence at 2.4 GHz for both polarizations.](image)

It is clear that as the angle of incidence increases, temperature elevation decreases faster in the case of perpendicular polarization as compared to the case of parallel polarization. This may be due to the fact that (at a specific incidence angle) a higher wave reflection occurs in the perpendicular polarization case as compared to the parallel polarization case.
Chapter Four: Effect of Metallic Implants Inside Human Head

4.1 Introduction

In order to obtain an overview of metallic implants’ effects and a direct understanding of the consequences of using the cellular phones by people who have a metallic implant inside their heads, we adopt a simple model of a homogeneous sphere that has the same properties as the brain with radius of 4 cm for the human head. An aluminum implant of 10 mm thickness, 8 mm height and 3 cm length is set at distance of 5 mm inside the model as shown in Figure 4.1. This model is excited by an obliquely incident plane wave. Moreover, this model will be used to verify the results obtained using the planar multi-layered head model. To excite the model by an obliquely incident plane wave, scattered field formulation [45, 46] will be used. Also, temperature elevation will be calculated by solving the BHE equation in three dimensions with appropriate boundary conditions at the interface with the ambient environment. It is found that the SAR values increase around the metallic implant, but the change in the peak values and the temperature elevation is small. On the other hand, the results obtained are very close to what was found in previous chapters but the consumed excitation time is very high compared to the time needed when using the multi-layered planar head model. The effect of other tissues is taken in consideration by putting the electric field intensity of the source equal to that found at the start of the brain layer in the multi-layered head model.
4.2 Yee Cell Definition [14, 19]

Yee originated a set of finite-difference equations for the time-dependent Maxwell’s curl equations system. These equations can be represented in discrete form, both in space and time, employing the second-order accurate central difference formula. As mentioned before, the electric and magnetic field components are sampled at discrete positions both in time and space. The FDTD technique divides the three-dimensional problem geometry into cells to form a grid. A unit cell of this grid is called a Yee cell. Using rectangular Yee cells, a stepped or “staircase” approximation of the surface and internal geometry of the structure of interest is made with a space resolution set by the size of the unit cell.

The discrete spatial positions of the field components have a specific arrangement in the Yee cell, as demonstrated in Figure 4.2. The \( \mathbf{E} \) vector components are placed at the centers of the edges of the Yee cells and oriented parallel to the respective edges, and the
\( \mathbf{H} \) vector components are placed at the centers of the faces of the Yee cells and are oriented normal to the respective faces.

![Diagram of Yee cell with field components](image)

**Figure 4.2:** Arrangement of field components on a Yee cell indexed as \((i, j, k)\) [14].

Yee cell has dimension \(\Delta x\) in the \(x\) direction, \(\Delta y\) in the \(y\) direction, and \(\Delta z\) in the \(z\) direction. It can be easily noticed in Figure 4.2 that each magnetic field component is surrounded by four electric field components that are curling around the magnetic field component. Figure 4.3 gives a clearer view for the components surrounding \(\mathbf{H}_x(i, j, k)\). Similarly, as shown in Figure 4.4 for \(\mathbf{E}_x(i, j, k)\), each electric field component is surrounded by four magnetic field components that are curling around the electric field component.

The FDTD algorithm samples and calculates the fields at discrete time steps. However, as mentioned in Chapter 2, the electric and magnetic field components are not sampled at the same time instants. Interleaving must be used in order to make leap-frog
scheme as was shown in Figure 2.4. Therefore, the electric field components are calculated at integer time steps, and magnetic field components are calculated at half-integer time steps, and they are offset from each other by $\frac{\Delta t}{2}$.

![Diagram of field components](image)

**Figure 4.3:** Field components around $H_x (i, j, k)$ [14].

![Diagram of field components](image)

**Figure 4.4:** Field components around $E_x (i, j, k)$ [14].

### 4.3 Scattered Field Formulation [14, 45, 46]

Sources are the necessary components of an FDTD simulation, and their types vary depending on the type of the problem under consideration. In order to integrate a plane
wave source to excite the head model, scattered field formulation is used. Far-zone sources, such as plane waves, are the fields generated somewhere outside the FDTD problem space that illuminate the objects in the problem space. The incident field always propagates in free space (even when passing through the interaction object) and is defined as the field that would be present in the absence of the object. Therefore, they are the fields that can be described by analytical expressions and would satisfy Maxwell’s curl equations where the problem space medium is free space, such that

\[ \nabla \times \mathbf{H}_{\text{inc}} = \varepsilon_0 \frac{\partial \mathbf{E}_{\text{inc}}}{\partial t} \]  
(4.1)

\[ \nabla \times \mathbf{E}_{\text{inc}} = -\mu_0 \frac{\partial \mathbf{H}_{\text{inc}}}{\partial t} \]  
(4.2)

We will apply a plane wave as the excitation source for the head model. When the incident field illuminates the head model in an FDTD problem space, scattered fields are generated and thus have to be calculated. The scattered field formulation is one of the simplest techniques that can be used for calculating scattered fields. The total fields satisfy Maxwell’s curl equations for a general medium, such that:

\[ \nabla \times \mathbf{H}_{\text{total}} = \sigma \mathbf{E}_{\text{total}} + \varepsilon \frac{\partial \mathbf{E}_{\text{total}}}{\partial t} \]  
(4.3)

\[ \nabla \times \mathbf{E}_{\text{total}} = -\mu \frac{\partial \mathbf{H}_{\text{total}}}{\partial t} \]  
(4.4)

Then, the scattered fields are defined as the difference between the total fields and incident fields. Therefore, one can write

\[ \mathbf{E}_{\text{total}} = \mathbf{E}_{\text{inc}} + \mathbf{E}_{\text{scat}} \]  
(4.5)

\[ \mathbf{H}_{\text{total}} = \mathbf{H}_{\text{inc}} + \mathbf{H}_{\text{scat}} \]  
(4.6)

Using equations (4.5) and (4.6), equations (4.3) and (4.4) can be rewritten in terms of incident and scattered field terms as
\[ \nabla \times \mathbf{H}_{\text{inc}} + \nabla \times \mathbf{H}_{\text{scat}} = \sigma \mathbf{E}_{\text{inc}} + \sigma \mathbf{E}_{\text{scat}} + \varepsilon \frac{\partial \mathbf{E}_{\text{inc}}}{\partial t} + \varepsilon \frac{\partial \mathbf{E}_{\text{scat}}}{\partial t} \] (4.7)

\[ \nabla \times \mathbf{E}_{\text{inc}} + \nabla \times \mathbf{E}_{\text{scat}} = -\mu \frac{\partial \mathbf{H}_{\text{inc}}}{\partial t} - \mu \frac{\partial \mathbf{H}_{\text{scat}}}{\partial t} \] (4.8)

The curls of the incident fields can be replaced by the time-derivative terms from (4.1) and (4.2). After replacing and rearranging the terms in (4.7) and (4.8) the obtained equations are:

\[ \varepsilon \frac{\partial \mathbf{E}_{\text{scat}}}{\partial t} + \sigma \mathbf{E}_{\text{scat}} = \nabla \times \mathbf{H}_{\text{scat}} + (\varepsilon_0 - \varepsilon) \frac{\partial \mathbf{E}_{\text{inc}}}{\partial t} - \sigma \mathbf{E}_{\text{inc}} \] (4.9)

\[ \mu \frac{\partial \mathbf{H}_{\text{scat}}}{\partial t} = -\nabla \times \mathbf{E}_{\text{scat}} + (\mu_0 - \mu) \frac{\partial \mathbf{H}_{\text{inc}}}{\partial t} \] (4.10)

At this point, the derivatives can be represented by central finite difference approximations, and updating equations can be obtained for the scattered field formulation. For the \( x \)-component, the updating equation of electric field can be expressed as:

\[
\varepsilon (i, j, k) \frac{E_{x,\text{scat}}^{n+1}(i,j,k)-E_{x,\text{scat}}^{n}(i,j,k)}{\Delta t} + \sigma (i,j,k) \frac{E_{x,\text{scat}}^{n+1}(i,j,k)-E_{x,\text{scat}}^{n}(i,j,k)}{2} =
\]

\[
\frac{H_{x,\text{scat}}^{n+1/2}(i,j,k)-H_{x,\text{scat}}^{n+1/2}(i,j-1,k)}{\Delta y} + \frac{H_{y,\text{scat}}^{n+1/2}(i,j,k)-H_{y,\text{scat}}^{n+1/2}(i,j,k-1)}{\Delta z} +
\]

\[
(\varepsilon_0 - \varepsilon (i,j,k)) \frac{E_{x,\text{inc}}^{n+1}(i,j,k)-E_{x,\text{inc}}^{n}(i,j,k)}{\Delta t} - \sigma (i,j,k) \frac{E_{x,\text{inc}}^{n+1}(i,j,k)-E_{x,\text{inc}}^{n}(i,j,k)}{2} \] (4.11)

Equation (4.11) can be arranged such that the new value of the scattered electric field component \( E_{x,\text{scat}}^{n+1} \) is calculated using the other terms as:

\[
E_{x,\text{scat}}^{n+1}(i,j,k) = C_{exe}(i,j,k)E_{x,\text{scat}}^{n}(i,j,k) + C_{exh}(i,j,k) \left[ H_{x,\text{scat}}^{n+1/2}(i,j,k) - 
\right. \\
H_{x,\text{scat}}^{n+1/2}(i,j-1,k) \right] + C_{exh}(i,j,k) \left[ H_{y,\text{scat}}^{n+1/2}(i,j,k) - 
\right. \\
H_{y,\text{scat}}^{n+1/2}(i,j,k-1) \right] +
\]

\[
C_{exe}(i,j,k)E_{x,\text{inc}}^{n+1}(i,j,k) + C_{xeip}(i,j,k)E_{x,\text{inc}}^{n}(i,j,k) \] (4.12)
where

\[
C_{\text{exe}}(i, j, k) = \frac{2 \varepsilon (i, j, k) - \sigma(i, j, k)\Delta t}{2 \varepsilon (i, j, k) + \sigma(i, j, k)\Delta t}
\]

\[
C_{\text{ehx}}(i, j, k) = \frac{2\Delta t}{\Delta y(2 \varepsilon (i, j, k) + \sigma(i, j, k)\Delta t)}
\]

\[
C_{\text{ehy}}(i, j, k) = -\frac{2\Delta t}{\Delta z(2 \varepsilon (i, j, k) + \sigma(i, j, k)\Delta t)}
\]

\[
C_{\text{exeic}}(i, j, k) = \frac{2 (\varepsilon_0 - \varepsilon(i, j, k)) + \sigma(i, j, k)\Delta t}{2 \varepsilon (i, j, k) + \sigma(i, j, k)\Delta t}
\]

\[
C_{\text{exeip}}(i, j, k) = -\frac{2 (\varepsilon_0 - \varepsilon(i, j, k)) + \sigma(i, j, k)\Delta t}{2 \varepsilon (i, j, k) + \sigma(i, j, k)\Delta t}
\]

In this equation the, \textit{ic} term in the subscript of the coefficient \(C_{\text{exeic}}\) indicates that this coefficient is multiplied with the \textit{current} value of the \textit{incident} field component, whereas the \textit{ip} term in the subscript of the coefficient \(C_{\text{exeip}}\) indicates that this coefficient is multiplied with the \textit{previous} value of the \textit{incident} field component. Similarly, the updating equation can be written for \(E_{y,\text{scat}}^{n+1}(i, j, k)\) as:

\[
E_{y,\text{scat}}^{n+1}(i, j, k) = C_{\text{eye}}(i, j, k)E_{y,\text{scat}}^n(i, j, k) + C_{\text{eyhx}}(i, j, k)\left[H_{x,\text{scat}}^{n+\frac{1}{2}}(i, j, k) - H_{x,\text{scat}}^{n+\frac{1}{2}}(i, j, k-1)\right] + C_{\text{eye}}(i, j, k)\left[H_{y,\text{scat}}^{n+\frac{1}{2}}(i, j, k) - H_{y,\text{scat}}^{n+\frac{1}{2}}(i, j, k-1)\right] + C_{\text{eyic}}(i, j, k)E_{y,\text{inc}}^{n+1}(i, j, k) + C_{\text{eyip}}(i, j, k)E_{y,\text{inc}}^n(i, j, k)
\]  \hspace{1cm} (4.13)

where

\[
C_{\text{eye}}(i, j, k) = \frac{2 \varepsilon (i, j, k) - \sigma(i, j, k)\Delta t}{2 \varepsilon (i, j, k) + \sigma(i, j, k)\Delta t}
\]

\[
C_{\text{eyhx}}(i, j, k) = \frac{2\Delta t}{\Delta z(2 \varepsilon (i, j, k) + \sigma(i, j, k)\Delta t)}
\]
The updating equation for $E_{z,scat}^{n+1}$ ($i$, $j$, $k$) can be written as:

$$E_{z,scat}^{n+1}(i, j, k) = C_{eze}(i, j, k)E_{z,scat}^{n}(i, j, k) + C_{ezhy}(i, j, k) \left[ H_{y,scat}^{n+1}(i, j, k) - H_{y,scat}^{n+1}(i, j, k - 1) \right] + C_{ezhx}(i, j, k) \left[ H_{x,scat}^{n+1}(i, j, k) - H_{x,scat}^{n+1}(i - 1, j, k) \right] + C_{ezeic}(i, j, k)E_{y,inc}^{n+1}(i, j, k) + C_{ezeip}(i, j, k)E_{y,inc}^{n}(i, j, k)$$  (4.14)

where

$$C_{eze}(i, j, k) = \frac{2 \varepsilon(i, j, k) - \sigma(i, j, k)\Delta t}{2 \varepsilon(i, j, k) + \sigma(i, j, k)\Delta t}$$

$$C_{ezhy}(i, j, k) = \frac{2\Delta t}{\Delta z(2 \varepsilon(i, j, k) + \sigma(i, j, k)\Delta t)}$$

$$C_{ezhx}(i, j, k) = -\frac{2\Delta t}{\Delta x(2 \varepsilon(i, j, k) + \sigma(i, j, k)\Delta t)}$$

$$C_{ezeic}(i, j, k) = \frac{2(\varepsilon_0 - \varepsilon(i, j, k)) + \sigma(i, j, k)\Delta t}{2 \varepsilon(i, j, k) + \sigma(i, j, k)\Delta t}$$

$$C_{ezeip}(i, j, k) = -\frac{2(\varepsilon_0 - \varepsilon(i, j, k)) + \sigma(i, j, k)\Delta t}{2 \varepsilon(i, j, k) + \sigma(i, j, k)\Delta t}$$

Following a similar procedure, the updating equations for the magnetic field components can be obtained as follows:
\[ H_{x,\text{scat}}^{n+\frac{1}{2}}(i, j, k) = H_{x,\text{scat}}^{n-\frac{1}{2}}(i, j, k) + C_{\text{hexz}}(i,j,k) \left[ E_{z,\text{scat}}^{n}(i, j+1, k) - E_{z,\text{scat}}^{n}(i, j, k) \right] + C_{\text{hexy}}(i,j,k) \left[ E_{y,\text{scat}}^{n}(i, j, k+1) - E_{y,\text{scat}}^{n}(i, j, k) \right] + C_{\text{hxic}}(i,j,k) H_{x,\text{inc}}^{n+\frac{1}{2}}(i, j, k) \]

\[ + C_{\text{hxhip}}(i,j,k) H_{x,\text{inc}}^{n-\frac{1}{2}}(i, j, k) \]  \hspace{1cm} (4.15) 

where

\[ C_{\text{hexz}}(i,j,k) = -\frac{\Delta t}{\Delta y \mu(i,j,k)} \]

\[ C_{\text{hexy}}(i,j,k) = \frac{\Delta t}{\Delta z \mu(i,j,k)} \]

\[ C_{\text{hxic}} = \frac{\mu_0 - \mu(i,j,k)}{\mu(i,j,k)} \]

\[ C_{\text{hxhip}}(i,j,k) = -\frac{\mu_0 - \mu(i,j,k)}{\mu(i,j,k)} \]

\[ H_{y,\text{scat}}^{n+\frac{1}{2}}(i, j, k) = H_{y,\text{scat}}^{n-\frac{1}{2}}(i, j, k) + C_{\text{hyez}}(i,j,k) \left[ E_{x,\text{scat}}^{n}(i, j+1, k) - E_{x,\text{scat}}^{n}(i, j, k) \right] + C_{\text{hynz}}(i,j,k) \left[ E_{z,\text{scat}}^{n}(i, j, k+1) - E_{z,\text{scat}}^{n}(i, j, k) \right] + C_{\text{hynic}}(i,j,k) H_{y,\text{inc}}^{n+\frac{1}{2}}(i, j, k) \]

\[ + C_{\text{hyhip}}(i,j,k) H_{y,\text{inc}}^{n-\frac{1}{2}}(i, j, k) \]  \hspace{1cm} (4.16) 

where

\[ C_{\text{hyez}}(i,j,k) = -\frac{\Delta t}{\Delta z \mu(i,j,k)} \]

\[ C_{\text{hynz}}(i,j,k) = \frac{\Delta t}{\Delta x \mu(i,j,k)} \]

\[ C_{\text{hynic}}(i,j,k) = \frac{\mu_0 - \mu(i,j,k)}{\mu(i,j,k)} \]

\[ C_{\text{hyhip}}(i,j,k) = -\frac{\mu_0 - \mu(i,j,k)}{\mu(i,j,k)} \]
\[ H_{z,scat}^{n+\frac{1}{2}}(i, j, k) = H_{z,scat}^{n-\frac{1}{2}}(i, j, k) + C_{hzey}(i, j, k)[E_{y,scat}^n(i, j + 1, k) - E_{y,scat}^n(i, j, k)] + C_{hzex}(i, j, k)[E_{x,scat}^n(i, j, k + 1) - E_{x,scat}^n(i, j, k)] + C_{hzhic}(i, j, k)H_{y,inc}^{n+\frac{1}{2}}(i, j, k) + C_{hzhip}(i, j, k)H_{y,inc}^{n-\frac{1}{2}}(i, j, k) \]

where

\[ C_{hzey}(i, j, k) = -\frac{\Delta t}{\Delta x \mu(i, j, k)} \]
\[ C_{hzex}(i, j, k) = \frac{\Delta t}{\Delta y \mu(i, j, k)} \]
\[ C_{hzhic}(i, j, k) = \frac{\mu_0 - \mu(i, j, k)}{\mu(i, j, k)} \]
\[ C_{hzhip}(i, j, k) = -\frac{\mu_0 - \mu(i, j, k)}{\mu(i, j, k)} \]

The updating equations for the scattered field formulation (equations (4.12)-(4.17)) are the same as the updating equations for the total field formulation except that these equations include additional incident field terms (the last two terms). So, it is easy to compute the total field using the scattered field formulation, because total field formulation is a special case of the scattered field formulation. All what we need to do is to save the values of electric field before adding the correction terms.

Figure 4.5 illustrates an incident plane wave travelling in the direction denoted by a unit vector \( \hat{k} \). The incident electric field in general may have a \( \theta \) component and a \( \phi \) component when expressed in a spherical coordinate system denoting its polarization. The incident electric field can be expressed at a given point denoted by a position vector \( \vec{r} \) as:

\[ \vec{E}_{inc} = (E_\theta \hat{\theta} + E_\phi \hat{\phi})f \left( t - \frac{1}{c} \hat{k} \cdot \vec{r} \right) \]

where \( f \) is a function determining the waveform of the incident electric field. In order to simplify the implementation of the source, incident electric field will be assumed to have a
\( \theta \) component only. \( \phi_{inc} \) will be set to zero also. Equation (4.18) can be rewritten by allowing a time delay \( t_0 \) and spatial shift \( l_0 \) as:

\[
\vec{E}_{inc} = E_\theta \theta f \left( (t - t_0) \frac{1}{c} (\vec{k} \cdot \vec{r} - l_0) \right)
\]

(4.19)

![Diagram of an incident plane wave with \( \theta \) and \( \phi \) components][1]

Figure 4.5: An incident plane wave with \( \theta \) and \( \phi \) components [14].

The parameters \( t_0 \) and \( l_0 \) are used to shift the given waveform in time and space such that the incident field in problem space is zero at the start of the FDTD iterations. As time proceeds, the incident field propagates into the FDTD problem space.

### 4.4 Absorbing Boundary Conditions (ABC’s)

Because computational storage space is finite, FDTD problem space size is finite and needs to be truncated by special boundary conditions. However, the imperfect truncation of the problem space will create numerical reflections, which will corrupt the computational results. Several various types of ABCs have been developed. However, the convolutional PML (CPML) will be used in this chapter. CPML offers a number of
advantages, specifically: the application of the CPML is completely independent of the materials modeled in the FDTD space. Secondly, it has been shown that the CPML is highly absorptive of evanescent modes and can provide significant memory savings when computing the wave interaction [47]. Therefore, using the CPML, the boundaries can be placed closer to the objects in the problem space and a time saving can be achieved. CPML is a finite-thickness special medium surrounding the computational space based on fictitious constitutive parameters to create a wave-impedance matching condition, which is independent of the angles and frequencies of the wave incident on this boundary. The theoretical analyses of the CPML and other forms of PML can be found in [48]. During the simulation, an air gap of 5 cells thickness is surrounding the head model, and the boundaries are terminated by 10 cells thickness of the CPML.

4.5 Discretization of Bioheat Transfer Equation

It is assumed that the cell size is equal in all the three dimensions. The cell resolution used in this chapter to calculate electric field and SAR is 2 mm. The discretization of the bioheat equation must have the same cell resolution. The bioheat equation was presented in Chapter 2 and is written for three dimensions as follows:

\[
C_p(i, j, k) \rho(i, j, k) \frac{\partial T(i, j, k, t)}{\partial t} = K(i, j, k) \nabla^2 T(i, j, k, t) + \rho(i, j, k) \text{SAR}(i, j, k) - B(i, j, k)(T(i, j, k, t) - T_b) \tag{4.21}
\]

The boundary conditions for the BHE are represented in the following equation:

\[
K(i, j, k) \frac{\partial T(i, j, k, t)}{\partial n} = -h(T(i, j, k, t) - T_a) \tag{4.22}
\]

By expanding the bioheat equation in its finite difference approximation, (4.21) and (4.22) can be written as follows:
\[ T^{m+1}(i,j,k) = T^m(i,j,k) + \frac{\Delta t}{C_p(i,j,k)} \text{SAR}(i,j,k) - \frac{\Delta t}{C_p(i,j,k)} \rho(i,j,k) (T^m(i,j,k) - T_b) + \frac{\Delta t k(i,j,k)}{C_p(i,j,k) \rho(i,j,k) \Delta z} \left[ T^m(i+1,j,k) + T^m(i-1,j,k) + T^m(i,j+1,k) + T^m(i,j-1,k) + T^m(i,j+1,k) - T^m(i,j,k) \right] \] (4.23)

\[ T^{m+1}(i_{\min},j,k) = \frac{K(i_{\min},j,k) T^m(i_{\min}+1,j,k)}{K(i_{\min},j,k) + h \Delta z} + \frac{T_a h \Delta z}{K(i_{\min},j,k) + h \Delta z} \] (4.24a)

\[ T^{m+1}(i,j_{\min},k) = \frac{K(i,j_{\min},k) (i,j_{\min}+1,k)}{K(i,j_{\min},k) + h \Delta z} + \frac{T_a h \Delta z}{K(i,j_{\min},k) + h \Delta z} \] (4.24b)

\[ T^{m+1}(i,j,k_{\min}) = \frac{K(i,j,k_{\min}) T^m(i,j,k_{\min}+1)}{K(i,j,k_{\min}) + h \Delta z} + \frac{T_a h \Delta z}{K(i,j,k_{\min}) + h \Delta z} \] (4.24c)

### 4.6 Results

The electric field intensity, SAR and temperature elevation distributions have been calculated inside the spherical model at incidence angles of \((0^\circ, 45^\circ)\) at frequencies of 900 MHz, 1800 MHz and 2.4 GHz. In order to give an idea about the general distribution of the electric field intensity, SAR and temperature elevation, these distributions are presented for the case of zero angle of incidence at frequency of 900 MHz in Figures 4.6, 4.7 and 4.8 (without the presence of the metallic implant). Figures 4.9, 4.10 and 4.11 give the distributions when the metallic implant is present.

The presence of the metallic implant decreases the peaks of the electric field intensity and SAR by about 15% and 12%. The general shape of electric field intensity
and SAR distributions changes also. There is an increase in these values in the area around the metallic implant by about three times; which is due to the reflections caused by the implant. Moreover, there is a decrease in the values of SAR and electric field intensity in the position of the implant. Cooper and Hombach [28] investigated the effect of the presence of implants on SAR distribution and found that the presence of a metallic passive element within a homogeneous dielectric lossy sphere can enhance local values of SAR. From Figures 4.8 and 4.11, it is noticed that there is no remarkable difference in the distribution of temperature elevation. This might be due to that (based on the heat transfer properties) the presence of a metallic implant was considered to have no effect on the BHE equation except its effect on the SAR distribution which is considered the external heating source.
Figure 4.6: Electric field intensity for normal incidence at 900 MHz (without implant).

Figure 4.7: SAR distribution for normal incidence at 900 MHz (without implant).
Figure 4.8: Temperature elevation for normal incidence at 900 MHz (without implant).

Figure 4.9: Electric field intensity for normal incidence at 900 MHz (with implant).
**Figure 4.10:** SAR distribution for normal incidence at 900 MHz (with implant).

**Figure 4.11:** Temperature elevation for normal incidence at 900 MHz (with implant).
For normal incidence case, the peaks of the temperature elevation are 0.16 °C, 0.12 °C and 0.15 °C at 900 MHz, 1800 MHz and 2.4 GHz, respectively. These values are in good agreement with what was found in Chapter 2 where it was found that the peak of temperature elevation inside the brain is 0.14 °C, 0.095 °C and 0.13 °C at frequencies 900 MHz, 1800 MHz and 2.4 GHz, respectively. It is worth mentioning that the CPU time needed to find the distributions using the spherical model was about 70 minutes. On the other hand, it took less than 3 minutes to get results using the multi-layered model.

When the plane wave is obliquely incident with an angle of 45° (the plane cut is kept the same at x=0), temperature elevation peak decreases by about 9%, 2% and 3% at 900 MHz, 1800 MHz and 2.4 GHz, respectively. These results agree with what was found in Chapter 3, where it was noticed that as the angle of obliqueness increases, the peaks of SAR and temperature elevations decrease. Another interesting note is that the temperature elevation decreases very quickly at 2.4 GHz compared to the case of 900 MHz. This is due to the fact that the brain has a higher conductivity at 2.4 GHz.
Chapter Five: Conclusions and Future Work

5.1 Conclusions

In this thesis, the effect of RF wave radiated from cellular phones and WLAN antennas on human head was investigated. A multi-layered head model was used to evaluate SAR distribution and temperature elevation. This simple model enabled us to investigate the effect of thin layers on SAR and temperature elevation distributions without the need for very large computational resources. The distributions of electric field and SAR have different shapes at different frequencies. This results from the fact that the dielectric properties are frequency dependent. In SAR distributions, it is noted that the peaks appear in skin and CSF layers. The international level for the SAR is exceeded in these two places. This is due to the fact that these tissues have relatively high conductivity values than other tissues, which means high losses within them. This causes SAR peaks, since the higher conductivity implies a higher SAR. In the literature, the highest cell resolution used was 2 mm, which means that the actual SAR value in the tissues that have small thickness were averaged or even neglected. This averaging led to that the peak values of SAR found in this work are higher than those peaks found in the literature.

Bioheat equation with suitable boundary conditions was solved using FDTD method in order to find the temperature elevation and exposure time effect. It is found that the peak temperature increases exponentially over the first 7–8 minutes, and the rate of temperature rise then slows down. The steady state is reached after about 25 minutes of exposure. The peak temperature rise in the head occurs in the skull and it is up to 0.227°C at 900 MHz, 0.268°C at 1800 MHz, and 0.384°C at 2.4 GHz. It should be pointed out that
the peak SAR occurs in the skin tissue, while the peak temperature rise occurs within the internal bone tissue rather than skin tissue.

Using phones inside enclosed environments was tested and it was found that the peaks of SAR and temperature elevation can be doubled or vanish depending on the distance between the wall and the head model. The effect of the distance between the head model and the source was studied. There is an inversely proportional relationship between SAR (and temperature elevation distributions) and the distance between the head model and the excitation source. For instance, increasing the distance between the source and the head model from 1 cm to 2 cm causes the SAR to decrease by almost three times at 900 MHz and 2.4 GHz while it decreases three and half times at 1800 MHz. Moreover, it causes the temperature elevation to decrease to one fourth of its value at 1 cm.

Aging effect was studied too. The change in dielectric properties due to age does not significantly affect the temperature elevation. This could be explained as a cancellation of the increased conductivity and decreased electric field penetrating into the tissue. This is because of the same degree of increase between the conductivity and permittivity. Moreover, a smaller head size was used to represent the head of a 5-years old child. The results show that the SAR values exceed the international limit level at the beginning of the brain tissue. But, compared to the results of an adult head, the temperature elevation decreases in the whole head model.

In Chapter three, an oblique incidence plane wave was used to excite the head model. Both perpendicular and parallel polarizations were studied. For the same incidence angle, the electric field inside the human head in the case of parallel polarization is larger than that in the case of perpendicular polarization. This may be due to a higher field reflection by the skin layer in the perpendicular polarization case as compared to the parallel
polarization case. Moreover, it is found that as the angle of incidence increases, the temperature elevation decreases.

The obtained results confirm the importance of performing a thermal analysis along with the dosimetric one because the relationship between SAR and temperature elevation is not linear. However, the induced temperature elevation in the brain region, in all the examined conditions, never exceeded 0.4°C. This value is well below the threshold for the induction of adverse thermal effects to the neurons.

In order to obtain an overview of the consequences of using the cellular phones by people who have a metallic implant inside their heads, in Chapter four, a simple model of a homogeneous sphere that has the same properties as the brain was adopted. This model was used to verify the results obtained using the planar multi-layered head model.

The presence of the metallic implant will decrease the peaks of the electric field intensity and SAR and the general shape of these distributions will change also. There will be an increase in these values in the area around the metallic implant by about three times; which is due to the reflections caused by the implant. However, it is noticed that there is no remarkable difference on the distribution of temperature elevation. This is may be due to that the presence of the metallic implant was considered to have no effect on the BHE equation except its effect on the SAR distribution. Moreover, the obtained results are in good agreement with those found using the multi-layered head model.

5.2 Future Work

During this study, the excitation source was assumed to be a plane wave. To extend this work, one has to study the effect of near field sources. It is expected to have some difference on SAR and temperature elevation compared to the ones found using plane wave, but these differences are very small.
Moreover, as a future work, one may take into consideration the effect of sweating on temperature elevation. The effect of using the cellular phone while walking is an interesting topic. This is due to the fact that there is another source of heat inside the body of a walking person. The effect of this extra heat along with the heat due to the EM waves is a questionable issue.
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تقييم معدل الامتصاص الخاص والارتفاع في درجة الحرارة في أنموذج متعدد الطبقات لرأس إنسان متعرض لإشعاع راديوي

الملخص

لقد تزايد في الأونة الأخيرة استخدام أنظمة الاتصالات اللاسلكية التي تستخدم على إرسال واستقبال الموجات الكهرومغناطيسية. هذا الانتشار المتزايد للاتصالات اللاسلكية أدى إلى انتشار الفقлик بين العامة حول تأثير تلك الموجات على الصحة. يعتبر التأثير الحراري للموجات الكهرومغناطيسية هو أحد المؤشرات المثبتة علمياً و التي تستخدم لتحقيق التأثيرات البيولوجية على الإسقاطات الحيوية. هذه الدراسات هي أثيرتوفاء التأثير الموجات الراديوية المتنوعة من الاهتزازات الخلوي وشبكات الاتصال اللاسلكية المحلية (WLANs) (SAR) على رأس الإنسان. لقد تم تمثيل رأس الإنسان بنموذج متعدد الطبقات وتم استخدامه لتقييم معدل الامتصاص الخاص والارتفاع في درجه الحرارة. هذا الأنموذج البسيط نوعاً ما، يمكننا من استكشاف تأثير الطبقات الواقعة على توزيع الارتفاع في درجة الحرارة داخل الرأس البشري دون الحاجة إلى أجهزة حاسوب ضخمة أو وقت كبير. معادلة انتقال الحارة بالتدفق بчерوط مناسبة على حداً الأنموذج مع البيئة المحيطية من أجل إيجاد توزيع الارتفاع في درجة الحرارة وتأثير فترة التعرض للموجات الكهرومغناطيسية على هذا التوزيع. تحوي هذه الرسالة دراسة مكثفة ومعققة حول هذا الموضوع حيث سيبتمع اختيار استخدام الاهتزازات المحمولة داخل اليات الخفيفة كالمصدوع أو أجزاء الجدران الخرسانية. تأثير اسقاطة بين أنموذج رأس الإنسان ومصدر الإشعاع سيبتم دراسته كذلك. إضافة لذلك؛ تأثير العصر على الصورة الكهربائية للأنسجة الحيوية سيبتم دراسته أيضاً. كما سيبتم باستخدام نموذج أخيل رأس طالب وقت طويل من العصر لتوضيح التأثير على الأطفال. سيتم استخدام ألوان كهرومغناطيسية مسلحة ومتى بشكل مثالي على الأنموذج لفحص تأثير زاوية السقوط على شدة المجال الكهرومغناطيسى ومعدل الامتصاص الخاص والارتفاع في درجة الحرارة إلى جانب نموذجdatetime:2023-07-25T21:10:41.301Z

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