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MASTER

Development of a new hyperthermia application for treatment of head and neck tumors: a theoretical exploration

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2002

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Development of a New Hyperthermia Application for Treatment of Head and Neck Tumors: A Theoretical Exploration

door
M.M. Paulides
EM-6-02

Verslag van een afstudeeronderzoek, verricht in de capaciteitsgroep TTE, onder begeleiding van Ir. S.H.J. Vossen (TNO-FEL), Dr. G.C. van Rhoon (Erasmus MC) en Prof. dr. ir. A.P.M. Zwamborn in de periode van januari 2002 - september 2002.

Preface

This thesis describes the results of my final MSc project in Electrical Engineering for the Eindhoven University of Technology (TU/e). The project is initiated by Gerard van Rhoon of the Radiotherapy department of the Erasmus Medical Center in Rotterdam and Peter Zwamborn of the Physics and Electronics Laboratory of the Netherlands Organization for Applied Scientific Research (TNO-FEL). The work is mainly carried out at TNO-FEL situated in Den Haag, The Netherlands. The construction of the head model is done at the hyperthermia group of the Erasmus MC-Daniel den Hoed Cancer Center.

It is a common use to place "Acknowledgements" or "A word of thanks" in the last few pages of a MSc or PhD thesis. In my opinion this does not do justice to everyone who has surrounded and helped me during my entire period of education. In my opinion thanking people should be a major part of a thesis and should thus be done in the preface.

I would like to start with thanking Peter Zwamborn for providing the possibility to perform my project at TNO-FEL and for his hints during the project. I am indebted to Gerard van Rhoon for defining the project, for supplying his knowledge in hyperthermia and for introducing me to and into the hyperthermia part of "the world of medicine". I gained a lot from Stefan Vossen's tremendous effort in helping and guiding me. He was always prepared to discuss results or to review my work, so many thanks go to him.

Of course a pleasant environment is of great importance. Therefore I would like to thank the people at the Electronic Warfare and Electromagnetic effects group of TNO-FEL for helping, guiding and teaching me. In this respect I would like to mention especially Bart, René, Eugène, Guus and Chris for their various contributions. During my project we have had many interesting discussions and pleasant talks that I have enjoyed a lot. The people of the hyperthermia group at the Erasmus MC where always prepared to help as well. I would like to thank Maarten de Bruijne especially for providing me the results of SEMCAD-FDTD.

Last, but not least, I have the challenging task to do justice to all that my friends, my family, my girlfriend Petra and especially my parents have given me in my life. My friends deserve my gratitude for the interest that they showed in me and my work. I want to thank my family for their faith in me and for their moral support during my entire life and Petra for her support during my graduation project. I also want to thank my brother Johan for reviewing this thesis: I will reserve a beer for him. And finally I want to say that without my parents this work would not have been completed by me. Of course because of biological reasons but also because they have challenged me to develop various skills. Skills not only in the scientific sense but also in many other senses, e.g. social, theological and cultural. I would like to conclude this preface with expressing my gratitude to them.
Abstract

In this thesis a theoretical investigation to a new head and neck hyperthermia applicator is described. This project forms the first step towards the development of a head and neck hyperthermia treatment system.

Hyperthermia is defined as a temperature elevation by several degrees (3-7°C) above the normal physiological level. A hyperthermia cancer treatment aims to kill cancerous cells by deliberately increasing the tissue temperature. Because the vascularization is chaotic in certain regions of the carcinoma, an increase of 3-5°C is sufficient to kill the tumor cells. As a result, the temperature just needs a small increase, therefore not affecting healthy regions.

Research in the temperature distribution has to be undertaken to determine whether a hyperthermia treatment, using heating with electromagnetic power, can be effective. This is achieved by calculating the power absorption in the target region from which information of the local temperature can be obtained. The setups are gradually changed towards a physically feasible setup, that represents an actual clinical setup. The calculations are done using a Finite Difference Time Domain (FDTD) code developed at TNO-FEL: FEL-FDTD. This code is adapted for calculation of power absorption distributions in a human body. The antenna characteristics, the influence of the boundary conditions and the influence of the implementation of the FDTD code on the power absorption values are visualized and discussed. The results of FEL-FDTD are compared with a FDTD based, commercial software using the same test setup, however to fully verify the results further research would be necessary.

Results obtained by calculations, with the FEL-FDTD program on a human head model that is illuminated by electric field point sources or dipole antennae, indicate that a local rise in temperature can be achieved. The results obtained with hard sources compared to results with dipole antennae show a large variation, which indicates that additional calculations are needed to optimize the tuning. It is noted that further research is needed to calculate setups with the head model illuminated by dipoles that are embedded in water as a first step into a better modelling of the physical applicator setup.
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Chapter 1

Introduction

This thesis describes a theoretical exploration of applying hyperthermia using electromagnetic power in the head and neck region. The main goal of this project is to find the requirements and possible drawbacks of a head and neck hyperthermia applicator. This chapter is focused on explaining hyperthermia as a treatment and the current status of this application. Further, the problem definition of this project and the structure of this thesis are given.

1.1 Hyperthermia

Following the demonstrated significant advantages of combining radiotherapy with hyperthermia in superficially located cancer by the International Collaborative Hyperthermia Group (ICHG) [ICHG-96], it has become a standard treatment in several institutes. Until now, hyperthermia has mainly been applied to carcinomas in breast and chest walls and in the pelvic region.

Hyperthermia is defined as a temperature increase of several degrees (3-7°C) above the normal physiological level. A hyperthermia treatment aims to deliberately increase the tissue temperature. Cells, in a chaotic vascularized region with their temperature increased to 3-5°C above their normal growth temperature, die or become at least more sensitive to radiotherapy. After treatment with hyperthermia, also an increased effect of some of the chemotherapeutic treatments is reported. Tumor cells are generally equally sensitive to the effects of hyperthermia compared to normal cells. However, in some regions of the carcinoma the vascularization is more chaotic [Molls-00]. Compared to a healthy region, a chaotic vascularization results in:

1. a difference in physiology (low pH, poor nutrient status, small amount of oxygen), resulting in a higher heating sensitivity.

2. a difference in blood flow through the region resulting in a decreased cooling ability. As the cooling ability decreases, the local temperature rises to a higher level, which in turn results in an increased thermal dose.
A carcinoma can be divided into three regions (see Figure 1.1). The first region consists out of necrotic (dead) tissue. Obviously necrotic tissue is not sensitive to hyperthermia. The focal region is indicated by a "2". The conditions, as stated above, are satisfied here. This region is highly sensitive to hyperthermia but less sensitive to radiotherapy or chemotherapy. Region "3" is equally affected as the fourth region consisting of healthy tissue (the rest of the body). Both of the regions are little sensitive to hyperthermia. The tissue in region three is highly affected by a radiotherapy or chemotherapy treatment, which means an additional improvement when combined with a hyperthermia treatment.

1.2 Treatment

A raise in tissue temperature can be obtained by various techniques. One of these techniques uses microwaves for heating the tissue. This is done using one or more antennae depending on the location of the tumor. For the clinical application of hyperthermia, three levels are defined:

**Total body hyperthermia** The entire body temperature is increased to a level between 40.5 and 42°C.

**Regional hyperthermia** The temperature of a part of the body or an organ is increased to 40-43°C in order to kill the tumor cells. This technique is generally combined with chemotherapy.

**Local hyperthermia** The tumor and its surroundings are heated locally. This type of hyperthermia is found to be the most suitable in combination with radiotherapy.

Hyperthermia systems are categorized as semi-deep or deep when the system is capable of heating tumors located more than 4 cm under the skin. In this research, a first step is made towards the application of hyperthermia to carcinomas located in the head and neck region. Because carcinomas in the head and neck region can be located as far as 15 cm below the skin. The system to be developed is qualified as semi-deep.
1.2.1 Current situation

Up to date, little research has been undertaken to the optimization of the application of hyperthermia to the head and neck region. Research into the clinical effects of hyperthermia has been done by several groups. A summary can be found in Table 1.1. Most of the studies in Table 1.1 show a statistical significant higher tumor control and/or cure rate for the combined treatment. The authors that found a minor improvement (Emami et al. 1996, Perez et al. 1991) for the combined treatment, related this failure to substantial problems with the quality of the HT treatment. The most important conclusion from Table 1.1 is that the early non-randomized clinical results are confirmed by recent phase III-trials. Concerning the head and neck region it can be seen that research done by Valdagni et al. shows very promising results on the application of hyperthermia in the head and neck region combined with radiotherapy.

Table 1.1: Hyperthermia treatment — Comparison of the results of radiotherapy (RT) versus radiotherapy plus hyperthermia (RT&HT) in randomized (phase III) trials from western research groups.

<table>
<thead>
<tr>
<th>Author</th>
<th>Tumor</th>
<th>Endpoint</th>
<th>N</th>
<th>RT</th>
<th>RT&amp;HT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Van der Zee et al.</td>
<td>All Pelvic tumors</td>
<td>3 years overall survival</td>
<td>358</td>
<td>24%</td>
<td>30%</td>
</tr>
<tr>
<td></td>
<td>Bladder</td>
<td></td>
<td>143</td>
<td>22%</td>
<td>28%</td>
</tr>
<tr>
<td></td>
<td>Rectum</td>
<td></td>
<td>101</td>
<td>22%</td>
<td>13%</td>
</tr>
<tr>
<td></td>
<td>Cervix</td>
<td></td>
<td>114</td>
<td>27%</td>
<td>51%</td>
</tr>
<tr>
<td>Sneed et al. (1998)</td>
<td>Glioblastoma multiforme</td>
<td>2 years survival</td>
<td>112</td>
<td>15%</td>
<td>31%</td>
</tr>
<tr>
<td>Emami et al. (1996)</td>
<td>Various</td>
<td>2 years survival</td>
<td>184</td>
<td>34%</td>
<td>35%</td>
</tr>
<tr>
<td>Valdagni et al. (1988, 1993)</td>
<td>Head &amp; Neck</td>
<td>compl. resp. rate</td>
<td>44</td>
<td>41%</td>
<td>83%</td>
</tr>
<tr>
<td></td>
<td></td>
<td>5 years survival</td>
<td></td>
<td>0%</td>
<td>53%</td>
</tr>
<tr>
<td>Overgaard et al. (1995)</td>
<td>Melanoma</td>
<td>2 years local NED</td>
<td>134</td>
<td>28%</td>
<td>48%</td>
</tr>
<tr>
<td>Perez et al. (1989, 1991)</td>
<td>Various</td>
<td>compl. resp. rate</td>
<td>236</td>
<td>30%</td>
<td>32%</td>
</tr>
<tr>
<td></td>
<td></td>
<td>overall in small tumors</td>
<td>55</td>
<td>39%</td>
<td>52%</td>
</tr>
<tr>
<td></td>
<td></td>
<td>in large tumors</td>
<td>181</td>
<td>27%</td>
<td>25%</td>
</tr>
<tr>
<td>ICHG (1996)</td>
<td>Breast cancer</td>
<td>complete response rate</td>
<td>308</td>
<td>41%</td>
<td>59%</td>
</tr>
</tbody>
</table>

1 Statistical significant difference  
2 NED: No Evidence of Disease  
3 tumor diameter < 3cm  
4 tumor diameter ≥ 3cm

1.3 Problem definition

Research into the quality of a hyperthermia treatment has been a hot topic for a long time. An improved quality of the hyperthermia treatment is expected to increase the probability
of a positive clinical result. Because of this expectation, most of the applicators must be developed location-specific. In practice, this implies that specific and highly specialized research is focussed on the design and construction of an optimized hyperthermia applicator that is able to heat tumors located at a specific location. The head and neck region is complex, since it is a very heterogeneous region where the tissue surface contours are very irregular. This has consequences to the possible control on the steering capabilities and the occurrence of "hotspots" (e.g. unwanted high absorption areas in the healthy tissue). The work described in this report is meant to be a theoretical exploration to the requirements of an applicator that can be used to treat carcinomas in the head and neck region. It forms the first step into the development of a head and neck hyperthermia treatment system. Previous research, mainly done at other sites, already defines many design rules, which are discussed in the following sections.

1.3.1 Design rules for a general applicator

Two main design rules are defined for a general applicator, which applies hyperthermia. Firstly, as the inner geometry differs significantly from one patient to another, a hyperthermia system must be capable to adapt to this. Therefore the geometry-dependent Specific Absorbtion Rate (SAR) distribution needs to be adjustable. This ability gives the possibility to adjust the temperature distribution patient specifically. Secondly, a good mechanical flexibility is needed, to easily and efficiently position the applicator at the target area. For example, the outer dimensions of a patient can vary significantly.

1.3.2 Design rules for a head and neck applicator

The design rules that count for an applicator in general also apply to the head and neck applicator. Because of some special properties of this region, some additional design rules arise, these rules being:

1. The applicator must be able to heat tumors located at the entire head and neck region. Therefore it needs to be able to treat carcinomas that are located both at the center of the head and neck area and just under the skin.

2. The hyperthermia field widely covers the entire radiation field, e.g. the radiation field plus a minimum safety margin of 1 cm overlap. The hyperthermia field is defined as the area that is at least enclosed by a 25% iso-SAR contour.

3. The SAR distribution can be controlled with a good spatial resolution.

4. The applicator has the ability to conform to the outer tissue contour.

The first rule results from the advantage of one system for the entire region, which should provide more insight in the performance of one general system when compared to different systems for different regions. The second rule is important to be sure of the clinical effectiveness of the treatment. Because the target region is at least enclosed, a sufficient thermal dose is ensured. The third design rule is a direct result from the knowledge that the head
and neck region is very heterogeneous. To respond to large differences in SAR distribution, good spatial resolution of the system is necessary. Because the neck radius differs from patient to patient, the fourth design rule is defined. The system must be designed to be compliant with this last rule.

1.4 Organisation of the report

This report is divided in three main parts, i.e. the head segmentation, the Hyperthermia Treatment Planning System (HTPS) and predictions of power absorption. The second chapter explains the basics of transforming CT data to a model of the head and neck. This head segmentation can be used as input for the HTPS. Chapter three describes the used HTPS and explains the basics of such a program, and of the fundamental Finite Differences Time Domain code. The performance is shown by results of some elementary set-ups. Chapter four provides the Power Absorption distributions in the head and neck model. This chapter shows the ability of an array of antennae to achieve a local hyperthermia that can be used to treat carcinomas. The fifth chapter provides conclusions and the necessity for further research.
Chapter 2

Head segmentation

This chapter provides a possible way to perform a head and neck segmentation. To carry out the segmentation a program called "Hyperplan" developed at the "Charité Medical School, Berlin", is used. This segmentation is used to calculate Power Absorption distributions in the head and neck area. These Power Absorption distributions are given in Chapter 4. The description of the head and neck segmentation tool is important because it provides a better understanding of the conclusions from the power absorption distributions.

When hyperthermia is applied, both the heat-transfer distribution and the electric and magnetic-field distributions are needed to calculate the heat deposition in the head and neck region. Because of the head's heterogeneous structure and its shape, field computations cannot be done analytically. This implies the use of numerical methods when field computations in this region are useful. In general, numerical models divide the target volume into a finite number of volume elements (e.g. voxels). For FEL-FDTD these elements need to be spaced homogeneously along each of the x, y and z-axis (i.e. voxels do not need to be cubical). The dielectric properties of every voxel must be found for the calculation of the electric and magnetic fields. The total model containing the array of electrical properties is called a head segmentation.

2.1 The head and neck model

In practice, it is difficult to gain information about the head and neck structure via an interstitial way. Therefore, scans are made and are used to allocate properties to every voxel in the model of the head. This model, the head segmentation, is used as input for calculations of the electromagnetic field. From the electromagnetic field, the SAR distribution can be calculated.

2.1.1 Data acquisition

For the project described in this thesis, Computerized Tomography (CT) scans are used to find the properties of every voxel. This method uses a small beam of γ-radiation that
passes through the patient. The properties of a "line" from transmitter to receiver can be calculated using the $\gamma$-radiation ratio. This ratio is defined as the quotient of the received and the transmitted radiation. Out of the 1-D measurements at various angles, a 2-D representation can be computed. The scanner is moved axially in order to obtain several axial "slices". See Figure 2.1 shows a CT-scan of the body's cross-section where the grey-scale values indicate a low or high absorption of $\gamma$-radiation. The person to be scanned lies on a metal couch, where a specific head and neck mask keeps the head of the patient positioned against the couch during the scan. This is done to ensure a high accuracy. Both couch and mask are visible in the CT scan at the back of the patient's head.

2.1.2 Modelling

From axial 2-D slices a 3-D model of the head can be constructed. This is done automatically when loading the data of the CT scan in "Hyperplan", which will be used to perform the head segmentation. After the 3-D model is constructed within the program, four different regions of grey-scale values can be defined and labelled automatically thus obtaining a rough head segmentation. The four different regions that are allocated in this stadium are: exterior - muscle - fat - bone. The three threshold values of these regions are set manually. The program can also automatically change regions located within other tissues, that are originally allocated as the exterior region, to so-called (air) bubbles. The amount of different tissue types (grey-scale values), using certain threshold values, can be seen in Figure 2.2. The plot is a "Hounsfield histogram" that plots the amount of voxels, for every grey-scale value, on a logarithmic scale. This visualization can be used to iteratively find the three optimal threshold values for the segmentation.

The first step, after the automatic segmentation, is to remove couch and mask, which can be done automatically but this tool is not very robust. The couch and the mask are
therefore removed manually. This requires some knowledge about the patients anatomy. Taking into account the grey-scale values of the original CT scan provides a way to do this rather precisely without a lot of anatomical knowledge. The next step requires two main tools being:

**Fill:** To be used to "fill" surrounded regions: when an "island" of a certain tissue type is surrounded by a specified tissue type, this island is given properties equal to those of its surroundings

**Remove:** This tool results in the removal of islands with a voxel size up to a specified value, so changing the tissue type to the tissue type which is mostly present in the surrounding area

The program also contains a tool to re-sample the data in order to obtain larger (or smaller) voxels. After these steps are taken the actual manual segmentation starts. In this thesis over ten different tissue types are allocated (Table A.1 and Table A.2). To achieve this, the program provides a "paint-wise" tool. With this, a certain tissue type can be allocated to a single voxel or more voxels. This step is carried out mainly to define certain major organs and to remove artifacts. To be able to perform this step rigourously, knowledge of the patients anatomy is needed or needs to be obtained. The electrical properties of human tissue is frequency dependent so the frequency at which simulations will be done have to be determined first.
2.1.3 Operating frequencies of interest

The operating frequencies are chosen such that the results can be verified by measurements. Two set-ups that can be used for measurements are available at the hyperthermia group of the Erasmus MC-Daniel den Hoed Cancer Center. These set-ups operate at different frequencies:

1. 100 MHz: Frequency where the BSD, a commercial system, can operate at a high power (the total frequency range of the system is 70MHz-150MHz).

2. 433 MHz: Operating frequency of the superficial-hyperthermia waveguide-system developed at the Erasmus MC.

In literature the optimal operating frequency for a hyperthermia treatment, using electromagnetic power, is expected to lie between 100 MHz and 750MHz. Therefore, calculations using these two frequencies (100 MHz and 433 MHz) provide a relatively good estimation of the possibilities of a general head and neck applicator. Moreover, the requirements for the entire defined frequency range of interest can be extracted from results obtained using these two frequencies. Finding an optimal frequency for the head and neck hyperthermia applicator is beyond the scope of this research.

2.1.4 Electrical properties of the different tissue types

The head segmentation basically allocates numbers to the pertaining voxels and thus specifying the tissue types. Information about the actual electrical properties of the tissue types between 10 Hz - 100 GHz can be found at [EProp-02]. The electrical properties that are used for the work described in this thesis can be found in Appendix A: Table A.1 and Table A.2. In this thesis the temperature dependence of the electrical properties is left out of considerations. Many of the electrical tissue properties can be assigned straightforward but problems arise when assigning properties to the trachea. The properties of the trachea given in the two above mentioned tables are in fact of its border. Therefore assigning this value to the entire trachea would produce a homogeneous filled "cylinder". In fact it is filled with air so an option would be to assign the properties of vacuum to the interior of the trachea. This, however, is not entirely correct because the air inside the body has a very high water content. However, assigning a different value to the interior of the trachea, compared to the exterior value, leads to difficulties in choosing the location of the "exterior-trachea interior" boundary.

Until now, most of the hyperthermia applicators use waterboli or are filled with water. A waterbolus is a water filled bag that is positioned between the patient and the applicator. In some set-ups the applicators are placed within the bolus. This waterbolus is used to be able to cool the skin of the patient but, more important, also to decrease the reflections at the skin of the patient. Because of the presence of the waterbolus, an option is to assign the properties of non-conducting water to the exterior area. Doing so, an estimation of the performance of the treatment when applying the waterbolus can be obtained. The main practical difficulty lies in changing the absorbing boundary conditions in the HTPS that is used in order to simulate an infinite water-filled exterior region.
2.2 Conclusions

To provide the initial requirements of a head and neck semi-deep cancer treatment, using hyperthermia, limited knowledge about the human anatomy is needed. When, however, accurate and patient specific segmentations are needed the opposite is the case. To model the actual anatomy of the patient as accurate as possible, knowledge about the locations of, e.g., different organs is required.

"Hyperplan" is a useful program to perform the head segmentation. It contains many tools for obtaining a fast and accurate segmentation.

Calculations at 100 MHz and 433 MHz are sufficient because the aim of this project is not to find an optimal operating frequency but to obtain a general idea about the requirements of a head and neck applicator. These frequencies are chosen because at these frequencies measurements can validate the simulations.

To implement the effects of a waterbolus, the exterior value can be set to (non-) conducting water. Calculations using this exterior value approximate the effect of a waterbolus.
Chapter 3

Hyperthermia Treatment Planning System

This chapter will show what a Hyperthermia Treatment Planning System (HTPS) is and how it works. The basic Finite Difference Time Domain (FDTD) algorithm in the HTPS will be mentioned and possible difficulties of FDTD algorithms will be noted. Some calculations with the HTPS are included to illustrate limitations of FDTD. The last part of this section describes calculations with a muscle equivalent filled cylinder that is used to demonstrate the result of Power Absorption calculations. These are meant as a leg-up to calculations with a more realistic head model (Chapter 4) and to be able to verify results with a commercial computer program.

3.1 Introduction

A Hyperthermia Treatment Planning System (HTPS) is an algorithm that calculates the Power-Absorption (PA) distribution in the segmentation of the human target region when illuminated with antennae. Subsequently, the Specific Absorbtion Rate (SAR) distribution in the bounded volume is calculated. Finally, the SAR distribution together with a heat flow model is used to find an approximated local temperature distribution. The temperature distribution provides information of the expected killed tumor cells. These steps are used to iteratively find the optimal settings of the applicator.

In the case of this research, the HTPS will be used to show whether sufficient power absorption at the target region can be reached from which the amount of cells that are killed can be predicted. It can also be used to find possible difficulties when applying the hyperthermia treatment in the head and neck region.

A HTPS requires two major input data files, namely a head segmentation (see page 6) and the applicator properties.
3.1.1 Alternatives

The HTPS that will be used for this research has other priorities than one used at a clinic. Important features of the HTPS that will be used for this research are: available knowledge of the limitations of the program, adjustability, accuracy of the results, cost of the development, speed and system requirements.

At the beginning of the project two alternative HTPS programs were available that were potentially useful for this project:

1. FEL-FDTD
2. SEMCAD-FDTD

Both are not really developed as an HTPS but can calculate power absorption distributions. The calculation of the temperature is not implemented in FEL-FDTD but is implemented in SEMCAD-FDTD. The programs both are based on an FDTD (Finite Difference Time Domain) algorithm. The SEMCAD-FDTD program is commercially available and developed by Schmid & Partner Engineering AG [SEMCAD]. The FEL-FDTD code is developed by TNO-FEL researchers.

Based on the availability of the entire code of the program and based on the experience available at TNO-FEL, the calculations are done using the FEL-FDTD program. For other research at the Erasmus MC, SEMCAD-FDTD will be used. The FEL-FDTD results must, therefore be compared to SEMCAD-FDTD results.

An additional option that becomes interesting when trying to model physical antennae are programs based on the Integral Equation. This is beyond the scope of this work but can be implemented in future research.

3.2 The FEL-FDTD program

The FEL-FDTD program uses a FDTD algorithm and is originally developed as an electromagnetic solver, for example for radar cross-section calculations. Experiences obtained for temperature raise predictions in the human head caused by mobile phones and its universal setup makes the FEL-FDTD suitable to be used as an HTPS.

3.2.1 Computational areas

The first step when using FEL-FDTD algorithm is the definition of some computational areas. The different computational areas are shown schematically in Figure 3.1. Three main areas are defined:

- The region of interest which is defined by the inserted head segmentation, denoted by "Model region".
- Some extra voxels surrounding the model region which are needed to place antennae ("Additional space"). As the boundary condition never absorbs 100% of the incident
Figure 3.1: Schematic representation of the different computational areas.

EM waves in this case, see Section 3.2.4, this region is needed to decrease the influence of the unwanted backward scattering at the boundary on the PA distribution. These voxels are allocated the properties of free-space.

- The boundary conditions that are needed to absorb scattered power and therefore simulates the model to be in free space, see Section 3.2.4.

The electric and magnetic field in the model region and the additional space are calculated using an FDTD algorithm. This algorithm will be explained in following section. The Absorbing Boundary Conditions (ABC) region is discussed subsequently.

3.2.2 Basic FDTD algorithm

The FEL-FDTD program exploits a FDTD based algorithm. This algorithm is derived from Maxwell’s equations. Assuming a linear, homogeneous and isotropic medium, characterized by \( \sigma, \epsilon \) and \( \mu \), Maxwell’s equations for continuous fields in the time-domain are given by:

\[
\begin{align*}
\nabla \cdot \mathbf{E}(r, t) & = \frac{\rho_v}{\epsilon} \\
\nabla \cdot \mathbf{H}(r, t) & = 0 \\
\n\nabla \times \mathbf{E}(r, t) & = -\mu \partial_t \mathbf{H}(r, t), \\
\n\nabla \times \mathbf{H}(r, t) & = \epsilon \partial_t \mathbf{E}(r, t) + \mathbf{J}_D^e(r, t),
\end{align*}
\]

where \( \mathbf{E} \) denotes the electric field intensity, \( \mathbf{H} \) denotes the magnetic field intensity and \( \rho_v \) denotes the volume charge density. \( \mathbf{J}_D^e \) denotes the conduction current density and \( \epsilon \partial_t \mathbf{E} \) the displacement current density. \( \mathbf{r} \) denotes the Cartesian position vector and is defined by \( \mathbf{r} = x \mathbf{u}_x + y \mathbf{u}_y + z \mathbf{u}_z \), where \( \mathbf{u}_x, \mathbf{u}_y \) and \( \mathbf{u}_z \) are the Cartesian unit vectors, and \( t \) denotes time. The units are presented on page 52 of this thesis. In this thesis, bold face characterizes a vector quantity and calligraphic face characterizes the time domain. Eq. 3.3 is referred to as Faraday’s law and Eq. 3.4 as Ampere’s circuitial law. The derivation of the FDTD algorithm is done extensively in literature, e.g. [Tafel-95], and will therefore not be carried out here. However, a brief introduction to FDTD will be given below.
For the derivation of the FDTD algorithm, some notations are introduced. A space point at \( t = n \Delta t \) in a uniform, rectangular lattice is denoted by (see Figure 3.2) [Yee-66]:

\[
(i \Delta x, j \Delta y, k \Delta z) = (x, y, z)
\]

where \( n = 1, \ldots, N, i = 1, \ldots, N_x, j = 1, \ldots, N_y, \) and \( k = 1, \ldots, N_z \) with \( N \) the amount of time steps. \( N_x, N_y, \) and \( N_z \) are the maximum amount of voxels in the \( x, y \) and \( z \) direction. Any vector component in space and time is now denoted by:

\[
F_{q}^{n,i,j,k} = \mathcal{F}_{q}(i \Delta x, j \Delta y, k \Delta z, n \Delta t), \quad q = x, y, z
\]

where \( \Delta x, \Delta y \) and \( \Delta z \) are the space increments and \( \Delta t \) is the time increment. The basic, source-free, 3-D FDTD update equations are given by six equations, for the three spatial directions both the electric field and the magnetic field are calculated separately. The directions of the field components are shown in Figure 3.2. The update equations of the \( x \)-components are given by:

\[
E_{x}^{n+1,i,j,k} = C_{1E} E_{x}^{n,i,j,k} + C_{2E} \left( \frac{H_{x}^{n+\frac{1}{2},i,j+\frac{1}{2},k} - H_{x}^{n+\frac{1}{2},i,j-\frac{1}{2},k}}{\Delta y} - \frac{H_{y}^{n+\frac{1}{2},i,j,k+\frac{1}{2}} - H_{y}^{n+\frac{1}{2},i,j,k-\frac{1}{2}}}{\Delta z} \right), \quad (3.7)
\]

\[
H_{x}^{n+\frac{1}{2},i,j,k} = H_{x}^{n-\frac{1}{2},i,j,k} + \frac{\Delta t}{\mu_{i,j,k}} \left( \frac{E_{y}^{n,i,j,k+\frac{1}{2}} - E_{y}^{n,i,j,k-\frac{1}{2}}}{\Delta z} - \frac{E_{z}^{n,i,j,k+\frac{1}{2}} - E_{z}^{n,i,j,k-\frac{1}{2}}}{\Delta y} \right), \quad (3.8)
\]

where

\[
C_{1E} = \frac{1 - \sigma_{i,j,k} \Delta t}{1 + \sigma_{i,j,k} \Delta t} \quad \text{and} \quad C_{2E} = \frac{\Delta t}{1 + \sigma_{i,j,k} \Delta t}
\]

Similar equations for the \( y \) and \( z \)-components, respectively, are given in Appendix B. The field components are positioned according to Yee’s mesh [Yee-66], which is schematically shown in Figure 3.2. From both the formulas and Figure 3.2 it is easily seen that the magnetic field components are positioned half a voxel size away from the electric field components forming a "staggered grid". Another observation is that the three components of either the electric and magnetic field are not positioned at the same point. This means that the total electric and magnetic field can never be calculated exactly at a specified point but must be found by interpolation of neighboring values.

**Stability**

To ensure accuracy of the computed results, the cell size must be taken as a small fraction (e.g. \( 0.1 \lambda \)) of either the minimum wavelength expected or the minimum scatterer dimension. When the cell dimensions are chosen, the stability of the time stepping algorithm in the FDTD scheme is ensured by the *Courant criterion* (Eq. 3.9):

\[
\Delta t = \min \left( \frac{\Delta x}{\lambda}, \frac{\Delta y}{\lambda}, \frac{\Delta z}{\lambda} \right)
\]
\[ \Delta t_{\text{max}} \leq \frac{1}{c \sqrt{\frac{1}{\Delta x^2} + \frac{1}{\Delta y^2} + \frac{1}{\Delta z^2}}} \]  

in which \( c = c_0 (\varepsilon_r \mu_r)^{-\frac{1}{2}} \) denotes the maximum wave phase velocity. In most cases \( c = c_0 \), as the region outside the object is usually chosen to be free-space (\( \varepsilon_r = \mu_r = 1 \)). The maximum time step is \( \Delta t_{\text{max}} \) and \( \Delta x, \Delta y \) and \( \Delta z \) are the voxel dimensions. This criterion is only valid when the target region consists of loss-less, locally homogeneous material.

### 3.2.3 Antennae

In the available FDTD program it is very time consuming to insert physical antennae. Two main steps have to be performed. At first the geometrical (Perfect Electric Conducting) structure of the antenna needs to be inserted. Antennae normally are non-conformal to the FDTD grid and therefore approximations need to be made. The next step is to insert the electromagnetic source(s). These can be either an applied \( E \)-field or enforced current at, for example, the gap of a dipole antenna (Figure 3.3). The \( E \)-field or currents can be either time harmonic or finite in duration. Many different kinds of antennae can be modelled in this way but for this project only two kinds were used and will therefore be discussed. The hard sources are used as a first approximation because they are easy to implement, whereas the dipoles are used because of their great potential of being used in the hyperthermia applicator.
Hard sources

The first option the FEL-FDTD offers is the inclusion of an $E$-field point source without the surrounding geometrical structure of the antenna. Basically this means that at a certain specified voxel, an electric or magnetic field is enforced that "radiates", as a result of the FDTD algorithm, to other voxels. This field can be a periodic or transient function, e.g. a sine or a Gaussian pulse. In literature this source is referred to as a "hard source" [Tafl-95]. Because of the simplicity to implement it in FDTD the hard source is used to do some initial calculations. An advantage of this source is that it is not influenced by electromagnetic coupling effects with other point sources or the target object. The hard source however does cause "spurious nonphysical retro reflection" back to the material structure [Tafl-95].

Dipole antennae

A dipole is made of a Perfect Electric Conducting (PEC) wire with a certain length and thickness. The wire consists of two parts and is fed at the center. In analytical considerations the length $L$ of an infinitely thin resonant wire dipole is half a wavelength: $L = \lambda/2 = c_0/2f$. In practice the wires have a certain thickness. An increase in the radius of the antenna results in a decrease of the resonant length of the antenna. Thus the length of the antenna needs to be reduced. In FDTD (excluding using some special models) the thickness of the dipole must be the multiple of the voxel size, so the smallest possible size is that of one voxel (see Figure 3.3). This is obtained when setting $\sigma$ to infinite, i.e. PEC. In FDTD, this is normally approximated by allocating a high value, for this work 1 MS/m, to $\sigma$.

The dipole is usually modelled with a square cross-section instead of the actual circular cross-section. This is necessary to be conformal to the rectangular grid. To feed the dipole generally a an applied electric field is placed in the gap, which is at the center of the antenna. A second option is to apply a current by defining a circular magnetic field. This is done by enforcing magnetic field components at voxels surrounding the gap between the poles of the dipole.
3.2.4 Absorbing Boundary Conditions

As noted previously, using FDTD, the computational domain is divided into voxels. For an efficient computation, the size of the computational domain is restricted by the memory and CPU-speed of the computer. The main problem that arises from the bounded calculation space is the calculation of the field at the boundary. In most cases, the entire exterior of the FDTD calculation domain is taken to be free-space. This means in practice that waves travelling towards a boundary should not be reflected at that boundary. In literature, some methods are described on how to handle these boundaries. One of these methods uses the Absorbing Boundary Conditions (ABC) developed by Mur [Mur-81]. This method is implemented in FEL-FDTD. Better performing methods however, for example Berenger's Perfect Matched Layers (PML), exist. An important future step to gain accuracy of the results is implementing PMLs. Two important notes need to be made when applying ABC:

1. The grid is always truncated in the exterior medium, which is generally free-space.
2. Near the truncation, the field consists of plane waves that are almost normally incident on the boundary.

The article [Mur-81] describes methods to implement 1st and 2nd order ABC. Both 1st and 2nd order ABC are implemented in the FEL-FDTD program but 2nd order are used because they are found to be more accurate.

Influence of the ABC

The article cited previously [Mur-81] notes that ABC are not perfectly absorbing. This is related to the waves not being "almost normally incident" on the absorbing boundary. This scattering effect can have a large influence when positioning sources close to the boundary. When the sources are located too far from the boundary, computational effort increases unnecessarily. To find an optimal distance between the sources and the ABC a setup, using two hard, $E$-field sources, is used. This setup is schematically depicted in Figure 3.4. The $E$-field sources are polarized in the $z$ direction. $x_1$ and $x_2$ are the distances between...
the absorbing boundary region and the sources while $x_3$ denotes the distance between the sources. The absorbing boundary region is situated around the entire computational domain but just two sides are shown in the figure for simplicity. Properties of free-space ($\varepsilon_r = \mu_r = 1$) are allocated to the entire computational domain with the exception of the absorbing boundary. The distance between the absorbing boundary and the sources in the $y$ and $z$ direction is 50 and 40 voxels respectively. In this way computational errors from the ABC other than depicted in the figures are reduced to a minimum. The calculations are done with voxels of $2.65 \times 2.65 \times 2.11 \text{ mm}^2$ ($\Delta x \times \Delta y \times \Delta z$) so the absolute distances in the $y$ and $z$ directions, between source and absorbing boundary, are over 100 mm. The FEL-FDTD program is used to obtain the results.

Results with $X_1 = X_2$

Figure 3.5 and Figure 3.6 show results of calculations with the FEL-FDTD code using the previously described setup. The amplitude of the three different components of the $E$-field are given as a function of $x$, at a straight line through the two sources. In the figures the position in the $x$-direction is given in voxels. The sources are positioned at $x=0$ and $x=43$. The frequency is chosen to be 100 MHz. The number of computational time steps is 4000, however at "40acc", where the abbreviation "acc" stands for accurate, 10000 time steps are used. The values of "40acc" are taken as a reference since solutions with increased $x_1$ and $x_2$ iterate to these values. Afterwards an additional number of 2253 time steps ($1.05\lambda$) are used to find the maximal value, i.e. the amplitude in steady-state. Computations are done for several values of $x_1$ and $x_2$, which are increased simultaneously. The values 2, 10,
Figure 3.6: Influence of distance $x_1$ and $x_2$ on the amplitude of $E_x$ and $E_y$.

40 and 40acc in the figures correspond to the following values for $x_1$ and $x_2$:

1. $X_1 = X_2 = 2$ voxels
2. $X_1 = X_2 = 10$ voxels
3. $X_1 = X_2 = 40$ voxels
4. $X_1 = X_2 = 40$ voxels

Figure 3.5 and Figure 3.6 show significant differences in region $X_3$, which is the region in between the two sources. These differences are large when $x_1$ and $x_2$ are chosen below 10 voxels. Therefore, sources are always placed at least 10 voxels away from the absorbing boundary. The difference between the values using 10 and 40 voxels additional space is 0.3 dB V/m, which is small. In terms of wavelengths, the distance between the boundary and the source increases at higher frequencies. Figure 3.5 shows that the influence of the ABC decreases when $x_1$ and $x_2$ are increased. Therefore the influence of the ABC is expected to decrease at higher frequencies, and to increase at lower frequencies. Calculations with sources that operates at 433 MHz confirm this expectation.

**Results with $X_1 \neq X_2$**

To find the contribution of the ABC to errors in the region indicated with $x_3$, the electric field distribution is also calculated with a changed location of one of the sources. This source is positioned one voxel away from the original position. The values of $x_1$ are taken
equal to those of the previous section to obtain results that can be compared. An error in the position of the sources of one voxel occurs frequently. This is due to the fact that voxels are used and the field is imposed at the side of the voxel (see Figure 3.2). The remaining setup properties are taken equal to the properties that are used in the previous section. The legend shows the variation in $x_1$.

Figure 3.7 shows that the field distribution at $x_3$ is influenced dramatically when the distance $x_2$ is smaller than 5 voxels. This plot again shows that a distance of 10 voxels is a sufficient distance between the boundary and a source when the surrounding properties are set to those of free-space.

### 3.3 Modelling dipole antennae in FEL-FDTD

This research is an exploration to the requirements of an hyperthermia applicator. This applicator will consist out of an array of antennae in order to be able to focus electromagnetic power in the target area. Because of its flexibility, simplicity, available experience and low cost, a dipole antenna has great potential of being used as source. Problems with modelling dipoles in the FDTD's rectangular grid however could introduce unexpected results. This section gives some insight in the performance of dipoles modelled in FDTD.
3.3.1 Impedance calculation

To check the performance of the modelling of dipoles with FDTD, the impedance of a dipole is investigated at different frequencies. One way to find the frequency response of an antenna is via enforcing a Gaussian-pulse shaped E-field \( E = \frac{V}{\Delta z} \) with respect to time at the gap of the antenna. The Gaussian pulse is explained on page 22. The enforced E-field results in a current through the antenna. The impedance is defined as the quotient of the voltage and the current. Because the FDTD algorithm calculates the electric and magnetic field, the current must be calculated afterwards. In order to find the current, Eq. 3.4 is rewritten resulting in:

\[
\mathbf{J}_D(r, t) = \nabla \times \mathbf{H}(r, t) - \varepsilon \partial_t \mathbf{E}(r, t). \tag{3.10}
\]

The current follows from integrating the conduction current density \( \mathbf{J}_D \) over the cross-section \( S \) of the antenna according to:

\[
I(z, t) = \int_S \mathbf{J}_D^c(r, t) dS. \tag{3.11}
\]

Discrete Fourier Transform

A Fourier Transform (FT) algorithm can then be used to calculate the various frequency components in both the amplitude of the electric field (voltage) and the current. As mentioned in Section 3.3.1 the \( E \) and \( H \)-field are calculated at \( t = n\Delta t \) (see Eq. 3.3 and Eq. 3.4) so the current and the voltage can be calculated only at certain points \( n\Delta t \) in the time-domain, i.e. the current and the voltage are "sampled" in time. At this stage, using a Discrete Fourier Transform (DFT), the frequency components can be found out of the samples.

Initially a function \( \mathcal{F}(t) \) that is sampled according to \( \mathcal{F}(t) = \mathcal{F}(n\Delta t) \) is defined. Here \( n = 1, \ldots, N \), with \( N \) the total number of time samples. The FT can then be calculated according to:

\[
\mathcal{F}(\omega) = \int_{-\infty}^{\infty} \mathcal{F}(t) \exp(-j\omega t) dt, \tag{3.12}
\]

resulting in the frequency domain representation of the function \( \mathcal{F}(t) \). The Discrete Fourier Transform (DFT) can be found from Eq. 3.12 resulting in a sampled signal \( \mathcal{F}(m\Delta \omega) \), \( m = 1, \ldots, M \) where \( M \) is the amount of samples in the frequency domain. When the fact that \( \Delta \omega \Delta t = 2\pi/N \) is used, the DFT can be defined as:

\[
\mathcal{F}(m) = \sum_{n=1}^{N} \mathcal{F}(n) \exp\left(-j\frac{2\pi(m-1)(n-1)}{N}\right), \tag{3.13}
\]

with \( n = 1, \ldots, N \) and \( m = 1, \ldots, N \).

The DFT can now be applied to the time-domain electric field and current, resulting in their frequency domain counterparts. Afterwards the impedance per frequency can be found by dividing them according to Equation 3.14:
\[
Z(\omega) = \frac{\bar{E}_z(\omega)\Delta z}{\bar{I}_z(\omega)},
\]

(3.14)

where \(Z(\omega)\) denotes the impedance of the antenna. \(\bar{E}_z(\omega)\) is the amplitude of the \(E\)-field in the gap, \(\bar{I}_z(\omega)\) the amplitude of the current through the antenna and \(\omega\) the angular frequency, \([\omega] = \text{rad/s}\).

When using a Fourier Transform, care need to be taken to the sample frequency being over twice the highest frequency in the source signal, i.e. over two sample points per wavelength, which is known as Nyquist's criterion.

**Gaussian pulse**

A Gaussian pulse is very useful for finding the impedance characteristics as it includes many frequency components (Figure 3.8) and is a smooth function. Another property of a Gaussian pulse is its large slope at both sides of the pulse. Because a limited calculation time is desired, a large slope is necessary to achieve a small error when truncating the pulse. Truncation is done at \(t = 0\) and \(t = t_p\) in which \(t_p\) denotes the pulse length. Sometimes, however, a double Gaussian pulse is used because it has similar frequency characteristics compared to a single pulse but no DC component. DC currents do not contribute to a radiating electric field and can therefore be left out of impedance calculations. Note that frequency characteristics are similar only when the length of the double Gaussian pulse equals twice the duration of the single Gaussian pulse. Both pulses as described above are defined as:

**Gaussian pulse:**

\[
g(t) = \begin{cases} 
\exp \left[ -\frac{(t-t_1)^2}{\sigma^2} \right] & 0 \leq t \leq 2t_1, \\
0 & \text{otherwise}
\end{cases}
\]

(3.15)

or rewritten as:

\[
g(t) = \begin{cases} 
\exp \left[ -\alpha_1(t - \beta_1\Delta t)^2 \right] & 0 \leq t \leq 2\beta_1 \Delta t, \\
0 & \text{otherwise}
\end{cases}
\]

(3.16)

and

**Double Gaussian pulse:**

\[
d(t) = \begin{cases} 
\exp \left[ -4\alpha_2(t - \beta_2\Delta t)^2 \right] & 0 \leq t \leq 2\beta_2 \Delta t, \\
-\frac{1}{2} \exp \left[ -\alpha_2(t - \beta_2\Delta t)^2 \right] & 0 \leq t \leq 2\beta_2 \Delta t, \\
0 & \text{otherwise}
\end{cases}
\]

(3.17)

with \(\beta_2 = 2\beta_1, \beta_1 = 4\tau/\Delta t, \beta_2 = 8\tau/\Delta t, t_1 = \beta_1 \Delta t\) and \(t_2 = \beta_2 \Delta t\) in all three preceding equations (Eq. 3.15, 3.16 and 3.17). The parameters \(\alpha_{1,2}\) and \(\beta_{1,2}\) need to be chosen properly. The domain of \(t\), over which the calculations take place, is chosen to be finite for practical reasons. The domain of \(g(t)\) and \(d(t)\) are thus truncated at \(t = 0\) and \(t = 2\beta_{1,2} \Delta t\). Choose \(\alpha_{1,2} = (4/\beta_{1,2} \Delta t)^2\). For this value of \(\alpha_{1,2}\), the pulse has the values \(\exp(-16)\), for
the Gaussian, and $\exp(-64) - \frac{1}{2} \exp(-16)$, for the double Gaussian pulse respectively, at the truncation. This results in $-139$ dB and $-145$ dB, respectively with respect to their maximum value at $t = \beta_{1,2} \Delta t$. The Courant criterion determines the size of the time step $\Delta t$. Important is to know that $\beta_{1,2}$ determines the frequency content of the pulse.

**Results**

Figure 3.8 shows the Gaussian pulse and double Gaussian pulse in both the time and in the frequency domain see Section 3.3.1. The FT results are found using a Fast Fourier Transform (FFT), which is a fast DFT. In the plot the exact Fourier Transform of the Gaussian and double Gaussian pulse are plotted as well. A grid of cubical Yee-cells is assumed with a cell size of 1 cm. The Courant criterion gives the maximum time step size, $\Delta t = 19.25$ ps. In this example $\beta_1$ is taken to be 64 so the pulse widths are $2\beta_1 \Delta t = 2.464$ ns and $2\beta_2 \Delta t = 4.928$ ns. Note that the single and double Gaussian pulse match perfectly between one and 4.5 GHz. The results become unstable above 4.5 GHz, since the spatial discretization is 6.7 cells per wavelength at this frequency. This instability of the results at this frequency is due to the fact that 7 points per wavelength are needed for an accurate representation of the current (through the poles of the dipole antenna). Another observation is the (perfect) match between the FFT results and the exact solution. Out of preceding results, the FFT appears to be a fast and accurate approximation of the FT and therefore useful for determining frequency components as needed.
3.3.2 Dipole impedance calculation

In this section, the impedance of a dipole is calculated using FEL-FDTD. This illustrates the usefulness of a FFT when used to find the impedance per frequency of an arbitrary antenna. The dipole antenna is modelled with two wires, i.e., two conducting bars positioned in line (Figure 3.3), and the gap between them is one voxelsize. The two bars both contain three voxels in a row with the dimensions of one voxel being: \( \Delta x = \Delta y = \Delta z = 7 \text{ mm} \). The conductivity of the two bars is taken very high in order to approximate perfectly electric conducting bars. In this thesis \( \sigma = 1 \text{ MS/m} \) is used. Simulations are done by using a double Gaussian pulse \( E \)-field excitation at the gap of the dipole, with \( \beta_2 \) being 64. The size of the spatial grid surrounding the antenna is \( N_x \times N_y \times N_z = 30 \times 30 \times 37 \) voxels, so 15 voxels are taken between the antenna and the absorbing boundary. The entire computational domain surrounding the antenna is free space, so \( \varepsilon_r = 1, \mu_r = 1 \) and \( \sigma = 0 \) with \( \varepsilon = \varepsilon_r \varepsilon_0 \) and \( \mu = \mu_r \mu_0 \). The total number of time steps is 1024. The result of this calculation is plotted in Figure 3.9.

Figure 3.9 shows the frequency characteristic of the impedance of a dipole antenna [Leer-95]. The real and imaginary part of the impedance are shown, being respectively the resistance and reactance. The first resonant frequency is \( f_1 \). At this frequency the antenna reactance is zero so all the source energy is used for radiation. The resistance is approximately 75 \( \Omega \) at \( f_1 \).

The length \( L \) of the dipole antenna is 4.9 cm. In theory the antenna is infinitely thin. In the simulation with FDTD however, the length \( d \) equals 7 mm. The sample period is 13.44 ps. The first resonant frequency in theory is when \( L = \lambda/2 \) with \( \lambda = 2\pi c/\omega = c/f \) where \( f \) is the frequency and \( \lambda \) is the wavelength. Therefore the first resonant frequency \( f_1 \) should be at 3.1 GHz. In Figure 3.9, due to its cross-section of 7 \( \times \) 7 mm, \( f_1 \) lies at 2.6

![Figure 3.9: Impedance of a half-wave dipole antenna in FDTD, found using an FFT of a Gaussian pulse.](image)
GHz.

Another observation is the negative resistance between 0 Hz and $f_0$. This is a numerical error because an antenna with a negative resistance is physically not possible. The conclusion out of these results is that the antenna should be used at a frequency close to $f_1$.

3.3.3 Central frequency vs Fourier Transformed results

![Figure 3.10: Impedance of a half-wave dipole antenna in FDTD: central frequency and FFT results.](image)

Power-Absorption-distribution calculations are done at a single frequency. In the preceding section, the impedance characteristic of the dipole antenna is found by using the FFT. In this section the results found with FFT are compared with results obtained with a single frequency excitation. The setup is identical to that of the former section. The single-frequency results are obtained using a harmonic sine-wave as excitation in the gap of the dipole antenna. As the methods use the same FDTD scheme the results should match. The central frequency results uses 4000 time steps and some additional steps to provide the maximum value at every point.

Figure 3.10 shows a plot of the resistance and reactance of the dipole obtained with the FFT and at single frequencies. The results are almost identical, as expected. However, a horizontal offset in either the FFT results or the single frequency results can be noted. This effect is not further investigated as they are within a marginal error.

3.3.4 Contribution of the displacement current on the impedance

In Section 3.3.1 the conduction current density is found by subtracting the displacement current density from the curl of the magnetic field (Eq. 3.10). Often, however, the displacement current is not extracted because it is assumed to be negligible. This section will
show whether this assumption is valid for a dipole antenna. The same dipole as in section 3.3.2 is used. A double Gaussian pulse shaped E-field is enforced at the gap of the dipole. The impedance is calculated according to the steps described in Section 3.3.2.

Figure 3.11 shows the contribution of the displacement current on the impedance. At low frequencies the displacement current has little influence, however at higher frequencies this increases. When using frequencies close to the lowest resonant frequency the displacement current does not need to be extracted. The displacement current is extracted in this work because not all the simulated dipoles in this thesis where used close to their first resonant frequency.

3.3.5 Comparison different types of dipoles

Limited space to position antennae is expected when designing the applicator. Bowtie antennae have a lower resonant frequency compared to thin-wire dipoles of equal length. This results in a smaller antenna with an equal resonant frequency. Because of this pro-

![Diagrams of different dipole antennae](image)

Figure 3.12: Three dipole antennae with different shapes.
property, bowtie dipole antennae have a great potential of being used in the hyperthermia applicator. Since FEL-FDTD uses a rectangular grid, bowtie antennae can not be modelled properly. Adding some voxels at the ends of the poles of the dipole is done as an approximation of the "bowtie shape". Figure 3.12 shows three different structures of which the impedance characteristics will be compared in this section. Dipole a is equal to the example dipole of previous sections. Dipole b is extended at its ends with one voxel, i.e. eight extra voxels and dipole c with two voxels, i.e. 16 extra voxels. The impedance characteristics of the three antennae are calculated using a FFT and are shown in Figure 3.13. In Figure 3.13 the first resonant frequency is lower when increasing the amount of voxels at the ends of the poles of the dipole. The slope of the antenna-impedance characteristic close to the lowest resonant frequency increases, i.e. the band-width of the antennae decreases.

3.4 Comparison between results FEL-FDTD and SEMCAD-FDTD

The FEL-FDTD code was not validated for the purpose of the project. A full validation was outside the time frame for this research. As a compromise, results of FEL-FDTD are compared to those of SEMCAD-FDTD. The Power Absorption distribution in a homogeneous phantom is calculated with both programs. The PA is defined in Section 4.1. It should be noted that the programs use a different discretization of the phantom, therefore the size is not exactly the same. As a first step, the phantom is illuminated with hard sources and subsequently used in calculations with the FEL-FDTD code. The results are expected to be symmetrical due to the symmetrical setup. Afterwards the distributions obtained with both FEL-FDTD and SEMCAD-FDTD using dipoles will be presented and compared.
Homogeneous phantom

The geometry that is used to compare the code is a homogeneous, cylindrical, phantom (see Figure 3.14). This filled cylinder is a first approximation of the neck region.

![Figure 3.14: Schematic view of the homogeneous cylinder.](image)

FEL-FDTD uses a cubical grid for the model whereas SEMCAD-FDTD uses a grid that is non-homogeneous in the x, y and z-direction. Therefore the cylindrical setup described in this chapter is a model used in the FEL-FDTD program only.

The calculations have been performed at 433 MHz. At this frequency, comparison with measured values will be possible afterwards. The cylinder is filled with muscle-equivalent material, which according to literature means that $\epsilon_r = 56.9$ and $\sigma = 0.8$ S/m (see Table A.2). The diameter of the cylinder that is used in the FEL-FDTD program is 42 voxels ($x_3$) and the height is 60 voxels. By defining a voxel size in the x, y and z-direction, the "physical" geometry of the cylinder is defined. In this chapter, the voxel size is chosen to be 0.5 cm in the x, y and z-direction. As a result, the height of the cylinder is 30 cm and the diameter is 21 cm ($\approx 20$ cm). A schematic view of the cross-section of the cylinder is given in Figure 3.14. In this figure the homogeneous cylinder is white and the exterior is black. The properties of the exterior are chosen to be equal to those of free space: $\epsilon_r = 1$ and $\sigma = 0$ S/m.

3.4.1 Results with four hard sources

The first step is to illuminate the cylinder with four electric field "hard" sources. The hard sources are explained in Section 3.2.3. The positioning of these hard sources is schematically shown in Figure 3.15. The four electric field sources are placed on a circle at $z = 0$ cm and are polarized in the $z$-direction. The origin of the "source" circle lies at the central axis of the cylinder so a symmetrical setup is obtained. $x_1$ and $x_2$ are taken as 10 cm (=20 voxels) and $x_3$ is 21 cm. Mur's ABC are used in FEL-FDTD to simulate an infinite problem space. The absorbing boundary is positioned 15 cm (=30 voxels) away from the closest source in x and y-direction. In the z-direction the AB is placed 40 voxels away from the top and bottom of the cylinder. The results of the PA distribution calculation are plotted in Figure 3.16 and shows the PA distribution with respect to $x$ at $z = 0$ cm and $y = 0$ cm. Two curves are visible in the plot of the values at $y = 0$ cm. These are
Figure 3.15: Set-up of the four hard sources.

Figure 3.16: PA distributions (dB) at the "feed plane" (z = 0 cm) in a muscle-equivalent cylinder illuminated by four hard sources, normalized to the global maximum value. The results are obtained with the FEL-FDTD code.

the actual values and the actual values mirrored at x=0. This is done to visualize that the results are non symmetrical. The reason for this will be explained in Section 3.4.3.

The PA distribution clearly shows an increased absorption in the center region of the phantom. Muscle has the highest conductivity compared to the other tissues in the head and neck region. A high conductivity results in a lower penetration depth of the EM waves. Therefore, this figure is a first indication that a focus of energy in the head and neck region with the aid of an antennae system is possible at this frequency.

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3.4.2 Results with four dipole antennae

The second step is the use of the dipole model of Section 3.2.3. The calculation of the PA distribution is performed with the FEL-FDTD program and the SAR distribution with the SEMCAD-FDTD program. The PA and SAR values only differ a tissue dependent factor, the mass density: ρ. All tissues have a high water content so this factor is taken 1000 (kg/m³) for all tissues. Subsequently, the normalized results can be compared. In this simulation four antennae, fed with four hard sources, are used. The dipole antennae are placed around the phantom as in

Figure 3.17: Positions of the four dipoles around the muscle-equivalent phantom

x₁ and x₂ are 10 cm and the size of d is 0.5 cm. The poles of the dipoles are 14.5 cm and the gap between them is 0.5 cm so L = 29.5 cm. Note that SEMCAD-FDTD uses a variable voxel size whereas in the FEL-FDTD program a cubical grid is used. This means that both the muscle-equivalent phantom and the model of the circular, conducting poles of the dipole antennae are placed on a different mesh in the programs. The electrical properties of the phantom however are equal, thus εᵣ = 56.9 and σ = 0.8 S/m. The height is 30 cm in both cases and the diameter (x₃) is 20 cm in SEMCAD-FDTD and 21 cm in FEL-FDTD.

According to [Leer-95], the distance between a large object and an antenna should be more than 0.27λ to avoid cross coupling. At 433 MHz, 0.27λ equals 19 cm, so the object does influence the impedance of the antenna because the antennae are positioned only 10 cm away from the object.

The PA distributions obtained with both FEL-FDTD and SEMCAD-FDTD at the "antenna-feed plane" (z = 0 cm) are shown in Figure 3.18. The values are normalized to the maximum value of the entire computational domain. Figure 3.18 shows that the global shape is the same in both programs with high values at the central region and just beneath the surface of the cylinder.

Cross-sectional plots at y = 0 cm and y = 5 cm are shown in Figure 3.19. As mentioned before, the input of the programs was not exactly identical so differences, though
small, were expected. These differences can clearly be seen in the plots but appear to be small. Further, it can be seen that the SEMCAD results are symmetrical opposed to the FEL-FDTD results that are non-symmetrical around $x = 0$ cm.

3.4.3 Differences between FEL-FDTD and SEMCAD-FDTD

In this section the differences between FEL-FDTD and SEMCAD-FDTD will be discussed. The first and largest difference is that the FEL-FDTD results are non-symmetrical around $x = 0$ cm. This is probably due to the way the PA is calculated from the $E$-field components. The $E$-field components are, according to Yee's grid, calculated at different
locations, therefore interpolation is needed to find the values at one location (see also Section 3.2.2). After interpolation, the PA at that point can be calculated. This interpolation is not implemented in FEL-FDTD but it is implemented in SEMCAD-FDTD. The second difference is the use of Higdon 2nd order boundary condition by SEMCAD-FDTD where FEL-FDTD uses Mur's 2nd order ABC. The third difference is the use of a variable grid in SEMCAD-FDTD and the FEL-FDTD's cubical grid. The variable grid of SEMCAD-FDTD allows the non-uniform reduction of the size of the voxels in three directions at "interesting" regions.

3.4.4 Comparison of hard source and dipole antenna results

The results of previous sections are compared in this section to estimate the influence of the antenna modelling. In previous sections the antennae are modelled as E-field hard sources and, afterwards, as two conducting wires exited by an E-field hard source. The modelling with the hard source is easier to implement and is faster because no extra material properties need to be assigned.

![Figure 3.20: Comparison between FEL hard source, FEL dipole and SEMCAD dipole approximation results.](image)

Figure 3.20 shows results of the FEL-FDTD program and SEMCAD-FDTD program. The FEL-FDTD results are obtained with a hard source and a dipole approximation. The SEMCAD results are obtained using a dipole approximation. The most important observation is that higher PA results are found at the target region in the center when dipoles are modelled. The overall shape of the "FEL hard source" curve is similar to that of both "FEL dipole" and "SEMCAD". From Figure 3.20 it can be concluded the E-field hard source can be used to find results that are an appropriate first approximation for "dipole antenna results". At a later stage, this can be verified through measurements with a muscle-equivalent phantom.
3.5 Conclusions

This chapter has shown the possible difficulties in using FEL-FDTD as HTPS. The main topics that are discussed are: the implementation of the boundary conditions, the modelling of the dipole antennae and a discussion of the results of FEL-FDTD.

The main conclusion concerning the ABC is that additional voxel-space is needed, which has to be more than 10 voxels at frequencies above 100 MHz to achieve accurate results in the central region of the model. A better way is to implement PMLs in FEL-FDTD in order to decrease the influence of reflections.

Further, the implementation of antennae in FDTD is mentioned. It is time consuming to model antennae accurately with FDTD. An E-field hard source can be used to fast and fairly accurately do initial calculations of the power absorption distributions in the head and neck area.

It is found that, near to the lowest resonant frequency, the influence of the displacement current on the impedance of the antenna is small.

Some impedance calculations indicate that a bowtie shaped antenna has a lower first resonant frequency compared to a "normal" dipole antenna.

The last section of this chapter mentioned the results with a muscle-equivalent filled cylinder and a comparison between FEL-FDTD and SEMCAD-FDTD. The symmetrical setup result in power absorption distributions that are approximately symmetrical. Furthermore the results are almost the same for SEMCAD-FDTD and FEL-FDTD applied to a filled cylinder.
Chapter 4

Prediction Power Absorbtion distributions

The project described in this thesis is meant to be an exploration of the requirements of an hyperthermia system. One of the key elements of these requirements is knowledge about to what extend the temperature of the target region can be raised. Calculations are done with the FEL-FDTD program resulting in the Power Absorbtion (PA) distribution. This is useful because the temperature values are related to the PA values. The head segmentation is obtained via the procedure described in Chapter 2.

4.1 Theory

Firstly the theory involved with the PA calculations will be discussed and secondly the results of the PA calculations at various settings will be shown. The first section defines the Power Absorbtion (PA) and the Specific Absorbtion Rate (SAR). Afterwards the dependence of the SAR and PA values in a certain material on the polarization of the electric field is discussed. This will provide a better understanding of the obtained PA distributions. The temperature distribution can be calculated, e.g. with a heat-sink model, which was beyond the scope of this thesis. Therefore, the final part of this chapter will discuss the interpretation of the PA distributions. This is necessary to be able to estimate the temperature distribution from the PA distributions. In the future these temperature distributions might be used to predict a tumor cure probability.

4.1.1 Definitions

The Power Absorption (PA) is defined as:

$$PA = \frac{\sigma |E|^2}{2} \left[ \frac{W}{m^3} \right],$$

(4.1)
and the Specific Absorbtion Rate as:

\[
SAR = \frac{\sigma |E|^2}{2\rho} = \frac{PA}{\rho} \left[ \frac{W}{kg} \right],
\]

where \(\rho\) is the mass density (kg/m\(^3\)), \(\sigma\) the electric conductivity (S/m) and \(|E|\) the amplitude of the electric field intensity (V/m).

Using the FDTD algorithm, the field distribution can be computed. When a complete steady-state maximum field distribution inside the computational domain has been computed, the 3-D PA value at a point is found by:

\[
P_{A,i,j,k} = \frac{1}{2} \left[ \sigma_{i+\frac{1}{2},j,k} \cdot |E_x|_{i+\frac{1}{2},j,k}^2 + \sigma_{i,j+\frac{1}{2},k} \cdot |E_y|_{i,j+\frac{1}{2},k}^2 + \sigma_{i,j,k+\frac{1}{2}} \cdot |E_z|_{i,j,k+\frac{1}{2}}^2 \right],
\]

where \(|E_q|\) is the absolute value of the field component and \(q = x, y, z\). The notation is presented in Eq. 3.5 and Eq. 3.6.

A more accurate way to calculate the PA distribution is to interpolate the \(E\)-field and \(\sigma\) components to obtain the values at a single point. From these interpolated values can be used to calculate the PA value at that point. This feature is not implemented in the FEL-FDTD code because the voxel size is small and only the shape of the PA distribution is searched for. This leads to asymmetric results.

### 4.1.2 Boundary conditions

This section discusses the influence of the electric field polarization on the SAR and PA ratios at boundaries. These ratios have a large influence on the SAR and PA values in the various material types.

The SAR and PA of a dielectric material strongly depends on the polarization of the electric field [RhVi-88]. This can be shown by an assumed boundary between material \(a\), characterized by \(\sigma^a\), \(\varepsilon^a\) and \(\mu^a\) and material \(b\), characterized by \(\sigma^b\), \(\varepsilon^b\) and \(\mu^b\). The boundary conditions that are enforced at the interface between material \(a\) and \(b\) can now be derived for normal and tangential incidence.

1) **Normal incidence:** When the electric field is normal (perpendicular) to the interference, the electric flux density \(\mathbf{D}\) is continuous and given by:

\[
\mathbf{D}_n^a = \mathbf{D}_n^b = \epsilon_r^a \epsilon_0^a \mathbf{E}_n^a = \epsilon_r^b \epsilon_0^b \mathbf{E}_n^b
\]

and because of this, the electric fields in medium \(a\) and \(b\) are directly related to the difference of the permittivities.

The SAR and PA ratio can be calculated using:

\[
\frac{SAR^a}{SAR^b} = \frac{\sigma^a \rho^a (\varepsilon^b)^2}{\sigma^b \rho^a (\varepsilon^a)^2}
\]
and
\[
\frac{PA^a}{PA^b} = \frac{\sigma^a (\epsilon^b)^2}{\sigma^b (\epsilon^a)^2}
\] (4.6)

2) Tangential incidence: When the electric field is tangential (parallel) to the boundary surface, the electric field \(E\) is continuous and given by:
\[
E_i^a = E_i^b
\] (4.7)
in this case the relative PA ratio is directly related to the conductivity:
\[
\frac{PA^a}{PA^b} = \frac{\sigma^a}{\sigma^b}
\] (4.8)

When the density \(\rho\) is equal in both materials, the SAR ratio also depends on the conductivity ratio.

It is shown that the relative SAR ratio for an example structure using fat and muscle can differ up to 300 times when comparing both types of incidence [RhVi-88]. As this is a significant difference, special care is needed when applying hyperthermia after treatment planning indicates an appropriate PA pattern. In this thesis the polarization of the electric field with respect to various interfaces in the region of interest will be mainly tangential because \(z\)-polarized feeds are used.

4.1.3 Discussion value of PA distributions

The objective of this chapter is not to find exact temperatures but is expected to provide general knowledge about the temperature distributions. To meet this objective, in this chapter, only the PA values are calculated. This provides a good approximation of the local temperature distribution. However, some comments need to be made:

1. The difference between the PA and SAR values is solely a tissue-type dependent factor, namely the mass density \(\rho\)

2. Applying a heat-sink model to the SAR pattern (to find the temperature deposition) mainly results in two things. First, high SAR peaks are spread out due to thermal conductivity resulting in lower temperature peaks. Secondly, the temperature near blood vessels is lower than expected from the SAR value due to the cooling ability of the vessels.

When these comments are taken into account, the PA distribution can be used to see if a focal area can be created. The results of this chapter can subsequently be used to conclude whether it is possible to treat head and neck carcinomas with hyperthermia.
4.2 The head model

Chapter 2 describes the steps to built a head segmentation. This head segmentation is used as an input for calculations that result in the PA distributions. As only limited memory capacity was available at the available computer, the model was kept small. Therefore, the model is taken from the nose to the shoulders. A representation of the model is given in Figure 4.1. In this figure only the skin surface is shown and 10 free-space voxels (air) are added on each side for a clearer view. The size of the model is $83 \times 150 \times 69$ voxels, in the x, y and z-direction respectively. The voxelsize is $2.89 \times 2.89 \times 2.50$ mm ($\Delta x \times \Delta y \times \Delta z$) and therefore the head is "placed" in a rectangle of $239.87 \times 433.50 \times 172.5$ mm. The model is stored in a hierarchical data format. It is transformed to ASCII characters in the HTPS. Stored as a binary ".hdf"-file, the size of the model is 844 KB but stored in ASCII format its size increases to 3.0 MB. For the calculations 20 voxels extra space in all directions are added, which is explained in Section 3.2.1.

Figure 4.1: The skin surface of the head segmentation
4.3 Results with hard sources

The first calculations of PA distributions with FEL-FDTD are done using the hard sources that are mentioned in Section 3.2.3. In Section 3.4.4 it was shown that results with hard sources are a reasonable approximation of simulations with dipoles. The first step is to describe the setup and afterwards results with the head model placed in air and water are presented. The results are obtained using illumination of the head by one source and circular arrays of four and eight sources. The hard-sources are excited with a sinusoidal electric field. The target region is in the central part of the neck. All the calculations use 5000 time steps, but afterwards additional time steps are used to find the maximum values.

4.3.1 Setup

To obtain the results presented in this section, $z$-polarized $E$-fields are enforced at fixed points on a circle surrounding the neck: $E_z^{1,4,8} = E_0 \sin(2\pi ft)$ with $E_0$ an arbitrary unit that is of no importance due to the latter normalization of the results. The positions of these hard sources are visualized in Figure 4.2. In Figure 4.2 the point indicated with "1" is the position of the hard source when a single source is used. When an array of four sources is used, the sources at the locations indicated with a "4" are used. All shown hard sources are used when the array is formed by eight sources. The figure also shows the height at which the sources are positioned, i.e. at $z = 43$.

![Figure 4.2: Positioning of the hard sources around the neck at cross-sections through z=43 and x=75 (in voxels).](image)

4.3.2 Results in air

These results are obtained with hard sources that are positioned according to the previous section. They form a first approximation of the PA distributions that can be obtained with the head and neck applicator. To the exterior as well as the additional voxels properties of free-space are allocated.
Figure 4.3 shows the PA distributions of calculations at 100 MHz and 433 MHz. The results are normalized to the maximum value in the head model domain. The figures are shown using a logarithmical scale because in this way the patients geometry can be seen as well. Especially in Figures d, e and f an increase of the PA in the target region is obtained when more sources are used. These figures indicate that an improved PA distribution can be obtained at 433 MHz. At 100 MHz, when with 8 sources are used, the PA distribution is almost homogeneous in the neck slice. Taking into account the increased sensitivity of the carcinoma region, this could already lead to a successful treatment.

4.3.3 Results in water

The results described in this section are also obtained with hard sources that are positioned according to Section 4.3.1. To simulate the effect of a water bolus in which the antennae will be placed, the simulations utilize an environment filled with water. The first advantage is that the bolus matches to the head, so decreasing reflections. The second advantage is that the first resonant frequency of dipole antennae is at a lower frequency, which decreases antennae lengths. An additional advantage is that the water bolus also provides a possibility to cool the skin-surface.

Because the water bolus does not span the total outer region, two different heights of the water regions are used. To use the ABC of air properly the ten voxels next to the absorbing boundary are allocated electrical properties of free-space. This results in the head embedded in a water vessel, which in turn is embedded in an air environment.

The calculations are done at 433 MHz, since this frequency provided the most promising results as was mentioned in the previous section.

PA distributions at 433 MHz with the head model placed in water are shown in Figure 4.4. The properties of the surrounding water are taken equal to those used in the clinic so $\varepsilon_{r,\text{water}} = 78.0$ and $\sigma_{\text{water}} = 0.001 \text{ S/m}$. The target region is assumed to be in the central part of the neck. Again a logarithmic scale is used. Note that the results are normalized to the maximum value in the head model region. The results shown in the figures are normalized to their own highest value, i.e. they are normalized separately.

The most noticeable effect from the results is that Figure 4.4 a, b and c show a, seemingly, decrease of the absorbed power in the target region. This is probably due to high values at the nose region. These high values are used to normalize the values at. This normalization therefore causes the values in central region to be lower with respect to the values obtained in air. The high values at the nose region could be due to resonance effects in either the head or in the water. This can be examined by resizing the height of the water region. Resizing the water region results in higher power absorption values for the three setups. This effect can be seen in Figure 4.4 d, e and f. A further observation is that the difference between the results with 4 and 8 hard sources has vanished. In the results obtained using 1 hard source a lot of the power is absorbed in the back region just beneath the skin. This is due to the small distance from the sources whereas influences of other sources is missing.

Comparing Figure 4.3 to Figure 4.4, shows that the surroundings have a large influence on the PA distribution in the target region. It must be noted that from other simulations a local maximum in PA at the nose region is obtained, especially when one of the sources is positioned close to the nose.
Figure 4.3: Normalized PA distributions when illuminating the head model with hard sources, viewed at cross-sections through $z=43$ and $x=75$. The PA values are normalized to the maximum value in the head domain. The model is placed in free space.
Figure 4.4: Normalized PA distributions when illuminating the head model with hard sources, viewed at cross-sections through $z=43$ and $x=75$. The PA values are normalized to the maximum value in the head domain. The model is placed in water: $\epsilon_r = 78.0$ and $\sigma = 0.001$ S/m, with the surrounding water 89 (a,b,c) or 19 (d,e,f) voxels high.
4.4 Results with dipole antennae

This section describes another step towards an actual head and neck applicator configuration. In this section dipole antennae are used. The calculation of the PA distribution is done using FEL-FDTD. The dipole antennae are exited with a sinusoidal function. The calculations all utilize 5000 time steps, and afterwards additional time steps are used to find the maximum values. The setup is described first and then the results, with the exterior properties set to those of free space, are given.

4.4.1 Setup

To obtain the results presented in this section, again z-polarized E-fields are enforced at points on a circle surrounding the neck of the model, see Figure 4.2. The dipole consists of 15 voxels along the z-direction. The dipole is fed at the center of the allocated voxels. The remaining voxels are filled with highly conducting material. This results in dipole antennae of 3.75 cm which, at 433 MHz, means that the antennae are not resonant. The actual values of the PA distributions are beyond the scope of this research, therefore his dipole can be used to obtain normalized power distributions.

4.4.2 Results in air

In this section, calculations at 433 MHz are shown, where calculations at 100 MHz using an X86 machine with the environment of the head set to air are found to be non-stable, which was not investigated due to the time restrictions of this project.

When a dipole is placed close to an object, its impedance is influenced by that object. In [Leer-95], the threshold distance was found to be 0.27 times the wavelength between a dipole and a large object. At 433 MHz in air this threshold distance is 19 cm. The object will always have an influence on the impedance and the radiating properties of the antenna, because in the hyperthermia applicator the antennae will be closer to the head than 19 cm. The dipole is found to be non-resonant at 433 MHz. However the real part of the impedance is positive and therefore the setup is physically feasible. The imaginary part is highly negative so the dipole acts as a capacitor. Figure 4.5 shows the normalized PA distributions obtained using 4 dipoles. a logarithmic scale is used for plotting this figure. Figure 4.5 shows that, when the dipoles are exited with 433 MHz, an increase of the localized PA can be obtained in the target region. This focussing effect is improved compared to the focussing effect in Figure 4.3.

4.4.3 Results in water

The last phase to approximate the real setup is to embed the head and antennae model in water. However the calculations using dipole antennae embedded in a water environment that in its turn is again embedded in free space are found to be unstable. To calculate this possibility an entire region consisting of water can be used. To do so, the boundary conditions need to be changed to match the water properties. This is one of the improvements that should be implemented further research.
4.5 Conclusions

In this chapter the basic theory of the PA distributions was discussed. Subsequently the results of calculations using hard sources and dipole antennae with the head embedded in air and water are presented.

From Section 4.1 it is important to repeat that the polarization of the field at material boundaries has a large influence on the transmission and reflection of the field, where insight in the temperature distribution can be deduced from the PA distribution. The difference between PA and temperature values is mainly the spreading of small hot-spots due to thermal conductivity. Also the cooling effect of the vascular structure results in a difference between the PA distribution and the temperature distribution.

From the PA distributions presented in Section 4.3 can be concluded that the environment in which the head is placed has a large influence on the PA distribution in the target region. The results also indicate that higher power absorption values located in the target region can be obtained by circumferential illumination.

Section 4.4 presents a PA distribution obtained by using four dipole antennae with the head model positioned in air. A local rise in power absorption can be seen, but as mentioned before, this can not be extrapolated to indicate a local increase when the dipoles are placed in water.

The calculations of dipole antennae embedded in a water environment are found to be unstable. Further research could improve this utility further.
Chapter 5

Conclusions and recommendations

This describes a theoretical exploration to the requirements of a head and neck hyperthermia applicator. To explore the possibilities and drawbacks of such an applicator the power absorption distributions in various setups are calculated. This is achieved using a program that is developed at TNO-FEL.

From this thesis can be concluded that an increase of the power absorption can be achieved in the target region, where the target region is placed at the center of the neck. The results however are an approximation because simulations, which are undertaken utilize setups with properties far from a feasible physical setup.

The results also show that the environment in which the patient and the antennae are placed has a large influence on the power absorption results in the target region.

It is clear that there is a need for more research, where some recommendations for further research are given in the next section.

5.1 Future research

A head model is constructed to be able to calculate the power absorption caused by a hyperthermia treatment. The head model extends from the nose to the shoulders due to limited memory capabilities. Moreover, many calculations needed to be done for the work described in this thesis and therefore the computational time is shortened by a reduction of the head model as indicated. Because at 433 MHz the wavelength in water is approximately 7.8 cm, which is close to the dimensions of the head or the waterbolus, resonances inside the head can occur as was demonstrated in Section 4.3.3. This phenomenon should be investigated further when using the entire head for simulations studies.

In this thesis power absorption distributions are calculated. In this way an estimation of the achievable thermal dose can be obtained. Of course a complete HTPS includes a thermal model to provide a more accurate prediction of the temperature distribution. Models to calculate the temperature distribution from the power absorption exist, e.g. a heat-sink model. Two groups of these models can be defined. Some only take into account the thermal conductivity but others also account for the cooling effect of the blood vessels.
It was beyond the scope of this research to incorporate such a model but it is useful to see the actual effect of the inserted power on the quality of heating inside the target region. A comparison between both methods could be performed to find their respective accuracy.

In the physical applicator dipole antennae have a great potential of being used. These dipoles are approximated in the code by two conducting bars exited by an electric field at the center gap. In the real situation possibly bowtie antennae will be used. Adding some additional conducting voxels at the ends of the dipoles could account for this.

In this thesis the antennae are placed regularly on a circle surrounding the neck. Due to mechanical restrictions of the physical applicator such a setup is not likely to be used. Calculations of various setups therefore need to show which setup is the better one given the mechanical restrictions.

The absorbing boundary conditions that are used at the lattice truncation are developed by Mur [Mur-81] and assume the outer region to be free space. To be able to embed the model and the applicator in water the absorbing boundary conditions must be changed to those of water, however other boundary conditions could be implemented in FEL-FDTD, which would possibly improve the model quality.
Appendix A

Electrical properties of human tissue

Table A.1: Tissue properties at 100 MHz — Electrical properties of the tissue types at 100 MHz that are used for the head segmentation [EProp-02].

<table>
<thead>
<tr>
<th>Number</th>
<th>Name</th>
<th>$\sigma$ (S/m)</th>
<th>$\varepsilon_r$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>Air (Free space)</td>
<td>0</td>
<td>1</td>
</tr>
<tr>
<td>1</td>
<td>Fat</td>
<td>0.0363</td>
<td>6.074</td>
</tr>
<tr>
<td>2</td>
<td>Muscle</td>
<td>0.7076</td>
<td>65.97</td>
</tr>
<tr>
<td>3</td>
<td>BoneCortical</td>
<td>0.06431</td>
<td>15.28</td>
</tr>
<tr>
<td>4</td>
<td>Bubble</td>
<td>0</td>
<td>1</td>
</tr>
<tr>
<td>5</td>
<td>LungInflated</td>
<td>0.3057</td>
<td>31.64</td>
</tr>
<tr>
<td>6</td>
<td>Tooth</td>
<td>0.06431</td>
<td>15.28</td>
</tr>
<tr>
<td>7</td>
<td>Cartilage</td>
<td>0.4746</td>
<td>55.76</td>
</tr>
<tr>
<td>8</td>
<td>BrainGreyMatter</td>
<td>0.5595</td>
<td>80.14</td>
</tr>
<tr>
<td>9</td>
<td>BrainWhiteMatter</td>
<td>0.324</td>
<td>56.8</td>
</tr>
<tr>
<td>11</td>
<td>Gland</td>
<td>0.7943</td>
<td>68.81</td>
</tr>
<tr>
<td>12</td>
<td>Trachea</td>
<td>0.5476 (0)</td>
<td>52.96 (1)</td>
</tr>
</tbody>
</table>
Table A.2: **Tissue properties at 433 MHz** — Electrical properties of the tissue types at 433 MHz that are used for the head segmentation [EProp-02].

<table>
<thead>
<tr>
<th>Number</th>
<th>Name</th>
<th>( \sigma ) (S/m)</th>
<th>( \varepsilon_r )</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>Air (Free space)</td>
<td>0</td>
<td>1</td>
</tr>
<tr>
<td>1</td>
<td>Fat</td>
<td>0.04166</td>
<td>5.567</td>
</tr>
<tr>
<td>2</td>
<td>Muscle</td>
<td>0.8048</td>
<td>56.87</td>
</tr>
<tr>
<td>3</td>
<td>BoneCortical</td>
<td>0.09429</td>
<td>13.07</td>
</tr>
<tr>
<td>4</td>
<td>Bubble</td>
<td>0</td>
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<td>Gland</td>
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<td>12</td>
<td>Trachea</td>
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<td>43.94 (1)</td>
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Appendix B

FDTD update equations

\begin{align}
E_{x}^{n+1}_{i,j,k} &= C_{1E} E_{x}^{n}_{i,j,k} + C_{2E} \left( \frac{H_{z}^{n+\frac{1}{2}}_{i,j+\frac{1}{2},k} - H_{z}^{n+\frac{1}{2}}_{i,j-\frac{1}{2},k}}{\Delta y} - \frac{H_{y}^{n+\frac{1}{2}}_{i,j,k+\frac{1}{2}} - H_{y}^{n+\frac{1}{2}}_{i,j,k-\frac{1}{2}}}{\Delta z} \right) \\
E_{y}^{n+1}_{i,j,k} &= C_{1E} E_{y}^{n}_{i,j,k} + C_{2E} \left( \frac{H_{x}^{n+\frac{1}{2}}_{i+\frac{1}{2},j,k} - H_{x}^{n+\frac{1}{2}}_{i-\frac{1}{2},j,k}}{\Delta z} - \frac{H_{z}^{n+\frac{1}{2}}_{i,j,k+\frac{1}{2}} - H_{z}^{n+\frac{1}{2}}_{i,j,k-\frac{1}{2}}}{\Delta x} \right) \\
E_{z}^{n+1}_{i,j,k} &= C_{1E} E_{z}^{n}_{i,j,k} + C_{2E} \left( \frac{H_{x}^{n+\frac{1}{2}}_{i+\frac{1}{2},j,k} - H_{x}^{n+\frac{1}{2}}_{i-\frac{1}{2},j,k}}{\Delta x} - \frac{H_{y}^{n+\frac{1}{2}}_{i,j+\frac{1}{2},k} - H_{y}^{n+\frac{1}{2}}_{i,j-\frac{1}{2},k}}{\Delta y} \right) \\
H_{x}^{n+\frac{1}{2}}_{i,j,k} &= H_{x}^{n-\frac{1}{2}}_{i,j,k} + \frac{\Delta t}{\mu_{i,j,k}} \left( \frac{E_{y}^{n}_{i,j,k+\frac{1}{2}} - E_{y}^{n}_{i,j,k-\frac{1}{2}}}{\Delta z} - \frac{E_{z}^{n}_{i+\frac{1}{2},j,k} - E_{z}^{n}_{i-\frac{1}{2},j,k}}{\Delta y} \right) \\
H_{y}^{n+\frac{1}{2}}_{i,j,k} &= H_{y}^{n-\frac{1}{2}}_{i,j,k} + \frac{\Delta t}{\mu_{i,j,k}} \left( \frac{E_{z}^{n}_{i+\frac{1}{2},j,k} - E_{z}^{n}_{i-\frac{1}{2},j,k}}{\Delta z} - \frac{E_{x}^{n}_{i,j,k+\frac{1}{2}} - E_{x}^{n}_{i,j,k-\frac{1}{2}}}{\Delta x} \right) \\
H_{z}^{n+\frac{1}{2}}_{i,j,k} &= H_{z}^{n-\frac{1}{2}}_{i,j,k} + \frac{\Delta t}{\mu_{i,j,k}} \left( \frac{E_{x}^{n}_{i,j,k+\frac{1}{2}} - E_{x}^{n}_{i,j,k-\frac{1}{2}}}{\Delta y} - \frac{E_{y}^{n}_{i+\frac{1}{2},j,k} - E_{y}^{n}_{i-\frac{1}{2},j,k}}{\Delta x} \right)
\end{align}
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4.1 The skin surface of the head segmentation

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4.3 Normalized PA distributions when illuminating the head model with hard sources, viewed at cross-sections through z=43 and x=75. The PA values are normalized to the maximum value in the head domain. The model is placed in free space.

4.4 Normalized PA distributions when illuminating the head model with hard sources, viewed at cross-sections through z=43 and x=75. The PA values are normalized to the maximum value in the head domain. The model is placed in water: $\varepsilon_r = 78.0$ and $\sigma = 0.001$ S/m, with the surrounding water 89 (a,b,c) or 19 (d,e,f) voxels high.

4.5 Normalized PA distributions when illuminating the head model with four dipoles, viewed at cross-sections through z = 43 and x = 75. The PA values are normalized at the maximum value in the head domain. The model is surrounded by free space.
List of abbreviations, symbols and constants

List of abbreviations

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
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<tbody>
<tr>
<td>AB</td>
<td>Absorbing Boundary</td>
</tr>
<tr>
<td>ABC</td>
<td>Absorbing Boundary Conditions</td>
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<td>CT</td>
<td>Computerized Tomography</td>
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<tr>
<td>DFT</td>
<td>Discrete Fourier Transform</td>
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<td>EM</td>
<td>ElectroMagnetic</td>
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<td>Erasmus MC</td>
<td>Erasmus Medical Center</td>
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<tr>
<td>FDTD</td>
<td>Finite Difference Time Domain</td>
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<tr>
<td>FT</td>
<td>Fourier Transform</td>
</tr>
<tr>
<td>FFT</td>
<td>Fast Fourier Transform</td>
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<tr>
<td>ICHG</td>
<td>International Collaborative Hyperthermia Group</td>
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<tr>
<td>PA</td>
<td>Power Absorption</td>
</tr>
<tr>
<td>PEC</td>
<td>Perfect Electric Conducting</td>
</tr>
<tr>
<td>SAR</td>
<td>Specific Absorption Rate</td>
</tr>
<tr>
<td>SEMCAD</td>
<td>Simulation Platform for Electromagnetic Compatibility, Antenna Design and Dosimetry</td>
</tr>
<tr>
<td>TNO-FEL</td>
<td>Netherlands Organization for Applied Scientific Research, Physics and Electronics Laboratory</td>
</tr>
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<td>TU/e (EUT)</td>
<td>Eindhoven University of Technology</td>
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List of symbols

$PA$ power absorption ($\text{W kg}^{-1}$)
$SAR$ specific absorption rate ($\text{W m}^{-3}$)
$E$ electric field ($\text{V m}^{-1}$)
$H$ magnetic field ($\text{A m}^{-1}$)
$x, y, z$ spatial coordinates ($\text{m}$)
$x, Y, z$ spatial coordinates (voxels)
$\Delta$ difference; $\Delta x = \text{voxel dimension in x-direction}$
$\rho$ mass density ($\text{kg m}^{-3}$)
$\sigma$ conductivity ($\text{S m}^{-1}$)
$f$ frequency ($\text{Hz}$)
$t$ time ($\text{s}$)
$c$ speed of light in the medium ($\text{m s}^{-1}$)
$\lambda$ wavelength ($\text{m}$)
$\epsilon$ permittivity ($\text{F m}^{-1}$); $\epsilon = \epsilon_r \epsilon_0$
$\mu$ permeability ($\text{H m}^{-1}$); $\mu = \mu_r \mu_0$

List of constants

$\epsilon_0$ speed of light in free space ($3 \times 10^8 \text{ m s}^{-1}$)
$\epsilon_0$ permittivity of free space ($8.854 \times 10^{-12} \text{ F m}^{-1}$)
$\mu_0$ permeability of free space ($4\pi \times 10^{-7} \text{ H m}^{-1}$)
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