

Adaptive Finite Element Analysis for Optimized Design of Coaxial Antenna Fed DI Water-Stowed Waveguide Applicator for Therapeutic Local Hyperthermia System

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Abstract – In this paper, coaxial antenna is used to couple electromagnetic energy into waveguide coupled applicator. The water-stowed, indirectly cooled optimized coaxial coupled waveguide applicator has been developed for local hyperthermia applications using Adaptive Finite Element Method. High permittivity deionized (DI) water is used for indirect cooling in bolus for reduced size of waveguide applicator. The water loading also provides a good coupling of the radiated power into body tissue. The optimization is hence achieved and verified by measuring S parameter and specific absorption rates (SAR) which are found to be satisfactory for localized superficial hyperthermia treatments. The results indicate that coaxial coupling provides better focusing of microwave energy.

Keywords – Adaptive Finite Element Method (A-FEM), Deionized (DI) Water Waveguide Applicator, Co-Axial Coupling, Specific Absorption Rate (SAR).

I. INTRODUCTION

Hyperthermia (also called thermal therapy or thermotherapy) is a type of cancer treatment in which body tissue is exposed to high temperatures (up to 45°C). The process reduces the tumor size by killing cancer cells and damaging proteins and structures within cells [1]. Hyperthermia is used with other forms of cancer therapy, such as radiation therapy and chemotherapy [2]. The basic idea behind this treatment is to create an artificial fever of 41- 45°C in the body of the patient without damaging the benign tissue. Different studies have shown that moderate temperatures can damage and kill cancer cells, causing minimal injury to normal tissues. Hyperthermia may make some cancer cells more sensitive to radiation or harm other cancer cells that radiation cannot damage. Numerous clinical trials have studied hyperthermia in combination with radiation therapy and/or chemotherapy. The combination of hyperthermia with these therapies has been conducted for many kinds of cancers, including liver, cervix, sarcoma, appendix, melanoma, and cancers of the head and brain, neck, lung, esophagus, bladder, rectum, breast, and mesothelioma [1,3-7,10].

Several methods of hyperthermia are currently under study, including local, regional, and whole-body hyperthermia (1,3-9). In local hyperthermia, heat is applied to a small tumor area using various techniques for delivering energy to heat the tumor. Presently the methods used to apply heat include radio-frequency, ultrasound and microwave. Depending on the location of tumor within the

body, there are different approaches to local hyperthermia. Interstitial techniques are used to treat tumors deep seated within the body. This technique allows the tumor to be heated to higher temperatures than external techniques. The heat source, RF or microwave, is then inserted into the probe which is inserted into the tumor. External (Superficial) approaches are however used to treat tumors that are on surface or just below the skin. Accordingly, external applicators apparently placed near the affected area to focus energy on the tumor to raise its temperature. In this paper, water-stowed, indirectly cooled coaxial coupled waveguide applicator has been developed for external (superficial) hyperthermia applications [11-12].

II. COAXIAL TO WAVEGUIDE COUPLED APPLICATOR

The choice of a suitable model depends on the frequency of operation, between 300 MHz and 4.2 GHz, the stratified model may be the most appropriate one. In the present work, model is designed for working frequency of 485 MHz, which is one of the ISM (Industrial Scientific Medical) frequencies. At high frequencies the depth of operation is too small, while at lower frequencies, focusing of the field is difficult. It will also produce high surface temperature because the peak temperatures are located nearer to the surface. However, this is expected since by increasing the frequency of operation, the depth of field penetration decreases in the biomass. If, on the other hand, lower frequencies are used, peak temperatures will be sensed deeper in the muscle, but they will be lower, so more power will be required to achieve specified level of heating at lower frequencies [13-15]. Indirect cooling via water bolus helps regulating surface temperature homogeneity during the hyperthermia. Non-invasive hyperthermia systems using waveguide antenna are less traumatic to patients which also minimize the risk of mixing abnormal cells with healthy tissues. The electromagnetic energy radiated from the waveguide antenna is used to raise the temperature in tumor area in the therapeutic range of 40-45 °C to reduce the tumor volume.

A. Deionized Water

There are various benefits in using deionized (DI) water as stowed material in a waveguide-type applicator. Firstly, the dielectric permittivities of the tissues (except the fat layers) are close to that of water, strong energy coupling

can be easily achieved. Moreover, the rather high value of the real part of deionized water permittivity ($\epsilon_w \sim 81$) permits a considerable reduction (~ 9 times) in applicator aperture dimensions, which is very important in trying to achieve good directivity. Furthermore, it was decided to give priority to a small spillover radiation specification rather than the requirement to cover the maximum area of $10 \times 10 \text{ cm}^2$ with a single applicator. The maximum area can be heated by simultaneously using several applicators or even by scanning a single applicator on the tumor surface. Therefore, the waveguide dimensions were chosen to the minimum so that the dominant TE_{10} mode should be transmitted.

B. Coaxial To Waveguide Coupled Applicator Design

Fig. 1 depicts the perspective view of the developed coaxial coupled applicator is shown.

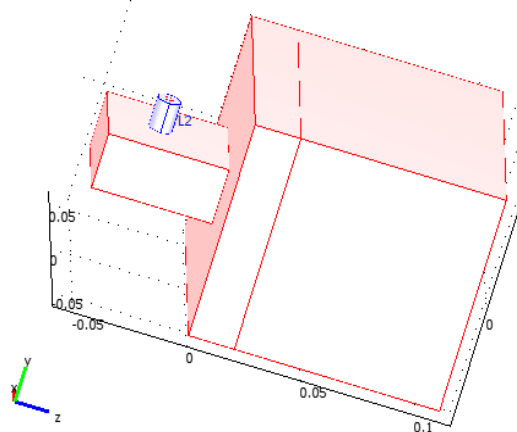


Fig.1. Perspective view of the developed applicator

The energy is fed into the cross-sectional area $a \times b$ ($a = 2.4 \text{ cm}$, $b = 4.8 \text{ cm}$) of the waveguide through a coaxial-to-waveguide coupled waveguide transition by utilizing a coaxial probe-type antenna inside the waveguide. The probe conductor, soldered to the central conductor of an N-type coaxial connector, was extended near to the opposite wall of the waveguide. The length l of one side of the short-circuited waveguide section was chosen equal to $\lambda_g/2$, where λ_g is the wavelength inside the waveguide. The λ_g of TE_{10} mode is given by the relation

$$\lambda_g = \frac{\lambda}{\sqrt{\epsilon_w'}} \frac{1}{\sqrt{1 - \left(\frac{f_{c10}}{f}\right)^2}} \quad (1)$$

where ϵ_w' is the real part of the DI water complex permittivity, λ is the free-space wavelength, $f_{c10} = c/(2a\sqrt{\epsilon_w'})$ is the TE_{10} mode cutoff frequency ($c = 3 \times 10^8 \text{ m/s}$), and $f = c/\lambda$ is the radiation frequency. The cooling chamber or water bolus is the portion between the waveguide and the muscle phantom as shown in Fig.1, isolates the waveguide cavity from the phantom. By circulating DI water through the cooling chamber, the antenna temperature was kept constant without any problem up to 100-W transmitted power levels.

III. METHODOLOGY

FEM has been extensively used in simulation of microwave antenna applicators because the computer simulations determine the optimal antenna design for the purpose of optimal power application and effectively localized thermal lesion. This method carries on the processing in three stages: Pre-processing, Processing and Post-processing. In Pre-processing the whole structure is divided into elementary sub domains, which are called finite elements and the field equations are applied to each of them. In processing the FE methods are essentially based on determination of the distribution of electric and magnetic fields in the structures under study, based on the solution of Maxwell's equations. The post processing performs analysis of results obtained by determination of other parameters based on the field distributions.

To solve electromagnetic problems with boundary values, it is essential to formulate the problem in terms of magnetic vector potential A and the electric scalar potential V .

$$B = \nabla \times A \quad (2)$$

$$E = -\nabla V - \frac{\partial A}{\partial t} \quad (3)$$

The electromagnetic field radiated in given tissue by an appropriate antenna can be determined by solving the Maxwell's equation stated below with the knowledge of tissue electromagnetic properties.

Discussing the most fundamental concepts of dielectric constant ϵ (relative dielectric constant) and the loss tangent $\tan \delta$, the dielectric constant tells a lot about reflection properties and reduction in wavelength whereas the loss tangent is reduced from the free-space value by the ratio c . Wave reflection at a surface of a material will increase as ϵ increases. For example, for a substance like water where $\epsilon \sim 60$, the power reflection is about 60%.

The loss tangent is a measure of how absorptive the material is, or how quickly it causes a wave to attenuate as it propagates in the material. A simple formula for penetration depth D , the depth at which the power density is reduced by $1/e^2$ or the fields by $1/e$, is

$$D = \frac{0.319\lambda_0}{\epsilon \tan \delta} = \frac{0.319\lambda}{\tan \delta} \quad (4)$$

where λ_0 is free space wavelength. The energy or power absorbed by the body as the wave attenuates produces local heating at a rate called the specific absorption rate, and is given by

$$\text{SAR} = \frac{1}{2\rho} \omega \epsilon_0 \tan \delta (E_{in})^2 \text{ w/kg} \quad (5)$$

Where,

SAR is the power transferred to the absorber by the electric field in the body

ρ is the mass density of the body, in kg/m^3

ϵ_0 is the permittivity of free space, in farads/m

ϵ is the relative dielectric constant $\tan \delta$ is the loss tangent

ω is the radian frequency given by $\omega = 2\pi f$

E_{in} is the electric field in volts/m at the point in in the body, with the subscript "in" to emphasize that the field inside the body is not the same as the field in the incident radiation.

A. Maxwell Equations

$$\nabla \times E = -j\omega\mu H \quad (6)$$

$$\nabla \times H = -j\omega\mu E \quad (7)$$

Where, E and H are Electric and magnetic field vectors respectively. μ and ϵ are permeability and permittivity respectively.

The total fields obtained due to electric and magnetic vector potential are obtained as:

$$E = -j\omega A - \frac{j}{w\mu\epsilon} \nabla(\nabla \cdot A) - \frac{1}{\epsilon} \nabla \times A \quad (8)$$

$$H = \frac{1}{\mu} \nabla \times A - j\omega V - \frac{j}{w\mu\epsilon \nabla(\nabla \cdot V)} \quad (9)$$

The position and orientation of the applicator are optimized in order to achieve better distribution of SAR (Specific Absorption Rate) in treated area. SAR determines the quantity of electromagnetic field exposition and it is related to E- field by [6].

The heat transfer inside a biological tissue is described by bio heat equation proposed by Penne's [13], which relates TE waves with temperature increase in tissue.

The bio heat equation is defined as:

$$\delta_{\tau} \rho C \frac{\partial T}{\partial \tau} + V(-k\nabla T) = \rho_b C_b \omega_b (T_b - T) + Q_{met} + Q_{ext} \quad (10)$$

where,

δ_{τ} is a time scaling coefficient

ρ is the tissue density

C is the specific heat of tissue

k is the tissue thermal conductivity tensor

ρ_b is the density of blood

C_b is the specific heat of blood

ω_b is the blood perfusion rate

T_b is the arterial blood temperature

Q_{met} is the heat source from metabolism

Q_{ext} is the specific heat source

The dimensions of the waveguide aperture are calculated using the following equations:

$$a = \frac{1}{2f_c \sqrt{\mu_0 \mu_r \epsilon_0 \epsilon_r}} \quad (11)$$

$$b = \frac{a}{2} \quad (12)$$

Where, a and b are the height and width of the waveguide aperture, f_c is the cut-off frequency,

μ_0 and μ_r are the permeability of free space and the material inside the waveguide respectively,

ϵ_0 and ϵ_r are the permittivity of free space and the material inside the waveguide respectively.

For calculating the waveguide height, it is necessary to know the wavelength, which is given by the equation:

$$\lambda = z = \frac{1}{2f_c \sqrt{\mu_0 \mu_r \epsilon_0 \epsilon_r}} \quad (13)$$

B. Model Formulation

Coaxial lines cross-coupled to a waveguide are commonly used power combining structures. The problem of a single coaxial line cross-coupled to a rectangular waveguide has been studied in [14,15]. In the problem considered in this paper, it is assumed that the waveguide and coaxial line are perfectly conducting and that only the TEM mode can propagate in the coaxial line at the frequency of interest. The analysis presented here is applicable to the case where the waveguide (assumed deionized water as dielectric) is perfectly matched at both waveguide ports.

FEM based COMSOL Multiphysics software has been used for the simulative implementation of applicator, which is a powerful interactive environment for modeling and solving all kinds of scientific and engineering problems based on partial differential equations (PDEs). With this software, one can extend conventional models for one type of physics into Multiphysics models that solve several coupled physics phenomena and do so simultaneously.

For more realistic implementation, 3D model has been created which consisted of a waveguide applicator, water bolus and the muscle phantom.

The wave excitation with $P_{in} = 100$ W is applied at the coaxial coupled input of the waveguide. Matched boundary conditions are selected for water bolus and muscle phantom. After defining the distinguish boundary conditions, meshing is performed with mesh statistics as detailed in Table 1. Meshing can be contemplated as the splitting of domains into smaller sub-domains made up of geometric primitives like hexahedra and tetrahedral in case of 3D and quadrilaterals and triangles in case of 2D. When trying to solve a problem that does not converge, or having solved a problem where the solution does not seem properly resolved, there is often a need for re-solving the problem using a finer mesh. In these situations it is sometimes easier to refine the existing mesh than to regenerate the mesh from scratch using finer mesh settings. Adaptive meshing automatically refines, coarsens or relocates and adjusts the basis to achieve a solution having a specified accuracy in an optimal fashion. The higher order adaptive mesh refinement technique has been implemented using Lagrange –Quadratic type element for meshing the geometry. The final mesh obtained after iterative refinement is as shown in Fig.2. Table 2 lists the scalar variables used in the problem formulation with their expression and descriptions.

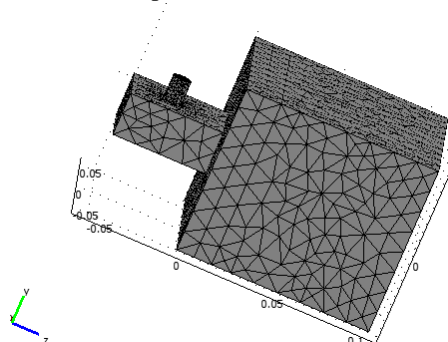


Fig.2. Generated mesh

The mesh consisted of 16377 elements and 109472 numbers of degrees of freedom. The governing equations of the physics are then solved inside each of these sub-domains. The parametric solver is used to solve the model. The execution has been performed using Intel(R) Core(TM) i7-3770k @3.40 GHz 3.90GHz CPU with simulation time of 1088.97 s.

Table 1: Mesh statistics of the generated mesh

Number of degrees of freedom	109472
Number of mesh points	3346
Number of elements	16377
Tetrahedral	16377
Prism	0
Hexahedral	0
Number of boundary elements	2998
Triangular	2998
Quadrilateral	0
Number of edge elements	343
Number of vertex elements	40
Minimum element quality	0.346
Element volume ratio	0

Table 2: Scalar variables

Name	Expression	Description
eps_w	81	water permittivity
sigma_w	.05[S/m]	water conductivity
eps_m	54	muscles phantom permittivity
sigma_m	0.8[S/m]	muscle phantom conductivity
rho_m	1000[kg/m^3]	muscle phantom density
eps_ptfe	2.2	
sigma_ptfe	0	

C. GMRES Linear System Solver

This linear system solver uses the restarted GMRES (generalized minimum residual) method. This is an iterative method for general linear systems of the form $Ax = b$. For fast convergence it is important to use an appropriate preconditioner.

There is no guarantee that a restarted GMRES converges for a small restart value. A larger restart value increases the robustness of the interactive procedure, but it also increases memory use and computational time. For large problems, the computational cost is often very large to produce a preconditioner of such a high quality that the termination criteria are fulfilled for a small number of iterations and for a small restart value. For those problems it is often advantageous to set up a preconditioner with a somewhat lesser quality and instead increase the restart value or iterate more steps. Doing so typically increases the condition number for the preconditioned system, so an increase in the error-estimate factor might be needed as well. Two slightly different versions of GMRES are available in COMSOL Multiphysics. The difference between these two versions is whether left or right

preconditioning is used. Normally, the two versions of GMRES have similar convergence behavior.

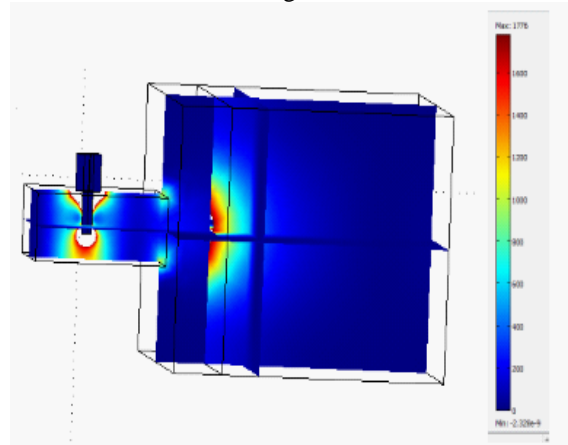


Fig.3. SAR heat distribution in the muscle phantom

IV. RESULTS AND DISCUSSION

The performance analysis is done by evaluating 3D heat distribution and reflection coefficient S_{11} parameters. Fig.3 show the 3D heat distribution within the muscle phantom depicting spherical lesion and localized heating pattern. Fig.4 further plots the values of S_{11} versus frequency to analyze the impedance matching for the different cases. It is evident from the plots that at operating frequency of 485 MHz, S_{11} reaches a minimum of -15.6 dB indicating good impedance matching for waveguide loaded with deionized water having relative dielectric constant, $\epsilon_r = 81$. Fig. 5 illustrates the convergence plot obtained during simulation of the problem using GMRES (Generalized minimal residual) method. Fig.6. depicts the resistive heating plot which gives the insight about the amount of power converted into heat at various penetration depths inside the tissue. The maximum power absorption is achieved and the focusing of heat distribution in the muscle phantom is therefore best localized. Also for commercial point of view water as dielectric is a better and cheaper option. As the permittivity of the material filled in the antenna applicator increases the optimized performance is obtained with reduced dimensions of the waveguide applicator.

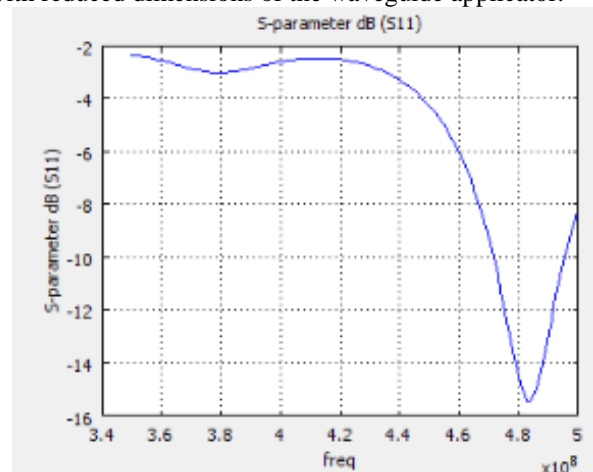


Fig.4. Impedance matching (S_{11}) plot of coaxial to waveguide coupled applicator.

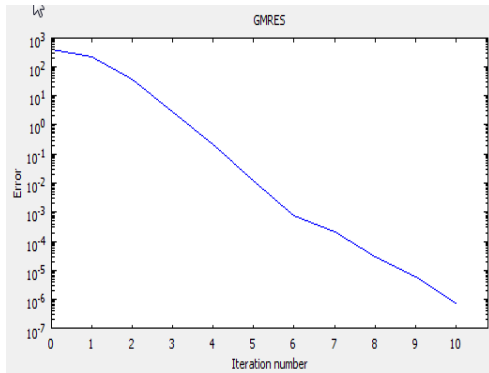


Fig.5. Convergence plot while solving the problem using GMRES (Generalized minimal residual) method.

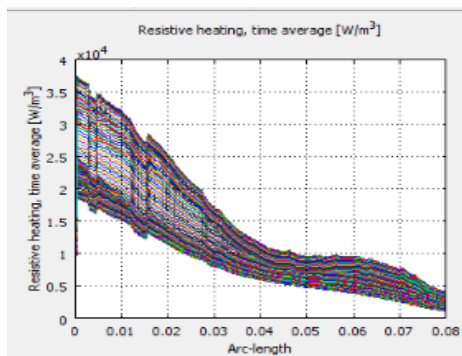


Fig.6. Resistive heating plot

V. CONCLUSIONS

The 3D simulation model of water-stowed, indirectly cooled coaxial coupled waveguide applicator waveguide for local hyperthermia treatment has been done using FEM based COMSOL Multiphysics ver. 3.5a. Adaptive Finite element full-wave modeling has been performed to predict to a high degree of accuracy the performance of 3D model of the applicator. Good thermal and electrical stability is achieved for radiation powers up to 100 W. The work also performs the analysis of field distribution in terms of resistive heating distribution. In the next step of research the formulation of 3D model will be extended for analysis of various favorable antenna types connected with the coaxial probe inside the waveguide to achieve better temperature goals and more spherical lesion inside the tissue.

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