Computation of Electromagnetic Field Inside a Tissue at Mobile Communications Frequencies

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***Abstract*—The increasing diffusion of mobile communications is stimulating the study of the interaction mechanisms between electromagnetic (EM) fields and biological systems at radio frequencies. This paper is devoted to the modeling of the interaction between EM fields and a tissue, represented with spherical cells. Different EM approaches have been used to analyze the problem and, in particular, the lumped-element finite-difference time-domain (FDTD) technique has been used to model the cell's membrane represented with the Hodgkin–Huxley (HH) model, and the Floquet theorem to study a tissue by analyzing only few cells. The EM problem has been solved by using the FDTD technique to be independent from the geometry. Computations have been performed at GSM900 and GSM1800 frequencies overcoming the problem relevant to the computation time by using the quasi-static FDTD technique. The results show the field distribution inside the tissue at global system for mobile communications frequencies; the effect of the application of the HH model to the membrane is also shown.**

***Index Terms*—Floquet, global system for mobile communications (GSM), Hodgkin-Huxley (HH) model, lumped-element finite difference time domain (LE-FDTD), quasi-static finite difference time domain (FDTD).**

I. Introduction

THE increasing diffusion of mobile communications has created great interest in the interaction between electromagnetic (EM) fields and biological tissues. The need to establish precise safety standards for regulating human exposure to EM waves has led to the development of research streams—both in modeling and experiments—in order to describe the different aspects of the problem, at various levels of definition such as at tissue level, cell level, and ionic level.

At a tissue level, a biological body is treated as a lossy medium and, as such, ascribes the EM properties to each tissue making up the body. This is the usual way to represent a tissue in dosimetry; even if the spatial definition is very high (even less than 1 mm), the interaction between cells and the EM field is not taken into account: the biological body is simply seen as a scatterer. In this area, the most frequently used technique for computing the EM field is the finite-difference time-domain (FDTD) method [1], [2]. Quite a few papers have been published, especially related to the radiation of cellular phones inside the human head (see [3]–[5]).

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At a microscopic level, several models have been developed describing the behavior of the cell membrane with an equivalent circuit that takes into account the complex mechanisms that govern the interaction of a cell with the surrounding environment. The first historic paper in this area was by Hodgkin and Huxley (HH) [6], who established the homonymous equivalent circuit for the membrane. Over the years, several additional contributions have been produced (see [7]–[9]). In [9], a model is shown that creates a link between the electrical circuit of the membrane and a finer model that describes the operational mechanisms of an ionic channel, which is represented as a finite state's Markov's machine (see also [10] and [11]).

At a more microscopic level, it is possible to study the effect of EM fields on the interaction between a ligand ion and a receptor site. These studies are based on the quantum Zeeman–Stark model (for further details, see [12]). The goal of these techniques is to analyze whether there is a modification in the probability of binding an ion to a receptor site in the presence of an exogenous EM field.

As can be noted, all the previous methods consider only one aspect of the problem, i.e., the interaction of the cells with the EM field is not analyzed at the tissue level, at the cell/sub-cell level, the presence of nearby cells and the distribution of the field inside the tissue are not taken into account.

This paper presents a new approach to calculate the EM field inside a tissue, composed of electrically excitable cells, by means of the FDTD method. For the sake of simplicity, the tissue has been represented with spherical cells, but the method can be applied to more general cases. The tissue has been discretized with a 1-μm step. In each cell pertaining to the membrane, the HH model has been implemented in order to include the membrane effects. To represent the presence of the remaining cells, the tissue has been represented as a stack of cells applying the Floquet theorem on the border of the structure. This approach makes it possible to limit the computational effort to the analysis of a small part of the structure. Cellular tissues are not perfectly periodic and the living cells are not precisely spheres. Anyway, aware of those limitations, the adopted approximations have been considered proper.

The space discretization step used in the simulations is very small compared to the wavelength of the excitation signals (GSM900 and GSM1800 bands). Consequently the standard FDTD requires a very short time step and, thus, an unaffordable computation time. The problem has been overcome by adopting a quasi-static FDTD approach, which makes it possible to compute (using sinusoidal excitations) the steady-state value of the EM field with only a few iterations.

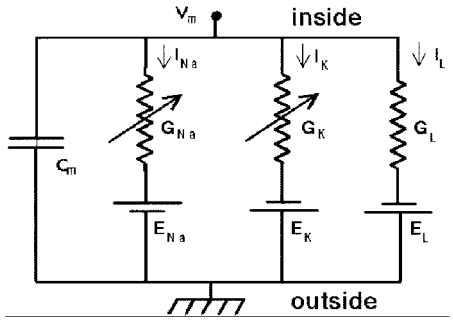


Fig. 1. Equivalent circuit of the cell's membrane developed by HH [6].

II. Cell Model

The plasmatic membrane defines the entity of the cell and maintains the essential differences between its interior and the external environment where the cell lives. However, it is not a simple barrier. It is also a very selective filter, which holds different concentrations of ions on the inside and outside.

The dimensions of a cell are a few tens of micrometers, depending on the type of tissue, and the thickness of the membrane is a few nanometers. Depending on the type of cell, a voltage of 20–200 mV is established across the membrane. The current (flux of ions) across the membrane, on average zero when the cell is in a resting state, depends on the variations of the membrane voltage (for further information, see [13]).

The first model describing the current flowing through the membrane with respect to the membrane voltage was introduced by HH in 1952 [6] for the giant squid axon. The equivalent circuit of the membrane is shown in Fig. 1. It consists of nonlinear conductances, voltage generators, and a capacitor that takes into account the distribution of the charges at either side of the membrane.

The various elements have the following meaning.

- C_m is the membrane capacitance.
- G_k, G_{Na} are the nonlinear conductances of the potassium and sodium channels, taking into account the flux of the potassium and sodium ions across the membrane.
- G_L is the leakage conductance; it does not correspond to any physical channel, but takes into account the flux of other ions across the membrane.
- V_m is the voltage across the membrane.
- E_k, E_{Na}, and E_L are the resting potentials of the various channels. They contribute to the steady-state voltage on the membrane for each channel.
- I is the current flowing across the membrane due to the flux of the ions. It is zero if the membrane voltage equals the resting potential.

The equations governing the model are [6]

$$I = C_m \frac{dV_m}{dt} + G_k n^4[V,t](V_m - E_k) + G_{Na} m^3[V,t]h\,[V,t](V_m - E_{Na}) + G_L(V_m - E_L) \quad (1)$$

$$\frac{d\Psi[V,t]}{dt} = a_\Psi(V)\big(1 - \Psi[V,t]\big) - b_\Psi(V)\cdot\Psi[V,t] \quad (2)$$

where

$$V = V_m - V_r \quad (3)$$

V_r is the resting potential of the membrane and Ψ stands for n, m, and h, which are a-dimensional parameters determining the percentage of open channels (an ion passes through the membrane). The percentage of the open Potassium channels is n^4, the percentage of open sodium channels is m^3h. Quantities appearing in (2) are defined as [6]

$$a_n(\mathrm{V}) = \frac{0.01(V+10)}{e^{((V+10)/10)} - 1} \quad (4)$$

$$a_m(\mathrm{V}) = \frac{0.1(\mathrm{V}+25)}{e^{((\mathrm{V}+25)/10)} - 1} \quad (5)$$

$$a_h(\mathrm{V}) = 0.07\, e^{(\mathrm{V}/20)} \quad (6)$$

$$b_n(\mathrm{V}) = 0.125\, e^{(\mathrm{V}/80)} \quad (7)$$

$$b_m(\mathrm{V}) = 4\, e^{((\mathrm{V}+65)/18)} \quad (8)$$

$$b_h(\mathrm{V}) = \frac{1}{e^{((\mathrm{V}+30)/10)} + 1}. \quad (9)$$

It should be specified that the voltage V is expressed in millivolts, as originally considered in the HH model [6].

It has been assumed that the HH model is valid for cells having excitable membranes. Table I quotes the values of the elements in the HH model, as reported in [6].

As shown by (4)–(8), the coefficients a_Ψ and b_Ψ in (2) depend on the voltage V. Assuming, however, that these coefficients are constant with the voltage in a limited interval of variation (as will be shown later) and rearranging (2) as

$$\frac{d\Psi}{dt} = -(a_\Psi + b_\Psi)\Psi + a_\Psi \quad (10)$$

we can find the expression for Ψ

$$\Psi(t) = \frac{a_\Psi}{a_\Psi + b_\Psi}\left(1 - \frac{e^{-(a_\Psi + b_\Psi)t}}{K}\right) \quad (11)$$

where K is a constant defined by the initial conditions. Considering the membrane at equilibrium ($\mathrm{V} = 0$) by (4) to (9), it follows that a_Ψ and b_Ψ are positive numbers, thus,

$$\lim_{t\to+\infty} \frac{a_\Psi}{a_\Psi + b_\Psi}\left(1 - \frac{e^{-(a_\Psi + b_\Psi)t}}{K}\right) = \frac{a_\Psi}{a_\Psi + b_\Psi}. \quad (12)$$

If the membrane has not been stimulated for a long time, much greater than the time constant $\tau = 1/(a_\Psi + b_\Psi)$, the parameter Ψ can be considered a constant as follows:

$$\Psi(t)\big|_{t>>\tau} \cong \frac{a_\Psi}{a_\Psi + b_\Psi}. \quad (13)$$

Fig. 2 shows the percentage variations of the quantities n, m, and h as a function of the voltage displacement with respect to the resting potential in the range of $\pm 1\,\mu$V (this variation is big enough to consider the signal range variations we will analyze). If we consider a sinusoidal plane wave (1-V/m peak value) impinging on a cell's membrane, being the transversal

TABLE I
VALUES OF THE PARAMETERS IN (1)

C_M	E_K	E_{Na}	E_L	G_k	G_{Na}	G_L
F/m^2	MV	mV	mV	S/m^2	S/m^2	S/m^2
0.01	72	-55	50	360	1200	30

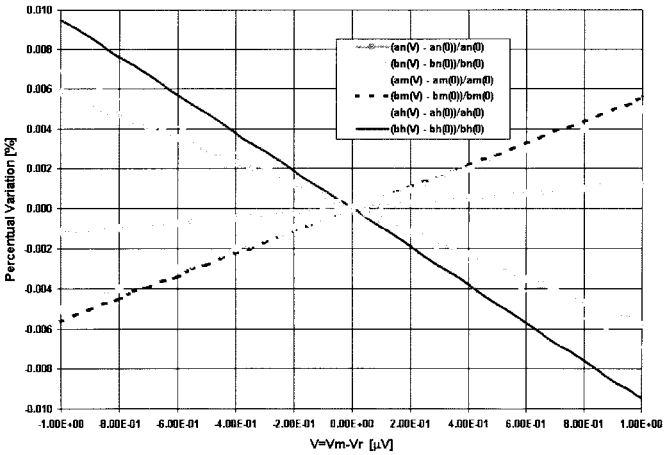


Fig. 2. Percentage variations of a_Ψ and $b_\Psi(\Psi = n, m, h)$ versus voltage.

TABLE II
STEADY-STATE VALUES FOR THE PARAMETERS n, m, AND h DRAWN FROM (3)

n	m	h
0.3177	0.0529	0.5961

dimension of the membrane of a few nanometers, a variation of the membrane potential can be estimated as a few nanovolts. This justifies the assumption that the parameters described by (6)–(11) can be approximated as constant; considering the discussion above, the quantities n, m, and h can then be considered fixed at the steady-state value. This leads to the conclusion that the HH model in this context can be considered linear. The values of the quantities m, n, and h in the steady state are reported in Table II.

III. IMPLEMENTATION

This section is devoted to the discussion of the methodology used to analyze a tissue radiated by an EM field. The FDTD technique has been chosen as the EM solver. In this way, the solution can be computed independently from the geometry of the problem. The methodology can be subdivided into the following three parts:

1) computation of the EM solution in a reasonable period of computational time;
2) representation of the membrane by means of the HH model and the interaction with the incident field;
3) modelization of a tissue by using only a few cells.

The diameter of a cell is a few tens of micrometers, thus, we have used a space discretization step of 1 μm. In this case, the standard FDTD, when cubic cells are used, requires a time step of

$$\Delta t \leq \frac{\Delta s}{v_o\sqrt{3}} \approx 10^{-15}[s] \tag{14}$$

where v_o is the speed of light in free space.

Since the signals of the mobile communication systems are located at 900- and 1800-MHz bands, the simulation of one single period of the incident wave would involve some millions of iterations, i.e., a full time-domain simulation is unpractical. Nevertheless, since the HH model and, thus, the entire system, can be considered linear, a quasi-static FDTD approach, well described in [15], can be adopted, which allows the EM solution to be computed in a few thousands of iterations.

It is well known that the lumped-element finite-difference time-domain (LE-FDTD) method incorporates lumped elements (linear and nonlinear) in the standard FDTD approach [14]. Since the HH model is basically made up of lumped elements, the LE-FDTD method can be used to represent the membrane in the FDTD space. After the discretization process of the space has been performed, each cell pertaining to the membrane contains a great number of channels. Consequently, even if each channel is described by a statistical law (see [11]), by exploiting the law of large numbers [17], the group of channels belonging to one FDTD cell can be described with the HH model, which is a deterministic one. For instance, if we choose a discretization that is so fine that only a few channels belong to the FDTD cell, we would not be able to use the HH model, but we would have to use a statistical model (such as the finite state's Markov's machine).

In this study, the cell is characterized as a sphere whose surface, representing the membrane, is modeled with the LE-FDTD method, while the remaining space is modeled by assigning the permittivity and conductivity. The transmembrane voltage is orthogonal to the membrane itself, hence, the HH model has to be included in the FDTD grid so that it is excited by the electric field normal to the surface of the membrane. The LE-FDTD method has consequently been modified in order to consider surface lumped elements whose normal is not aligned with the FDTD grid (see Fig. 3). This has been implemented by changing the local reference system for the field associated to the surface element, for each time step, as follows:

- transformation of the field derived from the FDTD method in a local coordinate system, associated with the surface element; one of the three unit vectors is chosen normal to the surface element;
- field components updating (on the membrane) by means of the LE-FDTD method;
- back transformation of the field components derived from the local coordinate system associated with the surface element to the original FDTD coordinate system.

The simulation of a complete tissue is not realistic because of the huge requirement of computer resources. On the other hand, the simulation of a single cell does not represent the real situation because there is a variation of the field distribution in the

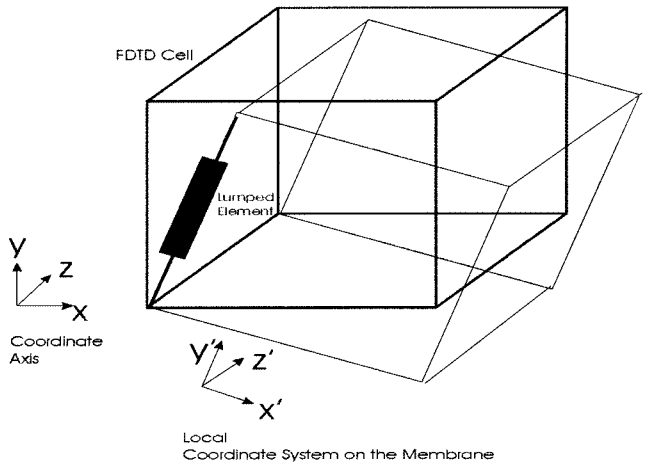


Fig. 3. Position of the lumped element inside a membrane's cell.

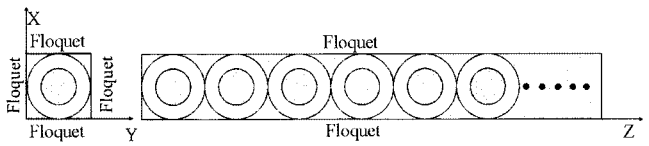
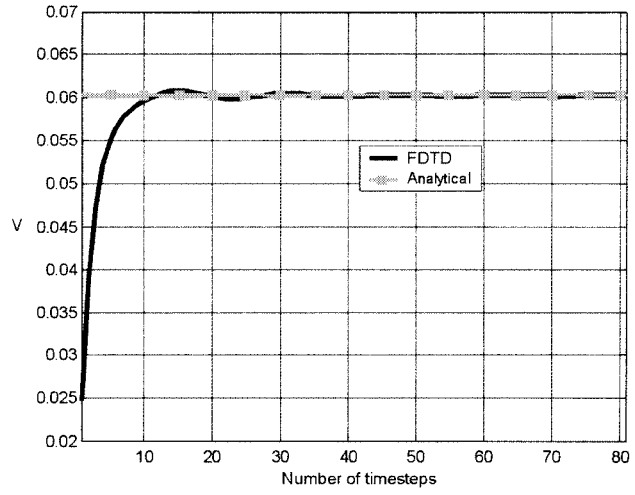


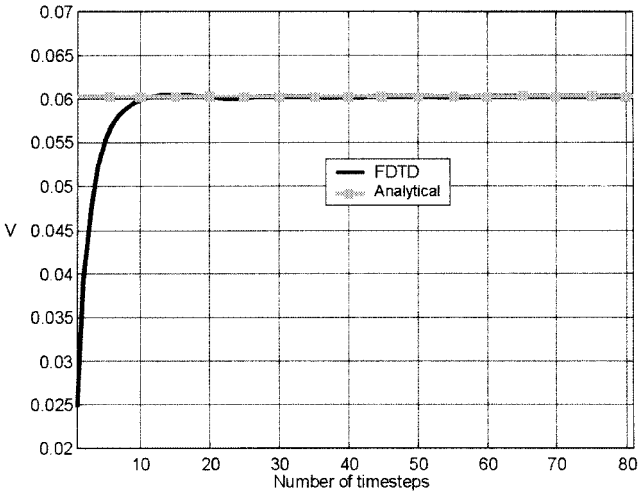
Fig. 4. Principle scheme of the simulated structure.

space due to the improper boundary conditions. This problem has been solved by applying the Floquet theorem [16] to a stack of cells. The stack takes into account the penetration of the field within the structure, while the application of the Floquet theorem on the four sides of the stack substitutes the structure of the entire tissue, as shown in Fig. 4. The use of Floquet boundary conditions may exhibit resonances due to the Floquet modes. Anyway, due to the small (some microns) transversal dimensions of the structure, the possible resonant behaviors, associated to Floquet boundary conditions, have characteristic frequencies of some orders higher than global system for mobile communications (GSM) frequencies and, thus, they are supposed to be out of our observation's frequencies.

To study EM properties of dense media, in other fields of analysis, such as remote sensing of snow and scattering from sea ice [18], [19], some authors have used random models along with Monte Carlo simulations. Even if some similitudes can be established between the EM simulation of a tissue and, for instance, the analysis of the scattering properties of the sea ice, the Monte Carlo method is not suitable to our purpose. In fact, this technique requires a lot of simulations (in [18], 40 every structure) to extract the statistical properties of the chosen structure. In our case, every simulation lasted 24–48 h on a 16-processor SGI Origin 2000 computer, thus, a complete Monte Carlo simulation would last up to 80 days, which is an unmanageable computational time. Furthermore, the Monte Carlo method is not suitable to be coupled with Floquet boundary conditions; hence, applying that to our case, it would force us to analyze not a simple stack of cells, but an entire portion of the tissue, making the simulation much heavier (in this case, every single simulation —considering, for example, a 3-mm^2 portion of a tissue, which is a structure 10 000 times larger that our stack of cells surrounded by Floquet boundary conditions—does not last 24–48 h, but ∼10 000 times more, also requiring an unrealistic quantity of memory).



(a)



(b)

Fig. 5. Time evolution of the voltage on the membrane without any external excitation. (a) The normal is directed along X. (b) The normal is $\hat{n} = \frac{1}{\sqrt{2}}\hat{z} + \frac{1}{\sqrt{2}}\hat{y}$.

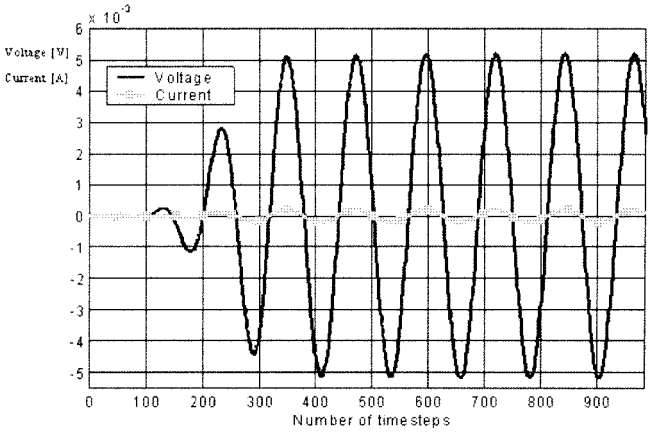


Fig. 6. Voltage and current on the resistor.

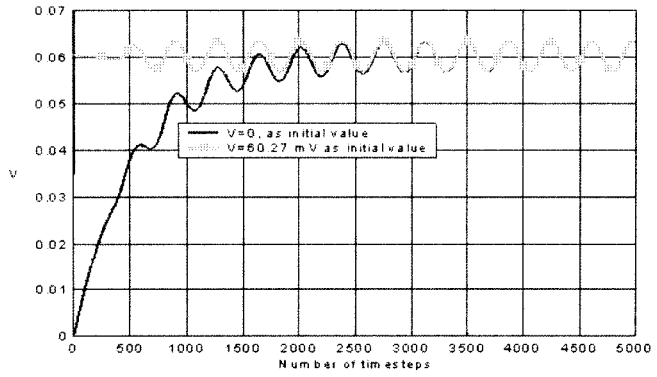


Fig. 7. Comparison between membrane voltages with different initial conditions.

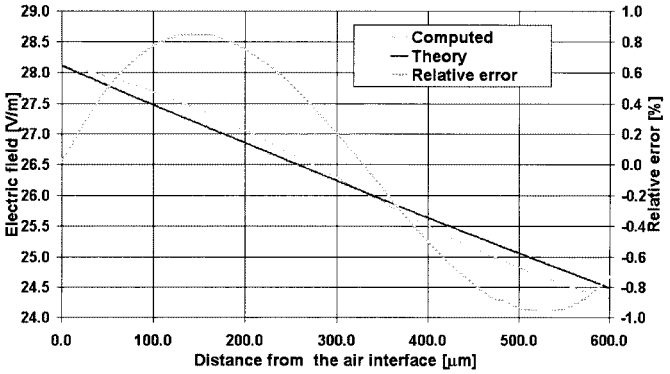


Fig. 8. Electric field inside the lossy medium. $\epsilon_r = 1.0, \mu = 25.0$ S/m. The space discretization step is 1 μm.

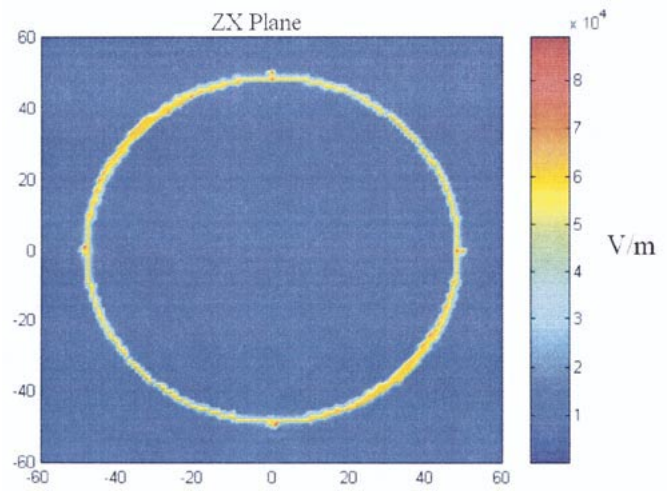


Fig. 9. Field distribution on a coordinate plane.

IV. Validation

A. HH Model Implementation

The validation of the implementation of the HH model within the FDTD framework has been performed by comparing analytically computed solutions for test cases.

In the first example, the HH model has been enclosed in a single FDTD cell. The system evolves without any excitation until the steady state is reached. A 10-cm-cell side has been chosen (this assumption does not affect the validity of the tests since the entire system is linear) in order to observe both the

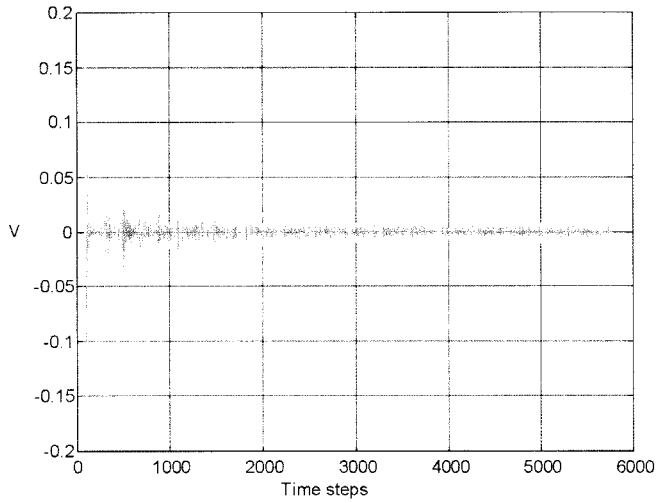


Fig. 10. Voltage versus time in the central FDTD cell of the structure.

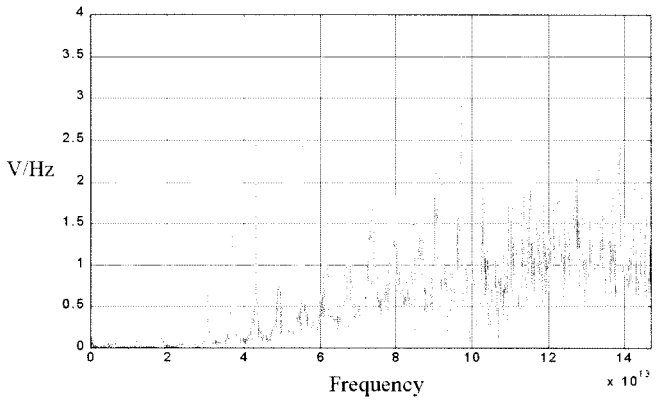


Fig. 11. Spectrum of the electric field at the center of the structure.

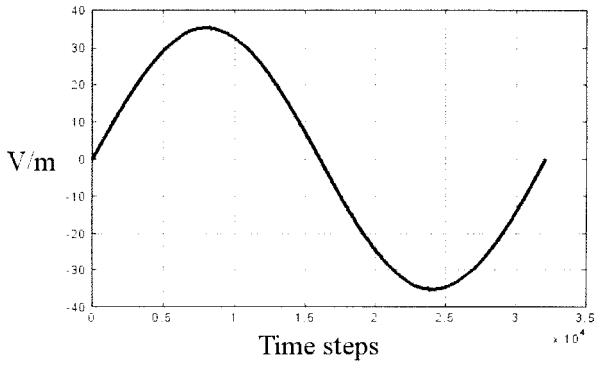


Fig. 12. Signal recovered from the numerical noise.

TABLE III
EM Features of the Simulated Structure

	Conductivity [S/m]	Permittivity
Water @ 900 MHz	0.173	78.24
Water @ 1800 MHz	0.66	77.59
Membrane	$13 \cdot 10^{-9}$	6.0

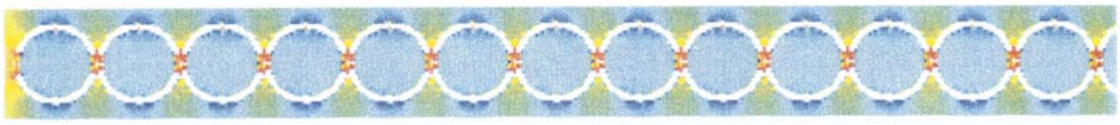


Fig. 13. 900 MHz, modulus of the electric field on the zx-plane (logarithmic scale).

transient of the phenomenon and the steady state with a reasonable period of computation time. By placing $I = 0$, which is the steady-state condition for the cell (the average rate of ions crossing the membrane is zero), in (2), and by considering the steady state, the voltage can be analytically computed and is given by (using the data given in Table II)

$$\mathrm{V}_r = \frac{G_K n^4 E_K + G_{Na} m^3 h E_{Na} + G_L E_L}{G_K n^4 + G_{Na} m^3 h + G_L} = 60.27\,\mathrm{mV} \tag{15}$$

Fig. 5 shows the comparison of the FDTD and analytically computed membrane voltage versus time for a cell whose membrane is aligned with the grid [see Fig. 5(a)] and for a cell membrane whose normal forms an angle of 45° with the grid [see Fig. 5(b)]. As can be observed, the numerical and analytical results correspond.

Another test has been performed by considering zero all the parameters in (2), except for $n = 1$ and $g_k = 1\,000\,000\,\mathrm{S/m^2}$ so that the network behaves as a resistor whose value is

$$R = \frac{1}{g_k \Delta s^2} = 25\,\Omega. \tag{16}$$

In this case, a plane-wave excitation has been used (E-field along X, propagation direction Z, $f = 30\,\mathrm{GHz}$, $\Delta s = 0.2\,\mathrm{mm}$). Fig. 6 shows the voltage and current on the membrane. The ratio between the peak value of the voltage and peak value of the current is 25.005 Ω.

One important aspect to be considered is the transient due to the time constant of the HH circuit; this transient can actually be very long (in terms of time steps) when considering simulation at cell's dimensions; consequently, to speed up the computational time, the value obtained at the steady state without any excitation can be considered as the initial value of the membrane voltage. To prove that this method gives the same results as the complete transient simulation, the HH model has been implemented in a grid cell (dimension of 10 cm) and it has been excited with a plane wave at the frequency of 10 MHz, considering firstly the membrane voltage $= 0$ and secondly $= 60.27\,\mathrm{mV}$. Fig. 7 shows the comparison of the two methods leading to the conclusion that the computation can be speeded up by considering the resting potential on the membrane of the initial state.

B. Floquet Boundary Conditions' Implementation

To validate the correctness of Floquet boundary conditions and the quasi-static FDTD method, the penetration of a plane wave in a half-space has been computed. A parallelepiped of lossy medium $(\epsilon_r = 1.0, \sigma = 25.0\,\mathrm{S/m})$ has been radiated by a plane wave normally incident at 900 MHz. On the four sides of the parallelepiped, the Floquet theorem has been applied to simulate the half-space. On the other two sides of the parallelepiped,

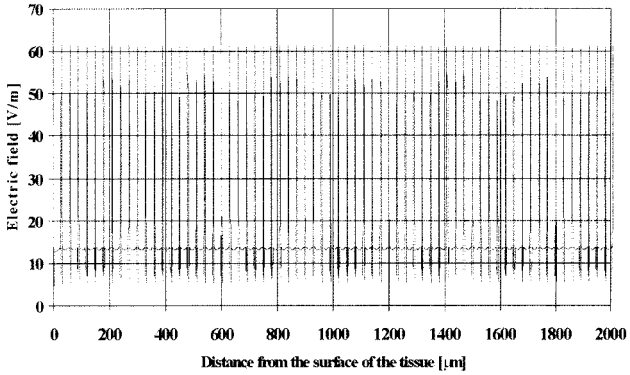


Fig. 14. 900 MHz, penetration of the electric field along the z-axis, through the center of the spheres.

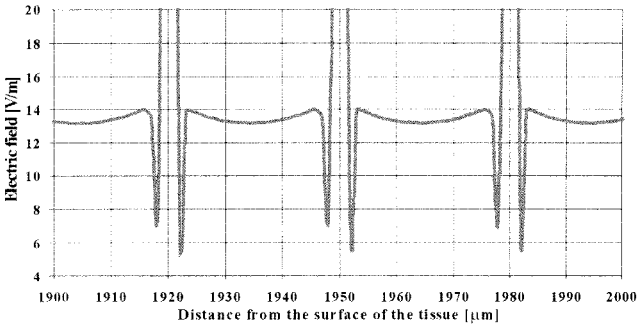


Fig. 15. 900 MHz, penetration of the electric field (enlargement).

the Mur absorbing boundary condition (ABC) have been applied to truncate the grid. The field in the grid has been determined by means of the quasi-static FDTD method. The simulated electric field inside the parallelepiped has been compared with the analytical solution. Fig. 8 shows the computed and analytical E-field and the relative error. As can be noted, a good agreement has been obtained.

C. Single Cell Simulation

A spherical structure, with a diameter of 50 μm has been discretized with a 1-μm step. All the cells pertaining to the spherical surface (approximately 30 000) include the HH model, while the volume inside the sphere has been considered a vacuum. This model has been implemented to check the stability of the procedure and the polarization value to 60.27 mV of the voltage on the membrane for a complex structure. Fig. 9 shows the magnitude of the electric field in a coordinate plane of the structure without any external excitation, while Fig. 10 shows the electric field in the central discretization cell of the structure.

It can be observed from Fig. 10 that the field in the central FDTD cell of the structure is not zero, showing a transient, and that there is a higher field level on the coordinate axis and on the

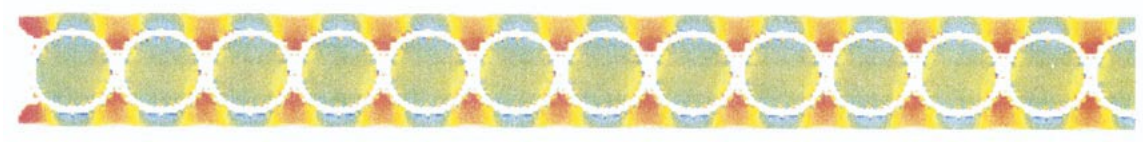


Fig. 16. 900 MHz, modulus of the electric field on the zx-plane (logarithmic scale).

directions $\pm 45^{\circ}$, as shown in Fig. 9. This effect can be ascribed to the numerical noise created by the geometry of the structure.

To support this hypothesis, a Fourier transform has been applied to the time samples of the electric field. Fig. 11 shows the spectrum of the noise showing peaks repeated every 6000 GHz corresponding to a wavelength $\lambda = 50\,\mu$m, the diameter of the simulated structure.

To verify whether the noise is random or deterministic, a simulation has been carried out radiating the structure with a plane wave at 1 GHz. The field in the central grid cell has then been computed as the difference of the situation with and without the excitation. Fig. 12 reports the result, showing that the noise is deterministic and that the harmonic excitation can be retrieved.

V. Results

The structure presented above does not represent a real situation; in fact, we have only one cell surrounded by Mur ABCs. This simulation has the following drawbacks:

- adopted boundary conditions do not account for the presence of a tissue around the analyzed cell;
- one cell simulation is unable to show the penetration curve of the electric field inside a tissue

To overcome these shortcomings, a more suitable simulation has, therefore, been set up. The new structure (see Fig. 4) consists of a stack of spheres with a radius of 15 μm (100 spheres in the case of 900-MHz simulations, 30 spheres in the case of 1800-MHz simulations) terminated by the Mur ABC. The application of the Floquet theorem [16] (hence, the simulation of a periodic structure along two dimensions) allows us to study a tissue by representing it with a small number of cells that is more suitable for the simulations.

Finally, the application of the quasi-static method [5] allows us to obtain the EM solution with a small number of time steps

Simulations have been carried out by using spheres to represent the cells with the same features (conductivity and dielectric constant) of the water at the analyzed frequency. Each cell has been enclosed within a spherical shell, representing the membrane that is simulated by the HH model. The values of the conductivity and dielectric constant of the water at the specified frequencies has been measured at the Specific Absorption Rate (SAR) Laboratory, Telecom Italia Laboratory, Turin, Italy. The values of the constants used for the membrane are in [20].

The structure has been excited by a plane wave propagating along the Z-axis, E-field along X, with an amplitude of 100 V/m. Table III shows the characteristics of the materials employed in the simulations.

Figs. 13–15 show the electric field inside the simulated tissue at the frequency of 900 MHz; from the inspection of Fig. 14 in particular, we can see that the field inside the cells is not constant.

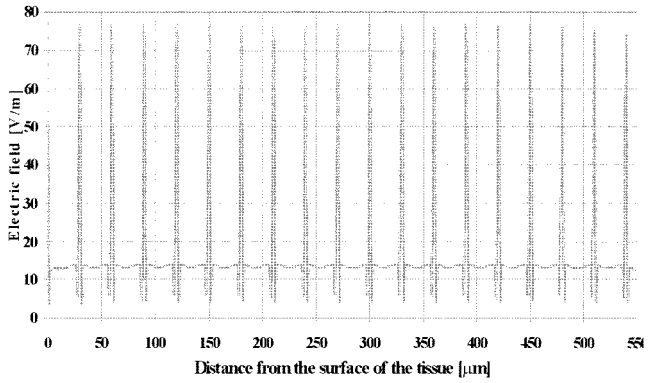


Fig. 17. 1800 MHz, penetration of the electric field along the Z-axis.

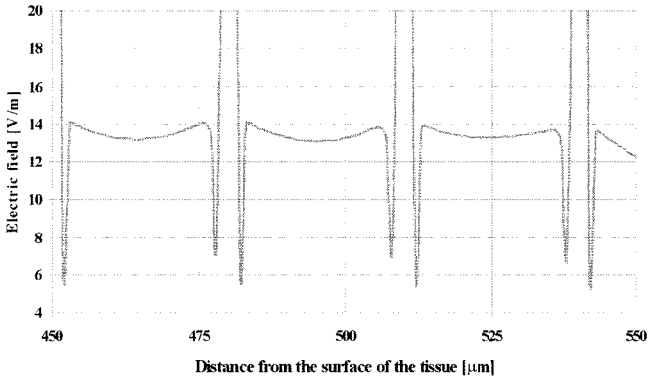


Fig. 18. 1800 MHz, penetration of the electric field (enlargement).

Figs. 16–18 show the results from the simulation at 1800 MHz. In Fig. 17. The effect of the field decay due to the penetration inside the tissue, more evident at 1800 MHz than at 900 MHz can be noted.

The effect of the polarization of the incident wave has been analyzed by applying an incident wave with the electric field rotated by 45° with respect to the X-axis. As can be noted in Fig. 19, there is a little difference in the internal field with respect to the previous case.

Finally, the effect of the application of the HH model to the membrane has been analyzed. As can be noted in Fig. 20, a difference of approximately 30% can be observed in the field due to the presence of the HH model to represent the membrane.

A. Real Signals' Simulations

Another aspect tackled in this paper is the simulation of real signals, such as the signals of GSM900 and GSM1800. For this purpose, we have focused our attention on one time slot of the GSM signal, which is phase modulated, with constant envelope, and we can mathematically represent it as

$$s(t) = A\sin\big(\omega t + \varphi(t)\big). \tag{17}$$

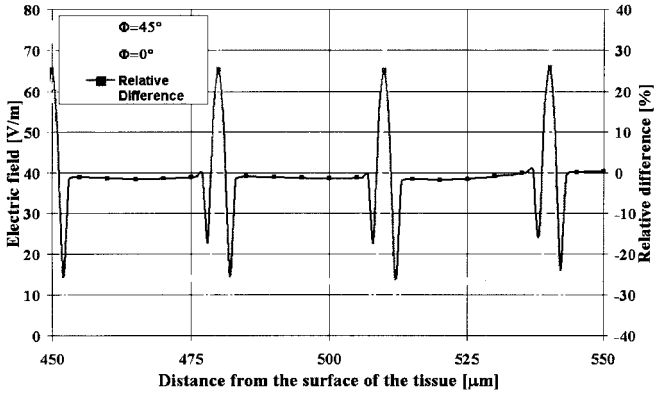


Fig. 19. 1800 MHz, penetration of the electric field along the Z-axis (polarization 45°, enlargement).

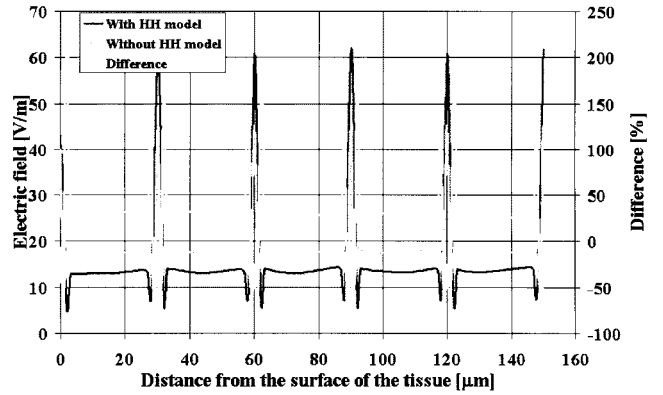


Fig. 20. 1800 MHz, comparison between simulations.

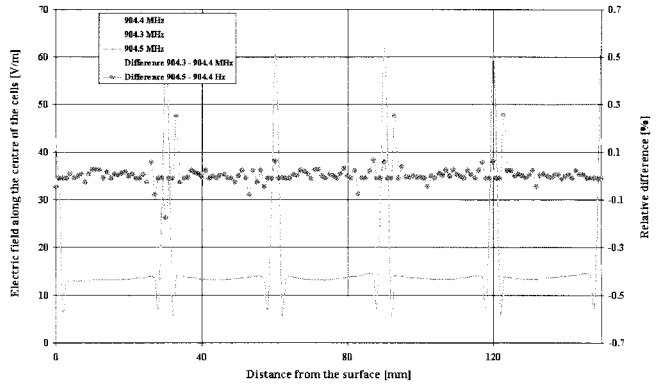


Fig. 21. Comparison of electric field at three different frequencies inside the structure.

The frequency in (17) depends on the GSM channel used by the signal and $\varphi(t)$ contains the digital information. If we consider, in a first approximation, that this function is linear at intervals, the associated frequency deviation (related to the instantaneous phase $\varphi(t)$ with $\Delta f(t) = 1/(2\pi) \quad (d\varphi(t))/(dt)$) is clearly constant at intervals.

The GSM signal with that approximation can, therefore, be seen as an ordinate sequence of sinusoids with different frequencies. By analyzing the behavior of the signal, we have seen that, in a GSM frame, we can observe 64 different frequencies, variously combined to form a sequence of 592 frequencies that constitute a GSM time slot.

In order to investigate whether the GSM signal can be approximated by a pure tone at the center frequency of the channel, we have estimated the electric field along the centers of the stack of cells that constitute the simulation domain at three frequencies (904.3, 904.5, 904.4 MHz), which represent the center and side frequencies of a GSM900 channel (geometric and EM features of the structure are discussed in Section V). From Fig. 21, we can see that the differences between the three behaviors at the three different frequencies are negligible. We have, therefore, retained that the effects of the GSM signal can be correctly simulated by studying only one single-tone signal.

VI. Conclusions

The sketch of the analysis methods in the field of bioelectromagnetism consists of several approaches that investigate the phenomenon of the interaction between EM fields and biological systems at various level of spatial definitions. In the depicted scenario, there are no methods able to conjugate a very accurate EM analysis of the tissues while taking into account the presence of interaction mechanisms.

In this study, we tackled the problem of analyzing the EM field inside a tissue at the GSM frequencies, proposing a new approach that combines the high accuracy in EM analysis of the FDTD method. The inclusion of the HH model of the cellular membrane and some procedures (Floquet theorem and quasi-static method) allow the simulation of micrometric structures and, hence, are smaller than the wavelengths of the GSM signals. Furthermore, the proposed method is completely general since it allows cells of arbitrary geometries to be handled and to embed all types of lumped elements models for the membrane. The simulations shown in this paper have been carried out for tissues made up of spherical cells with the HH model on the membrane.

The knowledge of the field inside the tissue at a microscopic level, considering the interaction mechanisms between the exogenous field and cellular membrane, can give answers to microdosimetry problems or permit the dosimetry of new biological experiments to be established where the knowledge of the local field can be taken into account.

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