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ORIGINAL ARTICLE

Numerical model for RF capacitive regional deep hyperthermia in pelvic tumors

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Abstract A numerical model based on Finite Element Method (FEM) for the prediction of power density distribution and temperature during RF-capacitive hyperthermia treatment has been presented in the paper and the results are discussed. In particular the models are related to the treatment of pelvic tumors where it is more difficult to localize and focus heat in deep regions. The geometrical and physical model of the patient is reconstructed with a segmentation procedure by means of dedicated software. The geometrical meshed model has been used as input for the solution of coupled electromagnetic and thermal problems. A deep analysis of different configurations derived from specific scientific literature of the last years has been presented in the paper and discussed. The results obtained by FEM analyses have demonstrated the suitability of this method for the prediction of power and temperature distribution during RF capacitive hyperthermia and that the calculation procedure is an efficient mean to evaluate the efficacy of the heating system.

Keywords RF-capacitive hyperthermia · Coupled electromagnetic field and thermal distribution · Finite element models · Treatment planning

1 Introducoduction

Regional RF capacitive hyperthermia is widely used in different countries as an adjuvant therapy for advanced pelvic tumors [4].

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Heat delivery to deep seated tumors, however, is much challenging: since regional RF capacitive hyperthermia techniques apply energy in an unfocused manner, energy is delivered to both tumor and normal tissues. Under such conditions, selective heating of tumor is only possible when heat dissipation by blood flow in normal tissues is much greater than in tumor tissues. Equipments, which are said to heat to a depth more than 5 cm from the surface, are the low RF (8 and 13.56 MHz) capacitively-coupled heating systems. Those equipments are more simple, and hence much cheaper, than the radiative devices for regional hyperthermia, using frequencies of about 100 MHz [5]. In capacitive-coupled heating devices the patient is placed between two or more electrodes connected to a power generator and the currents flow between the electrodes through living tissues and generate heat through losses in the tissues themselves, mostly due to ionic current. In the literature it is recommended to use electrode diameters of more than 1.5 times the distance between them to achieve a substantial heating at the center of the human body [7]. In order to conform the electrodes to the body surface, the metal plates of the electrodes are covered with flexible water pads, in which temperature-controlled water (5–10°C) flows, so that excessive heating of the skin and subcutaneous fat, which always accompanies this type of technique, can be reduced [5]. Besides, the use of boli larger than electrodes reduce edge effects, i.e. excessive heating near the edge of the electrodes. Nevertheless, the overheating of fat layers remains one of the main disadvantages of regional hyperthermia and clinical experience suggests that this technique is not appropriate for patients with fat layers thicker than approximately 1.5–2 cm [13]. Another considerable limitation of capacitive heating is that the distribution of current and power density in an electrically inhomogeneous body can lead to undesirable hot spots in normal tissues. In particular, the deep SAR distribution is mainly influenced by the shielding effect of the pelvic bone that causes insufficient energy deposition in the central area of the same zone [8].

Today it is clear that the evaluation of a capacitive hyperthermia device with an homogeneous agar phantom leads to overly optimistic results. For this reason, the use of numerical models to simulate the distribution of electromagnetic field and to calculate power density and temperature distribution in an anatomic patient model is fundamental for research purposes and improving of capacitive hyperthermia devices. Moreover, the prediction of temperature inside the patient tissues could bring to a sort of ''treatment planning'' which should be used to optimize the treatment [2].

In previous works some different techniques have been proposed to reduce the overheating of the fat layer, typical of capacitive and radiative RF hyperthermia devices and to reduce hot spots regions.

Proposed techniques consist in the inclusion of a third electrode [10], the use of distributed electrodes [6, 15], precooling of human skin [16], the use of overlay bolus bags filled with a liquid close to the freezing point of water [11].

The aim of this work is to present a general review of these proposed techniques and their performances, based on 3D Finite Element Method (FEM) analysis. In the paper a 3D treatment planning system has been presented with an heterogeneous patient model to evaluate the performances of traditional RF capacitive hyperthermia systems and possible improved devices.

2 Materials and methods

Electromagnetic and thermal field distributions, obtained during a RF hyperthermic treatment, can be calculated with a high resolution ''three-dimensional treatment planning system'' based on an anatomical model obtained from a CT data set. The authors developed a ''treatment planning system'' based on FEM analysis, that allows the calculation of the electric field distribution solving Laplace equation and the temperature distribution solving the bioheat transfer equation (BHTE).

2.1 Electromagnetic and thermal problems formulation

As the dimensions of the domain are small in comparison with the wavelength of the RF electromagnetic field in tissues, the condition of quasi-static electric field approximation can be applied.

At low frequency (8 MHz) the wavelength in human tissues is in the range between 3 and 7 m in relation to the electric permittivity, and this is the reason why it can be assumed the quasi-stationary condition.

With reference to the usual partial differential equations (PDE) which can be derived in this case, the electromagnetic governing equations can be written as follows:

$$\begin{aligned}
&\nabla \cdot \vec{J}_{\text{tot}} = 0,\\
&\vec{J}_{\text{tot}} = \vec{J}_{\text{c}} + \frac{\partial \vec{D}}{\partial t},\\
&\vec{J}_{\text{c}} = \sigma \vec{E},\\
&\vec{D} = \varepsilon \vec{E},\\
&\vec{E} = -\nabla V,
\end{aligned} \tag{1}$$

where $\vec{J}_{\text{tot}}$ is the total current density vector, $\vec{J}_{\text{c}}$ is its conductive component, $\vec{E}$ is the electric field, $\vec{D}$ the electric flux density, σ is the electric conductivity, ε is the electric permittivity and V the scalar electric potential.

Taking into account that the electrodes are supplied by sinusoidal voltage, the time derivative can be written in a phasor way and hence, by simple mathematical derivations, we obtain:

$$\nabla \cdot [\sigma(-\nabla V) + j\omega\varepsilon(-\nabla V)] = 0. \tag{2}$$

This formulation of the electromagnetic problem is based on a classical approach, and the material properties are also defined in this way. On the other hand, in literature the experimental values of conductivity are referred to a ''non-classical approach'' [9] and the equivalent electrical conductivity σ_{e} is defined as:

$$\sigma_{\text{e}} = \sigma + \omega\varepsilon'', \tag{3}$$

where the first term is related to the conductive phenomena (ionic current); it consists in the ''real'' conductivity of the tissue and it can be considered independent from temperature and frequency. The second term, depending on the frequency, corresponds to the dielectric contribution and it is negligible, in human tissues, for frequencies lower than some hundred MHz.

Moreover the physical model of real dielectrics describe the electric permittivity by means of a complex number:

$$\varepsilon = \varepsilon' - j\varepsilon'',$$

where ε'' is responsible of losses in the material.

Taking into account these considerations, the formulation of the electromagnetic problem can be described by the following equation:

$$\nabla \cdot [-(\sigma + \omega\varepsilon'' + j\omega\varepsilon')\nabla V] = 0 \tag{4}$$

which is the Laplace equation to be solved in a domain characterized by properties described by complex numbers, it means that the electric permittivity is expressed by means of a complex number.

The BHTE formulated by Pennes [12] describes thermal processes in the human body, taking into account the cooling effect of the blood flowing into capillaries:

$$\rho c \frac{\partial T}{\partial t} = \nabla \cdot (k \nabla T) + \sigma E^2 - c_b W_b (T - T_b) + Q_m, \qquad (5)$$

where E is the root mean square (''rms'') value of the electric field which is determined by the solution of the electromagnetic problem; ρ is density (kg/m^3); c is specific heat capacity (J/kg K); k is thermal conductivity (W/m K); T is temperature (K); c_b is specific heat capacity of blood (J/kg K); and W_b is blood mass flow rate (kg/s m^3); T_b temperature of blood (K) and Q_m is the specific power due to metabolic process (W/m^3). A constant value, for power density related to this last term, has been assumed, equal to 1 W/kg.

In the transient thermal conduction equation (5), the term $c_b W_b (T - T_b)$ represents the ''heat sink'' term that models the heat rate exchange between blood in capillaries and tissues as a heat source. The blood mass flow rate has been taken into account as function of the actual temperature in every tissue involved in the model to correctly evaluate the temperature-dependent response of vasculature in tissues to heat stress. In our work, we employ the following blood perfusion rate values [14]:

$$W_{\text{muscle}} = 0.5 + 3.5 \exp\left[\frac{-(T-45)^2}{12}\right],$$

$$W_{\text{fat}} = 0.36 + 0.36 \exp\left[\frac{-(T-45)^2}{12}\right],$$

$$W_{\text{tumor}} = \begin{cases} 0.8 & T < 37 \\ 0.8 - (T-37)^{4.8}/5400 & 37 \le T \le 42\,. \\ 0.38 & T > 42 \end{cases}$$

In Fig. 1 perfusion rate values versus temperature in different tissues are reported. Systemic heating and discrete vasculature are not included in the model, in fact the Pennes model describes blood perfusion with acceptable accuracy, if large vessels are not nearby. For localized vessels a different calculation is needed, which takes into account the hydrodynamic effects [1].

Moreover, we would underline the limits of the blood perfusion formulation, because in a more realistic model the blood mass flow rate should be variable also with heating time.

The values of electric and thermal properties for the human tissues used in the electric and thermal computation have been derived from literature (http://www.niremf.ifac.cnr.it/cgi-bin/tissprop/htmlclie/uniquery) [3, 8].

Electrical and thermal properties of tissues involved in the treatment are listed in Table 1.

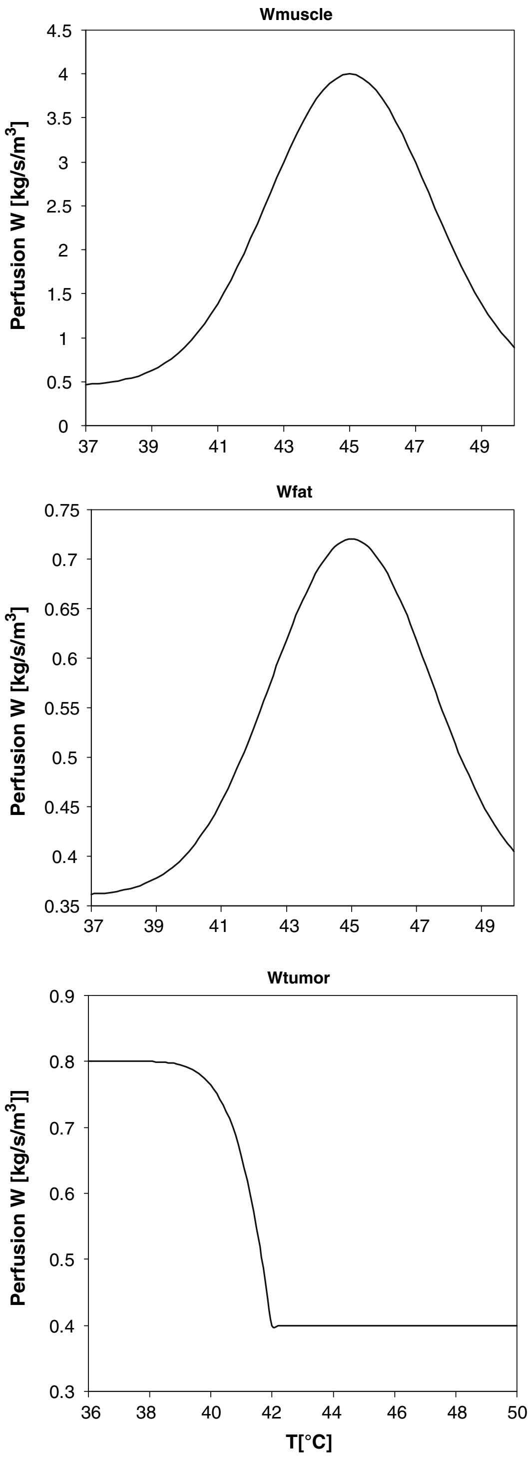


Fig. 1 Perfusion rate values versus temperature in different tissues

Table 1 Electric and thermal properties of the materials in the FEM models

	Electric conductivity σ_e (S/m)	Relative permittivity ε'	Thermal conductivity k (W/m °C)	Density ρ (kg/m^3)	Specific heat c (J/kg °C)
Fat	0.053	29.6	0.22	888	2,387
Muscle	0.64	160	0.56	1,050	3,639
Bone	0.043	36.8	0.65	1,595	1,420
Tumor	0.64	160	0.56	1,050	3,639
Bolus water	2	76.5	6.0	1,000	4,180

2.2 Model construction

The patient model has been derived from a computed tomography data set composed by 40 slices, with slice thickness of 5 mm of an Italian female patient with a rectal carcinoma.

Model reconstruction has been carried out with a semi-automatic procedure: the CT slices were segmented by Hounsfield Unit Thresholding and manually using the Amira package (Amira-TGS). The boli geometry is prepared directly inside Amira ambient too. Also the construction of the tetrahedral mesh is performed automatically by using Amira tools (Fig. 2). The geometry and mesh information are stored in a file which is used as input for the FEM solver.

In the simulation of thermal treatments with different techniques, the following parameters have been used: the electrodes dimensions are variable in the range of 20–30 cm in diameter, the electrodes voltages are chosen to obtain a total absorbed power in human body in the range between 400 and 1,000 W, according to clinical practice [11, 16]; the temperature of bolus bags at the constant value of 10°C and the ambient air at 25°C. A typical time duration for such a kind of treatment is in the range of 20–45 min: in order to compare the results of different configurations, the simulation time has been set at 35 min. The tumor, which can be considered as target zone in the patient, has a volume of 183 cm^3. The mean fat thickness of the patient is about 2.5 cm.

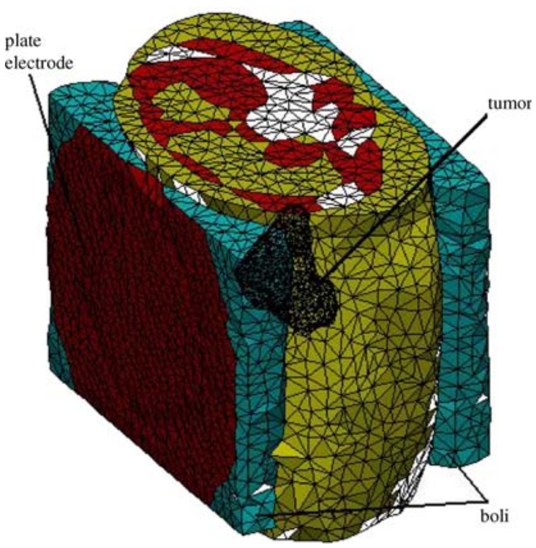


Fig. 2 3D geometric model and mesh for FEM calculations

The electric field and temperature distributions are computed using a FEM code, Flux 3D Cedrat, able to solve coupled time harmonic electromagnetic problems and thermal transient problems.

It should be pointed out that the heat sink term gives rise to a nonlinearity in the thermal transient problem which must be solved by Newton–Raphson method. The time step, in the solution of thermal problem has been set constant and equal to 10 s

3 Results and discussion

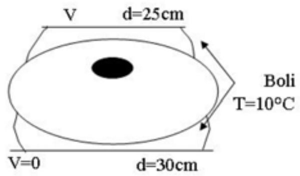
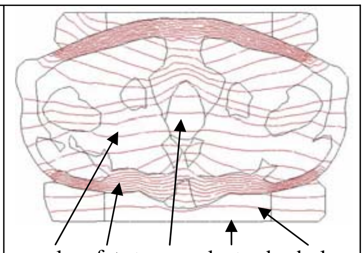
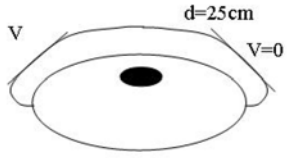
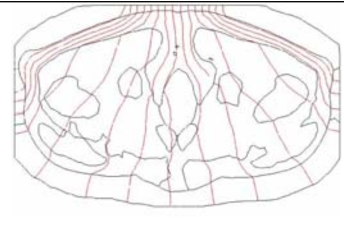
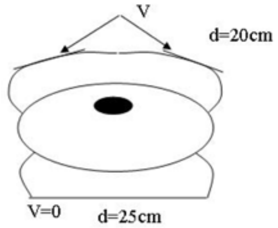
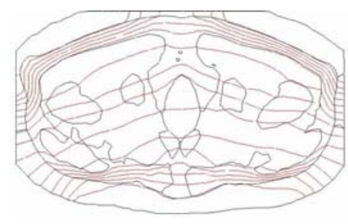
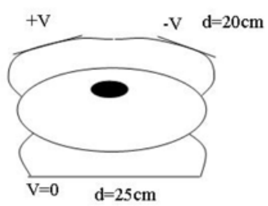
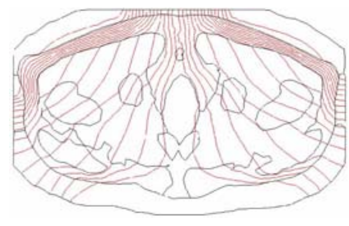
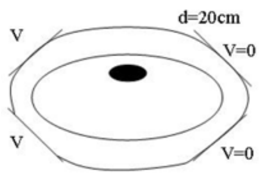
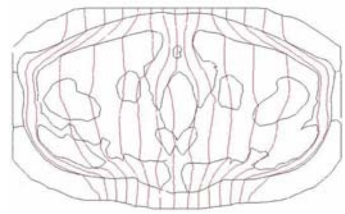
The results for different devices and strategies of intervention are presented and summarized in Tables 2 and 3.

For each simulated treatment, the total power dissipated in the human body, the power dissipated in the tumoral region, equipotential lines of the electric voltage on the a transversal section (Table 2) are shown; moreover the maximum temperature and the thermal distribution on the sagittal and transversal sections (Table 3) after 35 min of treatment are reported.

The CT slices used are the same for each simulation, but some geometric differences are due to different segmentations in the model construction.

The first treatment we simulated is constituted by two traditional plate electrodes attached to the skin of the patient by means of boli with circulated controlled temperature water at 10°C. Their dimensions are 25 and 30 cm diameter. The electric potential distribution shows that the electric field magnitude is higher in the region with a low permittivity and conductivity, mainly in the fat. From the thermal distribution map, representing temperature at the end of the treatment, excessive overheating of fat layers can be observed: bolus cooling effect is insufficient when fat layers exceeds 1.5 cm.

Table 2 Power dissipated in the all treated region (P_{tot}) and in the tumoral region (P_{tum}) and electric potential distribution, plotted as contours, on a transversal section

2 electrodes 	P_{tot} =509.3 W P_{tum} =3.56 W	
2 electrodes at 45° 	P_{tot}=386.9 W P_{tum}=2.78 W	
3 electrodes (common mode) 	P_{tot}=691.9 W P_{tum}=3.59 W	
3 electrodes (differential mode) 	P_{tot} =512.6 W P_{tum} =2.56 W	
4 electrodes 	P_{tot} =500.3 W P_{tum} =4.76 W	

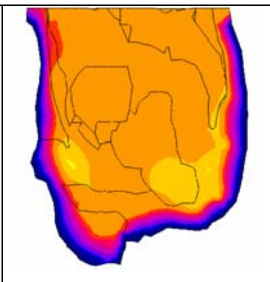
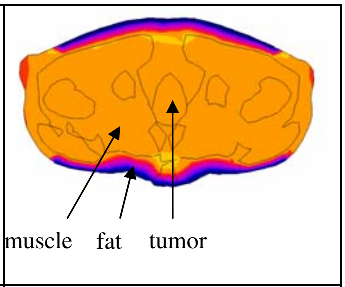
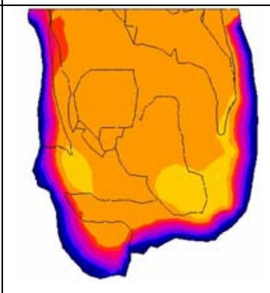
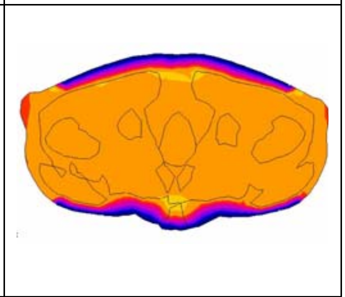
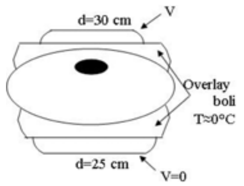
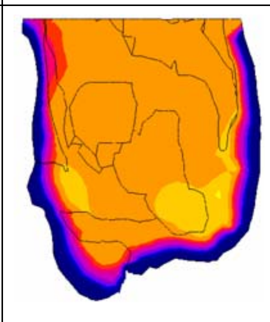
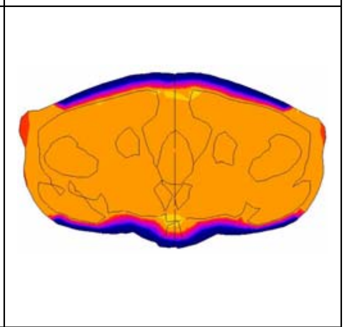
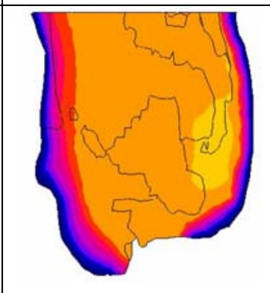
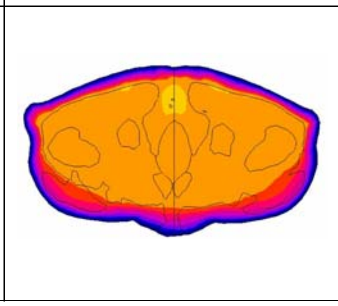
Considerable shielding effect of pelvic bone, besides, causes a SAR maximum in muscle layer between and below the bone of pubis. Tumor heating is inhomogeneous.

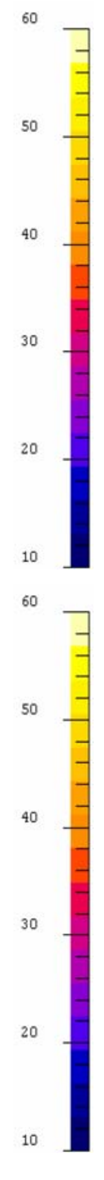
Boli pre-cooling, for 20 min at 10°C, effectively reduces the temperature of subcutaneous fat layers, but it cannot avoid preferential heating at the interface from fat and muscle tissue.

Overlay bolus bags can be used to reduce the excessive superficial heating. These are circulated with a liquid close to the freezing point of water. In order to simulate the effect of an overlay bolus, the temperature of the boli is set to 0°C and the ambient air is replaced with water at a fixed temperature of 0°C. The insertion of overlay boli causes a strong superficial cooling partially that reduces the temperature in the subcutaneous fat layers. Unfortunately this effect is only very superficial.

With *two electrodes placed at 45°* on the back of the patient, a more uniform heating of the tumor and a reduction of the overheating of fat and muscle in the

Table 3 Temperature values (T_{maxtot} = maximum temperature in the whole region, T_{maxtum} = maximum temperature in the tumoral region and T_{mintum} = minimum temperature in the tumoral region) and thermal distribution on sagittal and transversal sections at 35 min

2 electrodes T_{maxtot}= 53.6 °C T_{maxtum}= 49.9°C T_{mintum}=40.3 °C		
Pre-cooling for 20 minuts at 10°C T_{maxtot}= 52.4°C T_{maxtum}= 49.7 °C T_{mintum}= 40.4°C		
Overlay boli  T_{maxtot} = 52.4°C T_{maxtum}= 49.6°C T_{mintum}= 40.3°C		
2 electrodes at 45° T_{maxtot}= 51.4°C T_{maxtum}= 46.1°C T_{mintum}= 38.3°C		



anterior part of patient are observed. However, SAR and temperature maxima are in normal tissues around the sacrum.

The inclusion of *a third electrode* provides a power redistribution. With the posterior electrodes supplied in common mode the heating is more homogeneous and a reduction of ''hot spot'' is observed. With the electrode supplied in differential mode, the overheating of the anterior fat layer is cancelled, but the problem remains on the opposite layer.

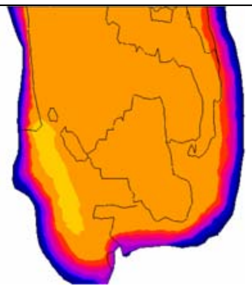
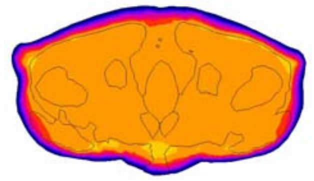
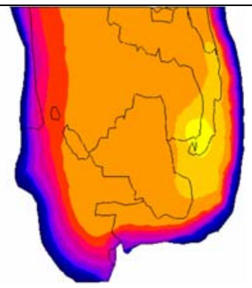
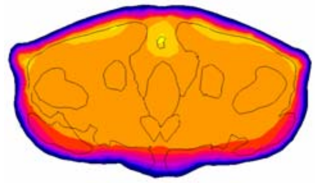
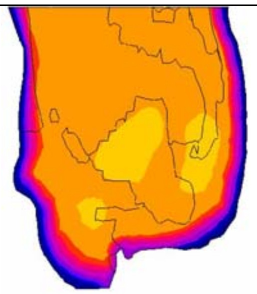
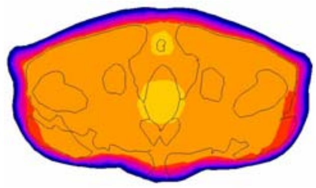
Finally, for *four electrodes configuration*, electric field distribution is totally changed: its direction is mainly perpendicular to the fat-muscle interface, reducing the overheating of fat. In this way the effect of boli is sufficient to prevent hot spots. Quite 1% of the total power is placed inside the tumor.

The shielding effect of the bone still remains as the main limitation to a successful application of this technique.

The maximum temperature reached in the calculations are in the range between 51 and 58°C, but the applied voltage has been chosen in order to obtain temperature in the range between 41 and 50°C in the tumor region.

On the basis of FEM simulations performed we can say that RF capacitive treatments for pelvic tumors have low efficiency. In fact, if a therapeutic temperature is reached in the target zone, some hot spots are present in other regions. Moreover the pelvic bone behaves like an electromagnetic shield.

Table 3 continued

3 electrodes *(common mode)* $T_{maxtot} = 57.8°C$ $T_{maxtum} = 44.7°C$ $T_{mintum} = 40.8°C$			
3 electrodes (differential mode) $T_{maxtot} = 65.2°C$ $T_{maxtum} = 47.1°C$ $T_{mintum} = 38.8°C$			
4 electrodes $T_{maxtot} = 57.7°C$ $T_{maxtum} = 48.6°C$ $T_{mintum} = 38.6°C$			

The aim of the modeling is to obtain temperature close to the therapeutic ones, as much as possible, in the target zone.

This technique seems to be more suitable for the part body treatment where an undifferentiated heating must be applied.

The overheating of fat layers, due to the perpendicularity of the electric field to the fat–muscle transition, is the main drawback of the RF capacitive heating. It should be pointed out that also the radiative heating systems present this problem, but the phase and amplitude steering of the antenna array can reduce the effect of hot spots.

4 Conclusion

The authors have presented a general review of the proposed techniques for RF capacitive regional hyperthermia based on the comparison of different types of applicator positions and configurations. A coupled FEM electric and thermal analysis has been performed on one patient anatomy. Results show the feasibility of FEM modeling of RF capacitive heating and give quantitative information on the effectiveness of simulated treatments.

Pelvic tumors are not very suitable for capacitive heating treatments, in fact, the simulation results put in evidence that in a patient model with pelvic tumor, the power deposition is mainly influenced in deep by pelvic bone structure, and in the superficial layers by the thickness of fat tissue. The main limitations of capacitive heating technique are the perpendicularity of the electric field to the longitudinal axis of the human body that causes subcutaneous fat overheating and the shielding effect of bone. Some authors suggest the possibility to apply an electric field parallel to the human body axis in order to prevent this effect [6]. The modification of size, number and position of electrodes gives a SAR redistribution in the body and the insertion of overlay boli, for strong superficial cooling, can reduce the edge effects of electrodes and the subcutaneous overheating. However the structure of pelvic bone causes an inhomogeneous distribution of current and power density and consequent SAR maxima in normal tissues.

The effective temperature can be obtained only causing overheating of fat and muscle tissues, in spite of the cooling effect of bolus bags and blood perfusion. In fact, calculations show that extremely high temperature, up to 60°C, are reached in normal tissues regions, if therapeutic temperature are obtain in deep-seated tumors. Such high temperature are not tolerable by patients during clinical practice.

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