Modeling of Planar Applicators for Microwave Thermotherapy

Julien Carlier, Vincent Thomy, Jean-Christophe Camart, Luc Dubois, and Joseph Pribetich, *Member, IEEE*

***Abstract*—In order to improve the external applicators used for microwave thermotherapy controlled by microwave radiometry in medical applications, we propose specific planar applicators developed for heating: either annular ones to be used at the frequency equal to 915 MHz or in the shape of a horseshoe (short-circuited ring) for 434 MHz. The final goal of this paper is the realization of a honeycomb network for the treatment of larger areas and greater volumes.**

***Index Terms*—Finite-difference time-domain (FDTD) method, medical applications, microwave heating, microwave planar applicators.**

I. Introduction

AN INCREASED interest in the applications of microwave energy for medical diagnostics and therapy has been observed over the past decade [1]–[4]. The objective of microwave thermotherapy is to obtain a well-confined raising of temperature within tissues. The research works undertaken in this domain aim at increasing the efficiency of the heating in volume and depth in order to treat larger areas located in various places of the human body. A large number of devices [5]–[7] have been designed and tested for these medical applications. Among these devices, we have been interested for over a decade in the study of external planar applicators [8], [9]. We present in this paper the latest results (theoretical study and experimental verifications) concerning the specific planar applicators we have developed for heating in medical applications. We present first in Section II the different applicators that have been studied. The theoretical determination of the thermal patterns is then briefly recalled in Section III as the experimental measurements. Section IV is concerned with results and discussion. Concluding remarks and perspectives are then presented in Section V.

II. Material

The applicator used in thermotherapy plays a fundamental role because the efficiency of the heating depends on its shape. In many cases, the applicators are external planar ones and present many advantages, i.e., small size, light weight, capable of conforming to the shape of the body, and low cost. Further, they are effective in coupling energy directly into tissues with minimal stray fields. The microstrip–microslot applicators are realized on a dielectric substrate on which is etched a feeding line ended by a subminiature version A (SMA) connector and connected to a microwave generator. On the other face, an aperture having different shapes (square, rectangular, circular, etc.) is opened in the metallic ground plane. This aperture is the radiating element, which allows the propagation of the electromagnetic waves in the dissipative medium on which the applicator is laid. Generally, a thermostated water bolus is placed between the applicator and the skin in order to avoid possible cutaneous burns. This bolus is realized from a very malleable plastic pocket, which allows the exact shape of the medium to be taken and offers good contact between the applicator and this medium while giving a reasonable superficial temperature.

For this study, several kinds of applicators have been realized, the shapes of which depend upon the frequency to be used. For the frequency equal to 915 MHz, we have studied the following planar annular applicators.

- A "single" applicator [see Fig. 1(a)] realized on a dielectric substrate of relative permittivity ε_r equal to 4.9 and a thickness equal to 1.58 mm. The aperture is of annular shape and its dimensions are the following: an internal diameter of 28 mm and an external diameter of 38 mm.
- A "twin" applicator [see Fig. 1(b)] realized from two "single" applicators set side by side. This configuration allows us to test the efficiency of the association of several applicators: the final goal is the realization of a honeycomb network for the treatment of large areas. The two rings have the same dimensions: an internal diameter of 25 mm and an external diameter of 45 mm. They are separated from a distance H equal to 4 mm.

For both applicators, the feeding line has a length equal to 82 mm and a constant width of 2.5 mm.

As to keep reasonable dimensions for the applicators at the frequency equal to 434 MHz, we have designed a new kind of applicators: they have the shape of a horseshoe, that is to say, a ring ended with a short circuit. We have also studied several planar applicators.

- A "single" applicator [see Fig. 2(a)] realized on the same dielectric substrate as that of the previous ones (relative permittivity ε_r equal to 4.9 and of thickness equal to 1.58 mm). The dimensions of the short-circuited ring are the following: an internal diameter of 20 mm, an external diameter of 40 mm, and width of the short circuit of 16 mm.
- A "twin" applicator [see Fig. 2(b)] realized from two "single" applicators set side by side. They have the same dimensions as that of the "single" one. This configuration

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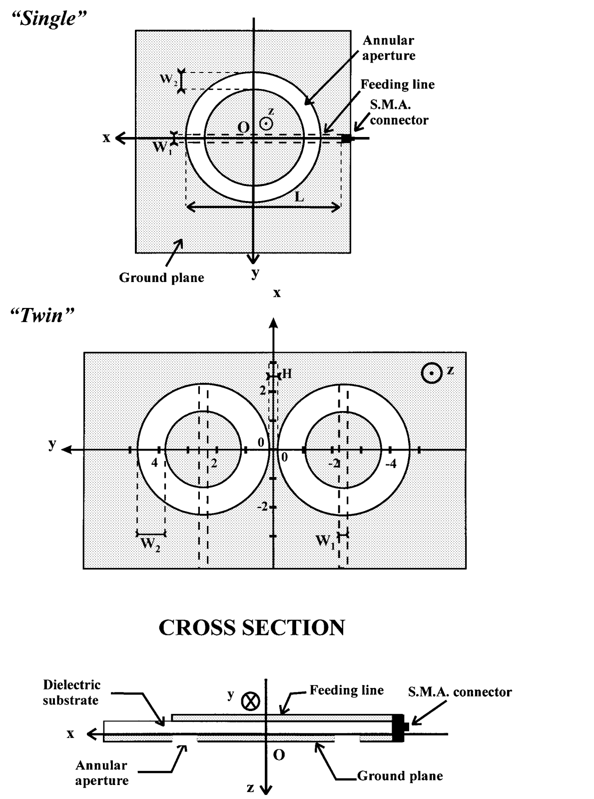

Fig. 1. Top view and cross section of the "single" and "twin" annular applicator. $W_1 = 2.5$ mm, $W_2 = 10$ mm, $L = 82$ mm, and $H = 4$ mm.

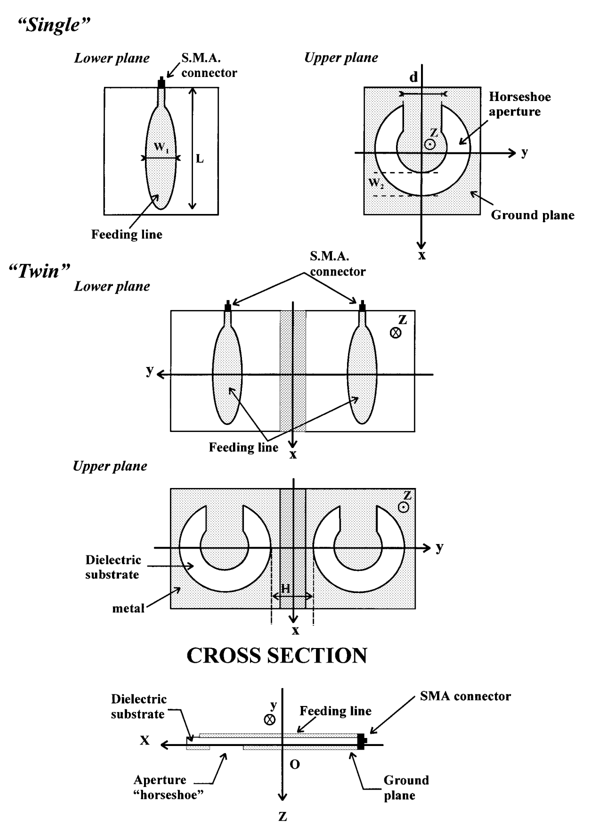

Fig. 2. Top view (lower and upper planes) and cross section of the "single" and "twin" horseshoe applicator. W_1 varying from 2.5 to 13 mm. $W_2 = 10$ mm, $d = 16$ mm, $L = 48$ mm, and $H = 14$ mm.

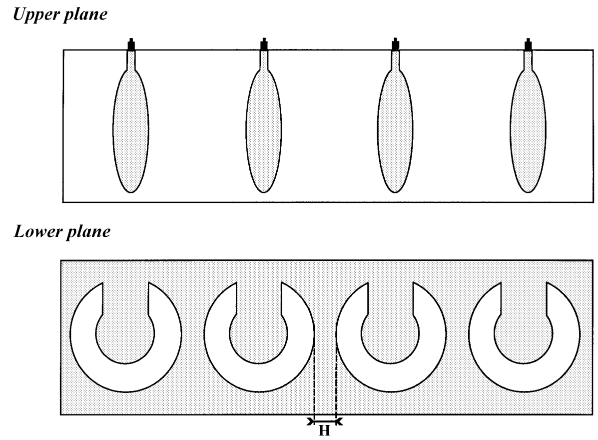

Fig. 3. Top view (lower and upper planes) of the "quadruple" horseshoe applicator. W_1 varying from 2.5 to 13 mm. $W_2 = 10$ mm, $d = 16$ mm, $L = 48$ mm, and $H = 8$ mm.

allows us to test the efficiency of the association of two applicators to be inserted in a box. In this case, the two rings are separated from a distance H equal to 14 mm.

- A "quadruple" applicator (see Fig. 3) realized from four "single" applicators set side by side. This configuration allows us to test the efficiency of the association of several applicators: the final goal is the realization of a honeycomb network for the treatment of large areas. Each ring is separated from its neighbor by a distance H equal to 8 mm.

In order to obtain a good matching at the chosen heating frequency equal to 434 MHz for all the applicators of this new kind, we have to change the shape of the feeding line. Indeed, if we keep its width constant, the resonant frequency is then equal to 1.5 GHz. Thus, the feeding line has a shape of a petal with a length equal to 48 mm and a width varying between 2.5 mm near the SMA connector and the maximum value equal to 13 mm.

III. Modeling and Measurements

The development of microwave applicators for medical applications requires the knowledge of interactions between electromagnetic waves and biological tissues. The studies at once theoretical and experimental allow us to know the electromagnetic behavior of the applicator coupled to the dissipative media and to obtain the thermal mapping directly coupled with the concept of therapeutic efficiency.

The main theoretical problem to characterize these applicators is the determination of the radiating pattern. Thus, in order to take into account the exact shape of tissues and also of the applicators, a three-dimensional (3-D) model based on the well-known finite-difference time-domain (FDTD) method [10]–[12] (which was first proposed by Yee [13]) has been developed.

The FDTD method technique proceeds by segmenting the studied volume, i.e., the structure and surrounding space, into a 3-D mesh composed of a number of rectangular unit cells. In the finite-difference procedure proposed by Yee, the E- and H-fields are evaluated at half-step intervals around the unit cell and at alternative half time steps, effectively giving centered difference expressions.

The temporal step is linked to the spatial one by a stability criterion [14]. Due to the dissipative media, the treatment of the field components at the lattice truncation is obtained using Mur's first-order absorption boundary condition [15] for each field component.

Once the electric fields are found, we obtain the absorbed microwave power density [specific absorption rate (SAR)] in the dissipative media at each space point $i,\ j,\ k$ from the following formula:

$$\mathrm{SAR}\,(i,\,j,\,k) = \frac{\sigma}{2}\left(|E_x(i,\,j,\,k)|^2_{\max} + |E_y(i,\,j,\,k)|^2_{\max} + |E_z(i,\,j,\,k)|^2_{\max} \right)$$

where $E_{x_{\max}}$, $E_{y_{\max}}$, and $E_{z_{\max}}$ are the maximum steady-state electric field components at cell $(i,\,j,\,k)$ and σ is the conductivity of the medium of this cell ($\sigma = \varepsilon_0 \cdot \varepsilon'' \cdot 2\pi f$; $\varepsilon_0 = 8.85.10^{-12}$ $\mathrm{F \cdot m^{-1}}$ and ε'' is the imaginary part of the permittivity).

The determination of the thermal mapping in the heated media is then deduced from the resolution of the heat-transfer equation in the steady state [16] using a model based on a finite-difference method

$$\rho c \frac{dT}{dt} = k_t \nabla^2 T + V_s (T_a - T) + Q_m + Q$$

where

- ρ density of the biological medium ($\mathrm{kg \cdot m^{-3}}$);
- c specific heat ($\mathrm{J \cdot kg \cdot {}^\circ C^{-1}}$);
- T temperature at the considered point (°C);
- t time (s);
- k_t thermal conductivity of the biological medium ($\mathrm{W \cdot m^{-1} \cdot {}^\circ C^{-1}}$);
- V_s blood–heat-exchange coefficient ($\mathrm{W \cdot m^{-3} \cdot {}^\circ C^{-1}}$);
- T_a internal temperature of the human body (°C);
- Q_m heat generated by the metabolism ($\mathrm{W \cdot m^{-3}}$);
- Q deposited power density ($\mathrm{W \cdot m^{-3}}$).

The heat generated by the metabolism Q_m has been neglected in comparison with the deposited power density Q, which is greater. In addition, the results are obtained at the thermal equilibrium. At the interface between the applicator and medium, the heat losses are expressed by the coefficient H ($\mathrm{W \cdot m^{-2} \cdot {}^\circ C^{-1}}$), of which its value depends upon the temperature at this interface. The polyacrylamide gel used as a phantom medium is characterized by a thermal conductivity k_t equal to 0.38 $\mathrm{W \cdot m^{-1} \cdot {}^\circ C^{-1}}$. The term V_s is set at 1500 $\mathrm{W \cdot m^{-3} \cdot {}^\circ C^{-1}}$ and the term H is constant and equal to 100 $\mathrm{W \cdot {}^{-2} \cdot {}^\circ C^{-1}}$. For the results which will be presented, we have used a water bolus placed between the polyacrylamide gel and the applicator with a thickness of 12 mm, and the temperature of the water have been fixed at 20 °C.

Confirmation of the theoretical approach is given by experimental measurements, which have been carried out on phantom models of human tissues (saline solution at 6 g/l or polyacrylamide gel). We have first measured the $|S_{11}|$-parameter as a function of frequency in order to control the level of impedance matching of the applicator by means of a network analyzer (HP 8510). We have then determined the power deposition in saline solution with a simple system for mapping the electric-field pattern created by the microwave applicator under test. Finally, the thermal performances have been characterized by using an automatic experimental system [9].

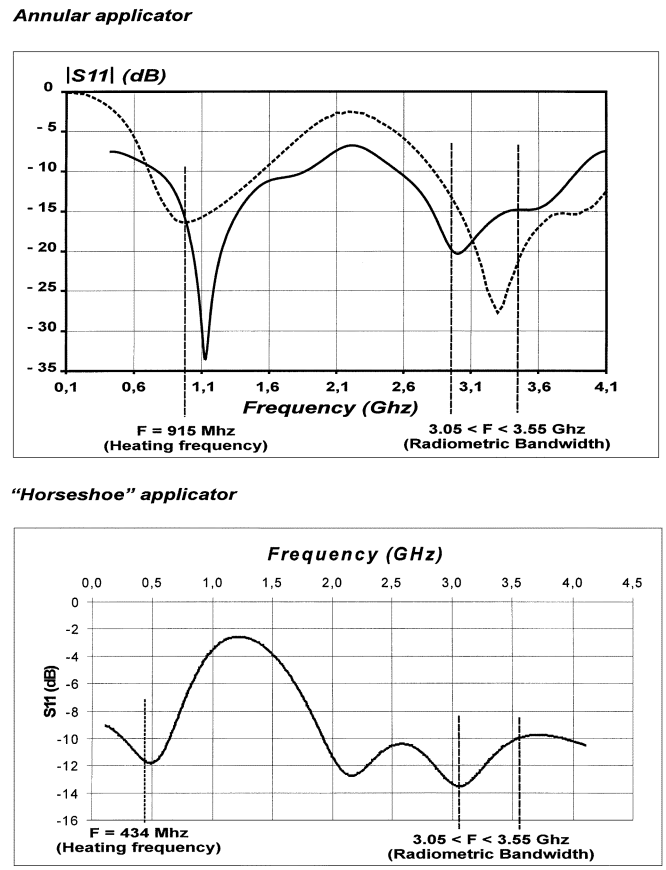

Fig. 4. Theoretical (—) and experimental (- - -) variations of the reflection coefficient $|S_{11}|$ as a function of frequency for the annular and horseshoe applicators.

IV. Results and Discussion

The theoretical determination of the evolution of the $|S_{11}|$-parameter as a function of frequency allows us to verify the good matching of the applicator at the heating frequency, but also in the radiometric bandwidth (in order to achieve noninvasive measurements of the temperature by radiometry [17]).

We consider that the matching is good when the reflection coefficient $|S_{11}|$ is less than −10 dB, i.e., that at least 90% of the incident power is delivered to the medium to be heated. An example of a comparison between theory and experiment is given in Fig. 4 for the "single" annular applicator. We can note that the values are less than −15 dB as well at the heating frequency (915 MHz) as in the radiometric bandwidth (3.05 GHz $< f <$ 3.55 GHz).

Another example of the evolution of the $|S_{11}|$-parameter versus frequency is given in Fig. 4 for the "single" horseshoe applicator. We can note that the values are near −12 dB at the

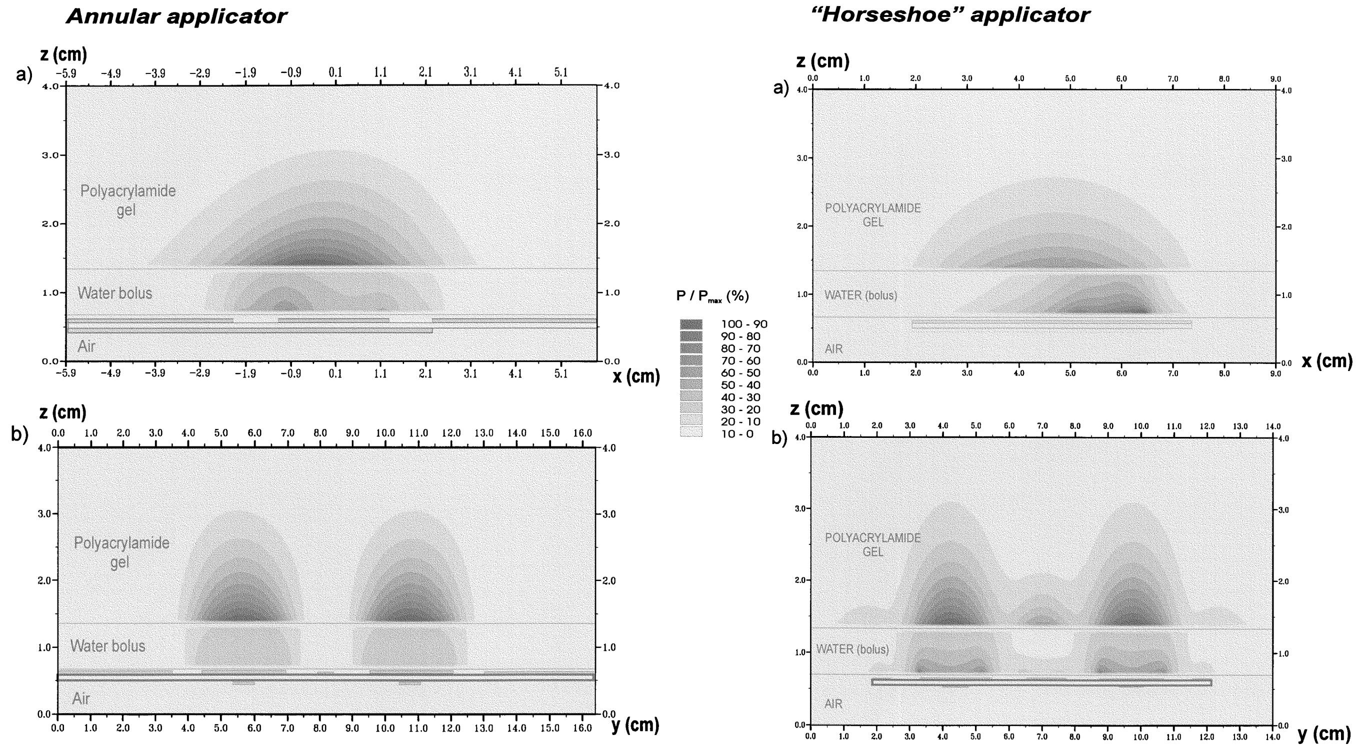

Fig. 5. Power deposition calculated from the FDTD for the "twin" annular and "twin" horseshoe applicators. (a) In the plane xOz. (b) In the plane yOz.

heating frequency (434 MHz) and from -10 to -14 dB in the radiometric bandwidth (3.05 GHz $< f <$ 3.55 GHz).

The second step of the study concerns the power deposition. We present in Fig. 5 the theoretical results concerning the power deposition for the "twin" applicators (for both annular and horseshoe) obtained in two planes: xOz along the feeding line and yOz perpendicular to this direction and situated at the level of the rings in the middle. We can observe a dissymmetry along the first plane. In the other one, the curves are symmetrical with respect to the feeding line and we can observe two lobes centered on each of the rings with a composition of the electromagnetic fields between the rings for the horseshoe applicator.

The last step of the study concerns the thermal mapping. We will first specify the notion of the 37% isothermal line. We consider that the therapeutic effect will be efficient if the temperature in tissues is situated between 40 °C–45 °C, the corporeal one being equal to 37 °C. Thus, an increase of at least 3 °C is needed: this one corresponds to 37% of the maximum increase of the temperature (equal to 8 °C).

As an illustration, we present the normalized thermal mappings (with respect to the maximum temperature) obtained for the "single" applicators (Fig. 6) and the "twin" applicators (Fig. 7) in the same planes defined previously.

For the "single" annular applicator, we can note that the zone delimited by the 37% isothermal line spreads until 26 mm in depth and has a surface of approximately 23 cm^2 [see Fig. 6(a)]. For the horseshoe applicator, it gets approximately the following results: the zone delimited by the 37% isothermal line spreads until 35 mm in depth (consistent with the decrease of the frequency, which involves an increase of the depth penetration) and has a smaller surface of approximately 19 cm^2. Thus, the therapeutic volume is roughly the same for the two "single" applicators (60 cm^3 for the annular applicator against 64 cm^3 for the horseshoe applicator).

For the "twin" annular applicator [see Fig. 7(a)], this zone spreads on a depth of 28 mm and covers an area of approximately 62 cm^2. Thus, the therapeutic volume is about three times greater for the association of two applicators (175 cm^3 against 60 cm^3 for the "single" one).

For the "twin" horseshoe applicators, the zone delimited by the 37% isothermal line spreads until 36 mm in depth and has a surface of approximately 54 cm^2 [see Fig. 7(b)]. Thus, we find the same results: the therapeutic volume is also three times greater for the association of two applicators (195 cm^3 against 64 cm^3 for the "single" one).

Thus, for both kinds of applicators (either annular or horseshoe), the association of two applicators involves an increase of the heating volume, which is about three times greater. We can note that, as expected, the depth is greater for the lower frequency. However, it is possible to modify this depth in varying the thickness of the water bolus, which is the same in this study for both frequencies.

In order to see what we can obtain with the quadruple horseshoe applicator, we present in Fig. 8 the power deposition calculation obtained from the FDTD in the two planes: yOz perpendicular to the feeding lines and xOy parallel to the applicator. In the first plane, the aspect of the curve looks like the ones given in the same plane for the "twin" horseshoe applicator: the lobes are centered on each ring and are symmetrical with respect to the feeding line. Thus, the heating zone will approximately have the shape of a rectangular given by power deposition given in the plane xOy. If we compare this surface to the one given by the "twin" horseshoe applicator in the same plane xOy, we obtain an increase of the area greater than twice. This study is still ongoing.

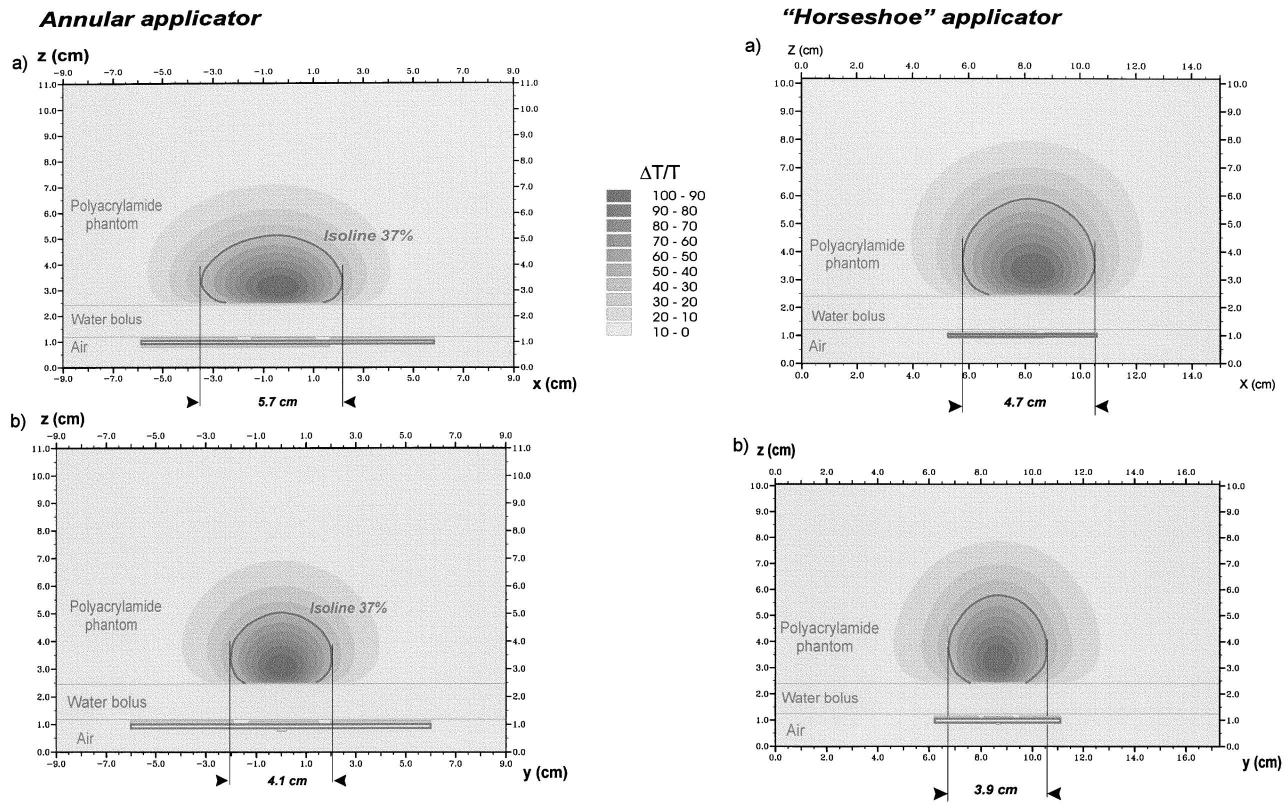

Fig. 6. Thermal mapping for the "single" annular and "single" horseshoe applicators with a water bolus of thickness 12 mm (at $T = 20$ °C). (a) In the plane xOz. (b) In the plane yOz.

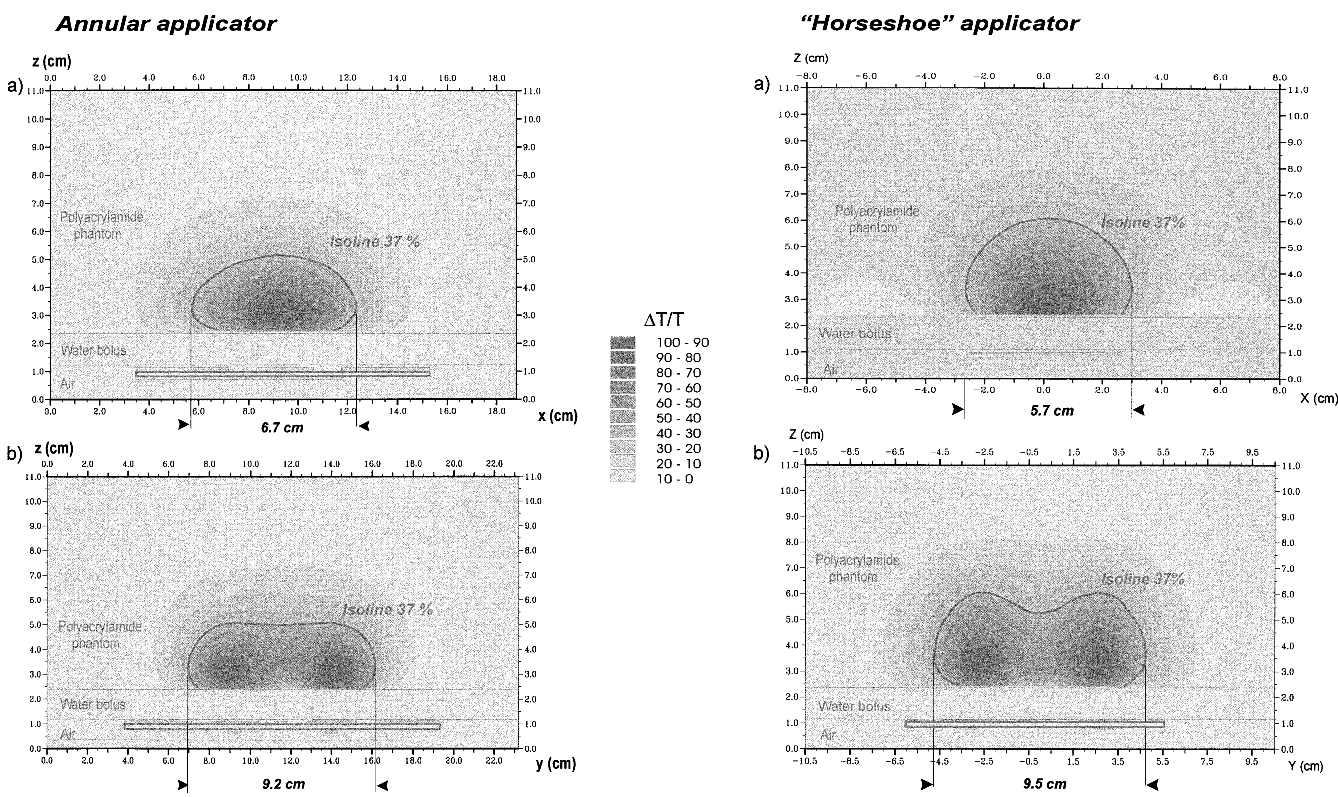

Fig. 7. Thermal mapping for the "twin" annular and "twin" horseshoe applicators with a water bolus of thickness 12 mm (at $T = 20$ °C). (a) In the plane xOz. (b) In the plane yOz.

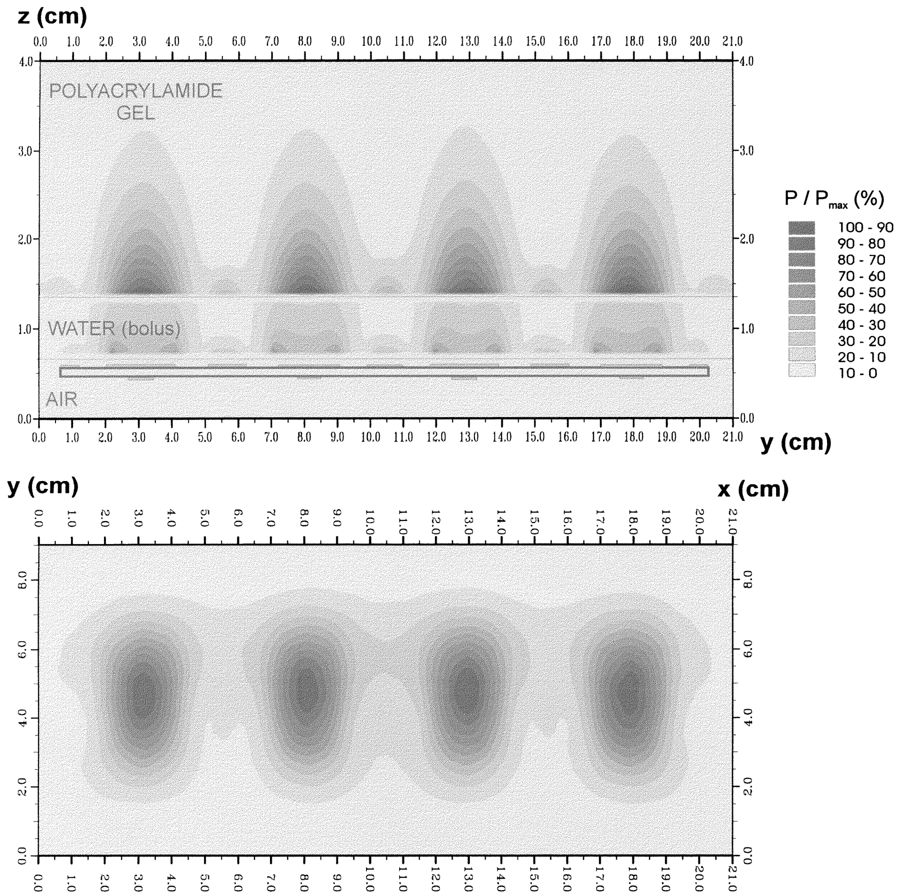

Fig. 8. Power deposition calculated from the FDTD for the "quadruple" horseshoe applicator with a water bolus of thickness 12 mm (at $T = 20$ °C). (a) In the plane yOz. (b) In the plane xOy.

V. Conclusion

We have presented the study of planar applicators with an annular or horseshoe shape to be used for microwave heating in medical applications. The comparison between theoretical results and experimental measurements obtained at the heating frequency either at 915 or at 434 MHz shows the efficiency of the 3-D FDTD as a simulation method. The thermal results point out the interest of the "twin" applicator for which the therapeutic zone is about three times greater than the one of a "single" applicator, whatever the type of applicator (annular or horseshoe).

The next step of this study concerns the development of arrays of applicators (four or more) in order to heat larger areas and greater volumes. We are also envisaging the possibility to introduce a phase difference for the current in the different feeding lines of the array of applicators in order to modify the shape of the heated zone. The study of the influence of the thickness of the water bolus on the depth of the heating pattern will also be studied.

References

[1] K. L. Carr, "Microwave heating in medicine," *Microwave J.*, pp. 26–36, Oct. 1991.
[2] A. Rosen, D. Rosen, G. A. Tuma, and L. P. Bucky, "RF/Microwave-aided tumescent liposuction," *IEEE Trans. Microwave Theory Tech.*, vol. 48, pp. 1879–1884, Nov. 2000.
[3] S. C. Hagness, A. Taflove, and J. E. Bridges, "Three-dimensional FDTD analysis of a pulsed microwave confocal system for breast cancer detection: Design of an antenna-array element," *IEEE Trans. Antennas Propagat.*, vol. 47, pp. 783–791, May 1999.
[4] E. C. Fear and M. A. Stuchly, "Microwave breast cancer detection," in *IEEE MTT-S Int. Microwave Symp. Dig.*, Boston, MA, June 11–16, 2000, pp. 1037–1040.
[5] Y. Nikawa, S. Mori, and F. Okada, "Flexible microstrip applicator for biomedical applications," in *Proc. ISAP*, Tokyo, Japan, Aug. 22–25, 1989, pp. 265–268.
[6] M. F. Iskander and T. S. Lind, "Electromagnetic coupling of microstrip lines and coplanar waveguides flexible to multilayer lossy media," in *IEEE MTT-S Int. Microwave Symp. Dig.*, Long Beach, CA, June 13–16, 1989, pp. 175–178.
[7] J. C. Kumaradas and M. D. Sherar, "Finite element method aided design of a beam shaping hyperthermia microwave applicator," presented at the XXVIth URSI General Assembly, Toronto, ON, Canada, Aug. 13–21, 1999.
[8] L. Dubois, P.-Y. Cresson, J. Pribetich, and M. Chive, "Analysis of microstrip–microslot applicator for biomedical applications," *Microwave Opt. Technol. Lett.*, vol. 5, no. 12, pp. 672–682, Dec. 1992.
[9] P.-Y. Cresson, C. Michel, L. Dubois, M. Chive, and J. Pribetich, "Complete three-dimensional modeling of new microstrip–microslot applicators for microwave hyperthermia using the FDTD method," *IEEE. Trans. Microwaves Theory Tech.*, vol. 42, pp. 2657–2666, Dec. 1994.
[10] D. M. Sullivan, D. T. Borupt, and O. P. Gandhi, "Use of the finite difference time domain method in calculating EM absorption in human tissues," *IEEE Trans. Biomed. Eng.*, vol. BME-34, pp. 148–157, Feb. 1987.
[11] A. Taflove and M. E. Brodwin, "Computation of the electromagnetic fields and induced temperatures within a model of the microwave-irradiated human eye," *IEEE Trans. Microwave Theory Tech.*, vol. MTT-23, pp. 888–896, Nov. 1975.
[12] D. M. Sullivan, O. P. Gandhi, and A. Taflove, "Use of the finite difference time domain method for calculating EM absorption in man models," *IEEE Trans. Biomed. Eng.*, vol. 35, pp. 179–186, Mar. 1988.

[13] K. S. Yee, "Numerical solution of initial boundary value problems involving Maxwell's equations in isotropic media," *IEEE Trans. Antennas Propagat.*, vol. AP-14, pp. 302–307, May 1966.

[14] A. Taflove and M. E. Brodwin, "Numerical solution of the steady-state electromagnetic scattering problems using the time-dependent Maxwell's equations," *IEEE Trans. Microwave Theory Tech.*, vol. MTT-23, pp. 623–630, Aug. 1975.

[15] G. Mur, "Absorbing boundary conditions for the finite difference approximation of the time electromagnetic field-equations," *IEEE Trans. Electromagn. Compat.*, vol. EMC-23, pp. 377–382, Nov. 1981.

[16] M. Nachman and G. Turgeon, "Heating pattern in a multi-layered material exposed to microwaves," *IEEE Trans. Microwave Theory Tech.*, vol. MTT-32, pp. 547–552, May 1984.

[17] M. Chive, "Use of microwave radiometry for hyperthermia monitoring and as a basis of thermal dosimetry," in *Methods of Hyperthermia Control, Series on Clinical Thermology, Subseries Thermotherapy*, M. Gautherie, Ed. Berlin, Germany: Springer-Verlag, 1990, vol. 3, pp. 113–128.

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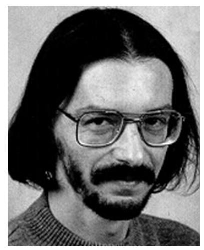

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