An Electric Field Converging Applicator with Heating Pattern Controller for Microwave Hyperthermia

YOSHIO NIKAWA, TOHRU KATSUMATA, MAKOTO KIKUCHI, AND SHINSAKU MORI, MEMBER, IEEE

Abstract —**A new type of applicator for microwave hyperthermia at 430 MHz has been developed. The applicator, which consists of an integrated waveguide array, has a convergent effect of the radiated electromagnetic (EM) field in lossy dielectric media, and it can change the heating pattern in the medium with ease. The electric field distributions from the aperture of the applicator were measured in a saline solution having a concentration of 0.2–0.6 percent which served as the model of simulated human tissues. The results show that when using the above applicator, the location of maximum intensity of the electric field could be generated inside of the dissipative medium. The calculations of temperature distribution were performed in the model by using the experimental results of the electric field distribution. The applicator could change the transverse electric field distribution inside the medium, and then could control the depth of tissues located to the hyperthermic range. The heating range available for hyperthermia using our single applicator is maximum 60 mm in depth theoretically.**

I. Introduction

FOR microwave hyperthermia treatment of cancer, deep and localized heating techniques for the human tissues are sometimes demanded. Using electromagnetic (EM) energy is one of the most effective ways to make a hot spot deep inside the human tissues. To apply microwave EM energy (with a frequency over 300 MHz and less than a few gigahertz) to the human body is one of the most interesting energy-coupling modalities, because the wavelength is moderate (between a few centimeters and around 10 cm) and the energy of the EM wave can be localized on the treatment area inside the human body. Since the depth of penetration into muscle tissues at microwave frequency is small (less than 4 cm), it is difficult to deposit EM energy deep inside human tissues noninvasively without excessive surface heating. Many attempts have been made in this region of the frequency spectrum to increase the penetration depth of the EM energy by a single or a multi-applicator system [1]–[5]. To attain deep heating, using the multi-applicator approach is one of the best means. However, the applicator set up is difficult. Single applicators which can deposit EM energy deep in the dissipative medium with the convergent effect of radiated EM field have been proposed [6]–[8].

This paper describes the study of a new type of integrated waveguide-array applicator for microwave hyperthermia at 430 MHz. Our applicator is a single unit, but it includes a waveguide array inside it. This applicator produces a convergent effect to make the spatial peak of the electric field appear deep in the dissipative medium. Also, the user of this applicator can control the heating pattern in the medium. The location of the hot spot can be changed easily by shifting metal plates in the applicator that are parallel to the E-plane. The distribution of the transverse electric field that is radiated from the applicator was measured in the saline model of human tissue. Furthermore, by using the results of the transverse electric field measurements to apply to a heat transport equation, the theoretical temperature distributions in a model of human tissue with blood flow were simulated.

II. Design of the Applicator

The general view of the applicator is shown in Fig. 1. The applicator is symmetric with respect to the x-z and y-z plane. Fig. 2 illustrates the applicator when viewed along H- and E-plane, respectively. This applicator is a waveguide array type, and the metal plates have been set inside the applicator to separate the aperture of the applicator into four zones. These zones will be called zones no. 1 and zones no. 2. When the propagation mode of the EM wave inside these zones is TE_{10}, the propagation constants in the medium between the metal plates in the applicator are expressed as

$$k_i^* = \sqrt{\omega^2\mu(\epsilon' - j\epsilon'') - \left(\frac{\pi}{d_i}\right)^2} \qquad (i=1,2) \tag{1}$$

$$k_{wj}^* = \sqrt{\omega^2\mu(\epsilon' - j\epsilon'') - \left(\frac{\pi}{d_{wj}}\right)^2} \qquad (j=1,2,3) \tag{2}$$

where ω is the angular frequency of the operating EM wave, μ is the permeability of the medium inserted in the applicator, $\epsilon' - j\epsilon''$ is the complex permittivity of it, and

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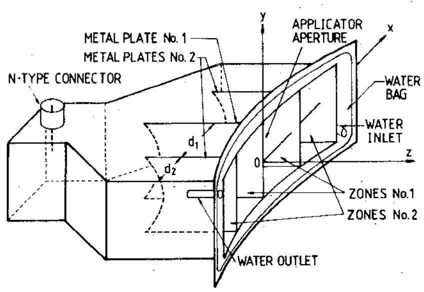


Fig. 1. General view of the applicator.

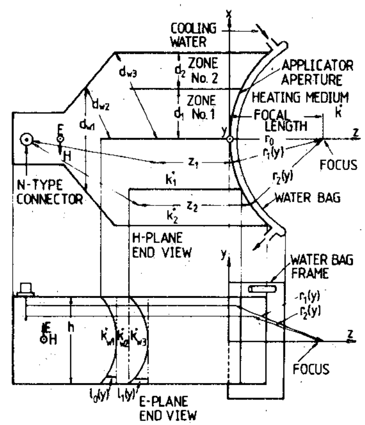


Fig. 2. End view of the applicator along H- and E-plane.

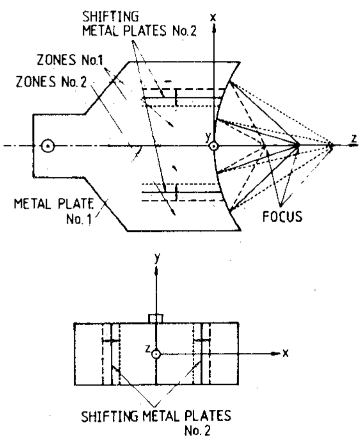


Fig. 3. Schematic of the relation between shifting metal plates and optical focal length.

d_i, d_{wj} are the widths of each zone (zones no. 1 and zones no. 2) and that of the tapered area of the applicator (see Fig. 2). The lengths (z_i), and the widths of the zones separated by the metal plates, are calculated with (3) to have an optical focus when the heating medium is loss free. Equation (3) shows the congruence of the phase of EM wave which is transmitted from the feed point through the zones no. 1 or no. 2 to the focus.

$$[\mathrm{Re}(k_{w1}^*)-\mathrm{Re}(k_{w2}^*)]\,l_0(y)+[\mathrm{Re}(k_i^*)-\mathrm{Re}(k_{w3}^*)]\,l_1(y)$$
$$+\mathrm{Re}(k_i^*)z_i+\mathrm{Re}(k^*)r_i(y)=\mathrm{Const.} \qquad (i=1,2) \quad (3)$$

where k^* is the propagation constant of the heating medium, $r_i(y)$ is the length between the optical focus and the aperture of the applicator, and z_i, $l_0(y)$, $l_1(y)$ are the lengths of the zones separated by the metal plates as indicated in Fig. 2. By shifting metal plates parallel to the E-plane to vary the width between the metal plates (d_i), the focal length r_0 can be changed. The narrower the widths of the zones no. 2, the faster the phase velocity of the EM wave transmitted in the zones no. 2, as a consequence, the focal length becomes shorter. On the other hand, the wider the width of the zones no. 2, the slower the phase velocity of the EM wave transmitted in the zones no. 2, as a consequence, the focal length becomes longer (see Fig. 3). The relation between the width of the zones (d_1) and the focal length (r_0) is shown in Fig. 4. Thus the focus can be changed when assuming the medium to be loss free.

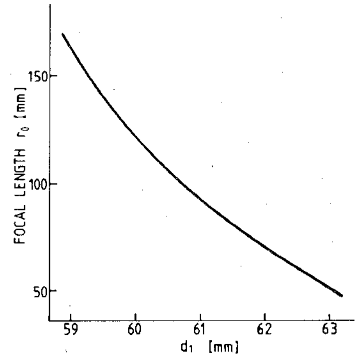


Fig. 4. Theoretical focal length (r_0) versus d_1 of the applicator.

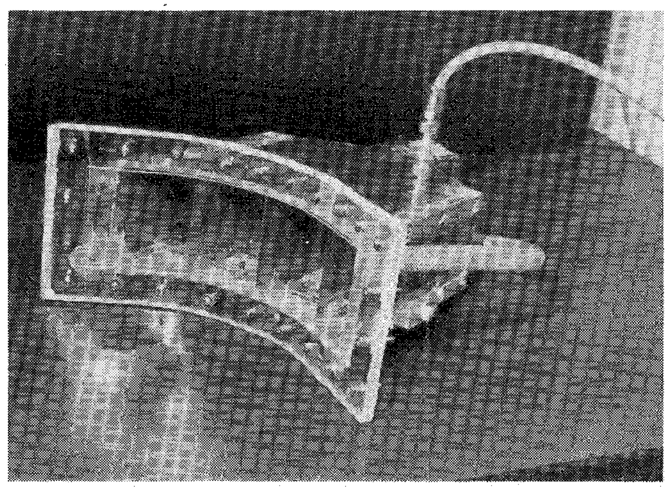


Fig. 5. Photograph of the applicator.

Furthermore, the applicator can be filled with water as a dielectric material whose relative dielectric constant is approximately $78-j1.6$ (at 400 MHz) [9]. By filling the applicator with water, it becomes compact in size and the impedance matching to the directly contacted medium to be heated (such as human tissue) is improved. A water bag is attached on the aperture surface to perform surface cooling of the medium to be heated. The concave-shaped aperture of the applicator is in good contact with a biological body (which can be approximated by a cylinder). The actual applicator, whose aperture size is 213×80 mm is shown in Fig. 5.

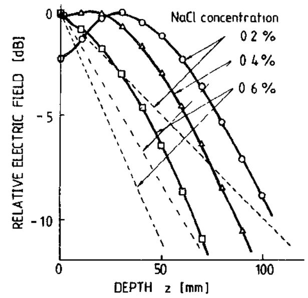


Fig. 6. Relative electric field distributions from the aperture of the applicator along the z-axis inside the saline solution having a concentration of 0.2–0.6 percent. Dashed lines are the distribution of a normally incident plane wave in the saline solution.

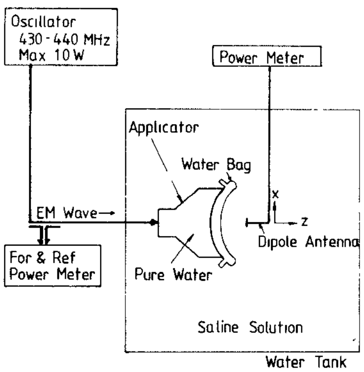
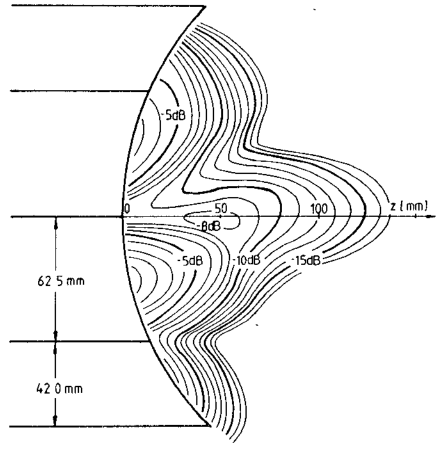


Fig. 7. Block diagram of electric field measurement system in the dissipative medium.

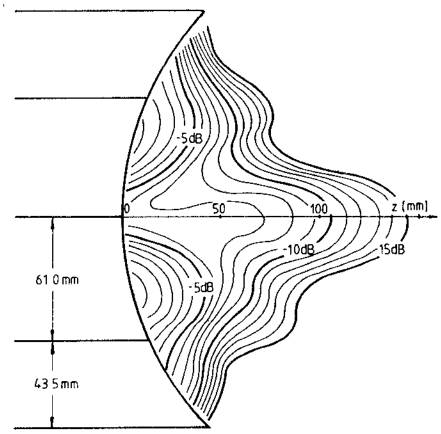
III. Measurements of Electric Field Distribution

Experimental measurements have been made to evaluate the electric field distribution in the medium. In all figures, the z-axis is perpendicular to the array plane and pointing to the forward direction. The measurements were performed along the z-axis inside the saline solution having a concentration of 0.2–0.6 percent such as a model of variable biological substances [10]. The results show the distribution of the plane wave propagated to the medium (see Fig. 6). The two-dimensional measurements of the electric field distribution radiated from the applicator were performed for various focal lengths (r_0) in a 0.2-percent saline solution (see Fig. 7). The complex permittivity of the 0.2-percent saline solution was measured as $79 - j35$ (at 400 MHz, 29.5°C). The reason why we used a 0.2-percent saline solution is that the dielectric loss is considered to be the mean value of muscle and fat whose relative dielectric loss is 60 and 5 at 433 MHz, respectively [11]. The measurement results of the squared electric field distribution are shown in Fig. 8(a)–(c) in the x-z plane (H-plane). The two-dimensional squared electric field distribution measured along the z-axis from the applicator is shown in Fig. 9.

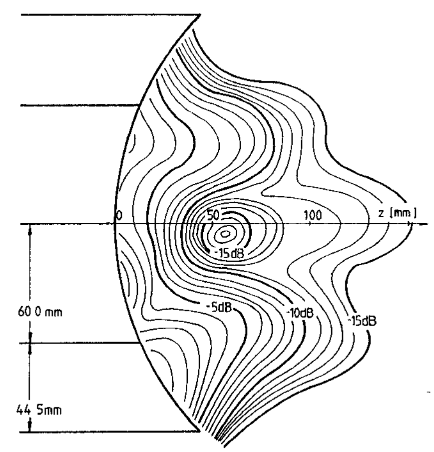
From Fig. 6, we can find that, for a 0.2- and 0.4-percent saline solution, the maximum intensity of the electric field could be generated below the surface of the dissipative medium. When measured in the 0.6-percent saline solution, the penetration depth along the z-axis, from the applicator, is deeper than that of a normally incident plane wave by 20 mm (see Fig. 6).



(a)



(b)



(c)

Fig. 8. Two-dimensional squared electric-field distribution measured in the 0.2-percent saline solution. The focal length settings of r_0 are (a) 69 mm, (b) 100 mm, and (c) 126 mm.

From Fig. 8(a)–(c), it is found that the location of the maximum intensity and distribution of the electric field inside the medium are changeable by shifting the metal

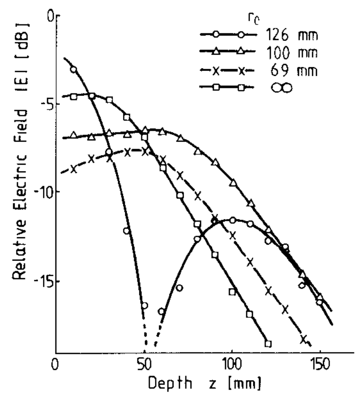


Fig. 9. Relative electric field distributions from the aperture of the applicator along the z-axis.

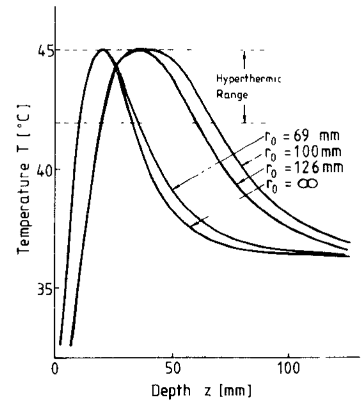


Fig. 10. Theoretical temperature distribution in the model of human tissue with blood flow for the variable set of the focal length of the applicator.

plates which are set inside the applicator. Fig. 9 shows that when we set the focal length (r_0) to 100 mm, the electric field penetration is the deepest in the medium.

IV. The Simulation of Temperature Distribution in the Model of Human Tissues

The effect of blood flow cannot be neglected in the actual use conditions when the medium to be heated is a human body. Therefore, calculations of the theoretical temperature distribution were made by adapting the experimental results of the transverse electric field distribution to the human tissue model with blood flow. The calculation method is using the conventional heat transport equation according to Kritikos *et al.* [12], Ho *et al.* [13], and ourselves [14]. The initial temperature and the temperature of the blood are 36.5°C. The temperature of the surface of the human tissue model is 15.0°C due to external cooling.

The results of the temperature calculation in the model which has a uniform blood-flow rate of normal muscle are presented in Fig. 10. From Fig. 10, we can see the difference among the temperature distributions along the z-axis for a variable set of focal lengths ($r_0 = 69$, 100, 126 mm, and infinity). The case where $r_0 =$ infinity is the same condition that exists in a continuous waveguide that is terminated with a reflectionless load. From these results, it is found that the newly developed applicator could probably heat deep areas of the human body, and could enable to change in the depth of heating. The maximum heating depth could probably be generated by a factor of two, using our applicator compared to a conventional waveguide-type applicator (see Fig. 10).

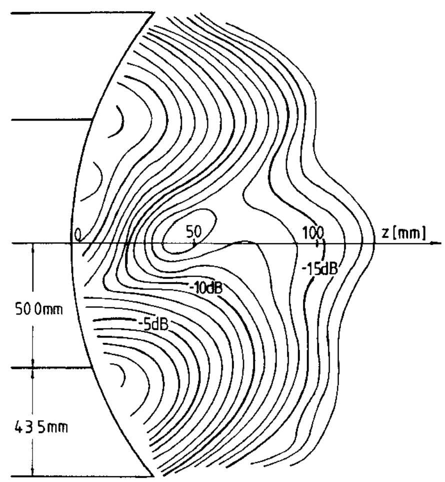


Fig. 11. Two-dimensional squared electric-field distribution measured in the 0.6-percent saline solution using an electric field converging applicator with an aperture size of 192×150 mm.

V. Discussion and Conclusions

An electric field converging applicator with a heating pattern control for microwave hyperthermia (operated at 430 MHz) has been developed. Experimental and theoretical results are presented for the applicator.

The results of transverse electric field measurements in a saline solution, a model of human tissue, show that the newly developed applicator could enable the user to change the location of the maximum intensity and the distribution of the electric field in lossy medium. This is done by shifting the metal plates which are built inside the applicator. These heating characteristics were also studied through the simulation of a living system's effect on temperature distribution. Data were obtained using a saline solution having a concentration of 0.2 and 0.4 percent. The results show that the maximum electric field could be produced below the surface of the lossy medium instead of at the surface. By using a 0.6-percent saline solution, a maximum electric field could not be obtained below the surface, but the penetration depth along the center axis is deeper than that of a normal incident plane wave by 20 mm.

To produce a maximum electric field deep location in a 0.6-percent saline solution as a human muscle model, an applicator of larger aperture size is needed. The measurement result of the electric field distribution in the 0.6-percent saline solution using the electric field converging applicator with aperture size of 192×150 mm, which was made following the design method described in this paper, is shown in Fig. 11. From Fig. 11, it is found that the electric field is not decreased in the region where the depth is over 50 mm and less than 90 mm along z-axis. From this result, deep heating in the human muscle layer may be realized by using our applicator. The problem of the relation between aperture size and the location of the maximum electric field will be discussed later in a separated paper.

Calculations of the temperature elevation in a theoretical human-tissue model with blood flow show that the newly developed applicator could heat deeply, and the depth of heating could be controlled. The maximum heating depth that could be generated with our applicator was over two times the depth that could be obtained using a conventional waveguide-type applicator.

Once the relation between the optical focal length and heating pattern is better understood, it may be possible to set the applicator's metal plates to selectively heat a certain area inside a dissipative medium such as a human body. As is stated above, using our new type of applicator, hyperthermia treatments may be improved because of the control of the EM energy that can be provided to focus electric fields deep within the target tissue.

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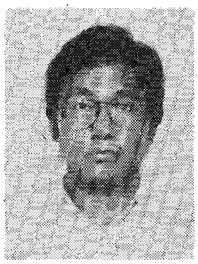

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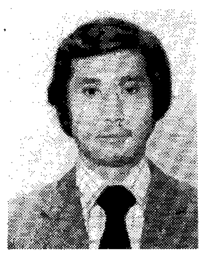

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