Short Papers

Capabilities of Multiapplicator Systems for Focussed Hyperthermia

REINHARD KNOECHEL

Abstract —**Features of the randomly phased scheme and the constructive interference approach for focussed applicator systems are evaluated and compared from a practical point of view. A mathematical expression is given to estimate the depth location of a focal point that depends on the number of array elements. Constraints are indicated that limit the physical construction of such arrays and allow their application only for special purposes.**

I. Introduction

Since localized hyperthermia with microwaves has been recognized as an adjunctive modality in cancer therapy, much work has been devoted to this topic during the last decade [1]–[3]. However, the microwave equipment is still not ready for clinical utilization at large scale. Due to the special problem of irradiating an inhomogeneous, stratified, and lossy dielectric like living tissue with microwaves, and heating to temperatures near to the threshold of irreversible damage, microwave applicators have been extensively studied [4] and are still under consideration.

Apart from the possibility of using a single applicator, attention has been drawn to multiapplicator systems that would allow the production of "focussed" hyperthermia, i.e., a deep seated hot spot, thus avoiding superficial burns of skin and healthy tissue. Essentially two operating schemes are envisaged:

1) a randomly phased multiple-beam antenna or sequential switching of power between a variety of antennas with beams aiming at a focal point [5], [6];
2) constructive interference of the electric fields from an antenna array at the focal point [7], [8].

The applicator arrays could be used in direct contact with the tissue, with air spacing or with a dielectric bolus of tissue-equivalent material. The different approaches are discussed in the literature. However, there seems to be no clear picture of what may be attained by the various heating schemes. Hence, this short paper will discuss the performance of multiapplicator arrays from a practical standpoint and try to answer the question of what heating depth may be obtained with a focal point in terms of the physical structure of the antenna array.

II. Heating Depth of Antenna Arrays

The frequencies to be considered are at the ISM bands (Industrial, Scientific, Medical) 2450 MHz (worldwide), 915 MHz (in USA), and 433 MHz (in Europe), i.e., the wavelengths are between 12 and 70 cm in air. As the lateral dimensions of a focusing array are several wavelengths, the size of such an arrangement is supposed to be impractical when it is driven with air spacing to the tissue surface. Such arrays are discussed theoretically in [8] and will not be considered here.

Arrays of direct-contact applicators with and without dielectric bolus have similar performance and shall now be explained with the help of Fig. 1. A spheroidal surface is sketched with N applicators all aiming at a focal point F. The irradiated material is supposed to be homogeneous, having a damping constant α. The radius R of the sphere is large compared to the size of a single applicator. The applicators are sufficiently far apart from each other and damping is high enough, so that, at the location of a distinct radiator, electric field contributions other than from the applicator itself, may be neglected. Hence, optimum heating depth is reached in the sketched arrangement when the N radiators are identical. A focal point is said to exist when the surface temperature is equal to or less than the temperature in the center.

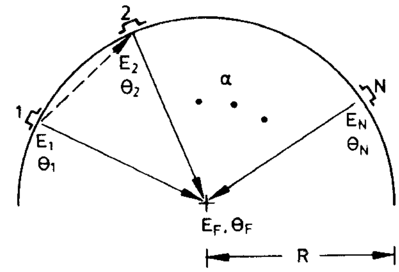

Fig. 1. N applicators on a spheroidal surface.

When the randomly phased approach is used, powers are superimposed at the focal point. The constructive interference system yields summation of the electrical field strength at the center. Thus for the latter, the electric field at the focal point is

$$E_F = N E_0 e^{-\alpha R} \tag{1}$$

where E_0 is the field of a single applicator at the surface, and co-phase driving of the radiators is assumed. Temperature θ is proportional to the irradiated power, hence

$$\theta_F \sim N^2 E_0^2 e^{-2\alpha R}. \tag{2}$$

When, as a limiting case, the surface temperature θ_0 is equal to θ_F, we deduce from (2) as the maximum heating depth with a focal point

$$R_{\max} = \frac{\ln N}{\alpha} = \delta \ln N \tag{3}$$

where δ is the penetration depth of a wave into the medium, i.e., the distance where the electric field drops to e^{-1}. The penetration depth is tabulated in [1] and in [9] for various kinds of tissue. Stuchly and Stuchly [10] tabulate electrical properties in terms of dielectric constant. For the randomly phased system, only half the value of (3) is obtained.

Of course, (3) is only a rough estimate and is valid only for a small number of radiators. In practice, the effect of beam spreading takes place, which is more distinct for small applicators [6] where the radiating cross section is less than a wavelength, and therefore the heating depth should be less than predicted by (3). On the other hand, surface cooling by convection and the dynamics of a living system such as blood flow tend to increase the penetration depth. Thus (3) may serve, as a rule of thumb, to estimate the minimum number of applicators once a certain heating depth is required. As a result, it should be stated, that a

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The author is with Philips Forschungslaboratorium, D-2000 Hamburg 54, West Germany.

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focal point can be produced at 70 percent of the wave penetration depth δ with two applicators and at 110 percent using three applicators. Operating at 2.45 GHz, this yields a focal point at approximately 7 and 11 mm in muscle and 40 and 60 mm in adipose tissue, respectively. The penetration depth of the plane wave has been assumed to be 10 mm (muscle) and 55 mm (fat) [9].

Considering the relatively small heating depths in muscle, one is tempted to reduce the applicator size, for instance, by dielectric loading of rectangular waveguide apertures, and to bunch a larger number of applicators together [6]. However, taking the results of [1] into account, this often proves to be not a practical solution. The reason lies in the fact that, in a majority of irradiated tissues, subcutaneous fat layers are present with thicknesses in the order of 1 cm and more. In [1] is stated that too small an applicator aperture yields excessive superficial heating of the fat layer due to strong near-field effects. To suppress the near field and obtain approximately the penetration depth of a plane wave, the lateral dimensions of an aperture have to be about one wavelength in the medium which is irradiated. That means, having a fat layer with a dielectric constant between 5 and 10 between the applicator and the muscle requires an applicator cross section of about 1/2 to 1/3 the wavelength in air in diameter. At 2.45 GHz, the diameter then is 40–60 mm. Due to these limitations, the utilization of a focusing array of applicators will often be prohibitive in practice. Only in special cases, superposition of the fields from two or three applicators may be possible.

III. Experimental Verification

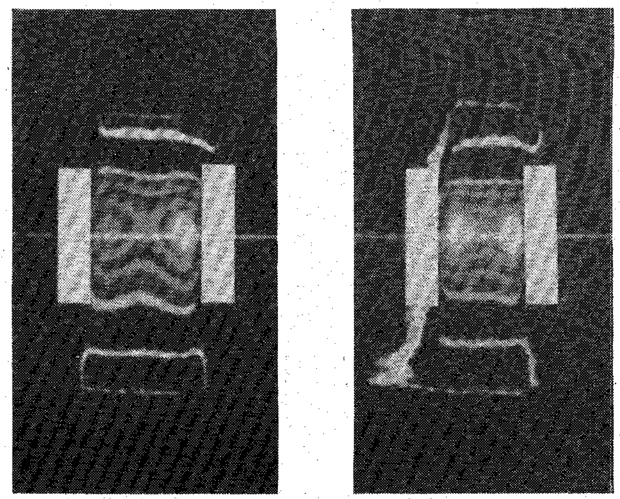
First, a recently published result in [11] shall be mentioned. Using the constructive interference scheme, a heating depth of about 50 mm could be reached in a square block of fat tissue with two applicators at right angles. This result compares well with (3).

Our measurements have been performed to check the validity of (3). The split phantom method and the thermographic camera have been used, as described in [1], to monitor the heating patterns. The muscle phantom material is the same as presented in [1], whereas as fat phantom, pork adipose tissue was used. Soft fat tissue was preferred for the experiments because it was readily available and could be easily cut into a suitable shape for the experiments. Heating was started with large blocks of tissue, which then were cut to smaller dimensions until a focal point could be observed.

In the experiments, all the powers had to be derived from one generator in order to be coherent with a well-defined phase relation. The power-dividing network had to be constructed from low-power components available at the laboratory and was inherently lossy. Low power levels per applicator and, consequently, long heating times resulted, which in principle lead to significant contributions of thermal diffusion to the measured heating pattern.

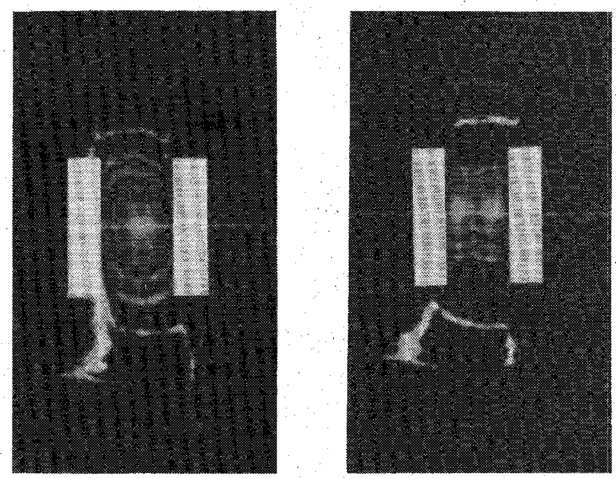
However, since the goal of the experiments was not the characterization of applicators, but the realization of the location of deep-seated hot spots, thermal diffusion effects were supposed to be of minor importance. Thermal diffusion mainly results in larger diameters of the measured spots and hardly in a shift of their location. On the other hand, thermal diffusion is always present in the real world, and, since hyperthermia sessions have a duration of 1/2–1 h, results including diffusion may even give a better picture of what happens in practice.

Measurement results in muscle phantom slabs are shown in the photographs of Figs. 2–4. Ceramic-filled ($Al_2O_3, \epsilon_r = 9.4$) rectangular waveguides (12.7×28.5 mm) were used as applicators.



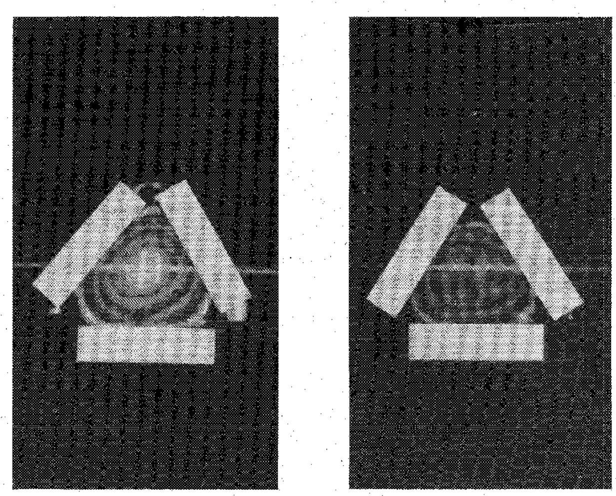
(a) (b)

Fig. 2. Dielectric slab with two applicators in co-phase action. Muscle phantom (a) 3 cm thick; and (b) 2.1 cm thick. Scale: 1.5°C per step of grey scale. Power: 10 W/applicator, $t = 2.5$ min.



(a) (b)

Fig. 3. Dielectric slab from muscle phantom, 1.7 cm thick. Applicators in (a) co-phase and (b) antiphase operation. Heating depth is 0.85 cm. Same scale as before. Power: 10 W/applicator, 2.5 min.



(a) (b)

Fig. 4. Dielectric slab from muscle phantom with equilateral triangular shape (length of side ≈ 30 mm, thickness 50 mm) and three applicators, driven in (a) cophase (b) 0°, 120°, 240° phase difference. Heating depth is 1.2 cm. Same scale as before. Power: 5 W/applicator, 3 min.

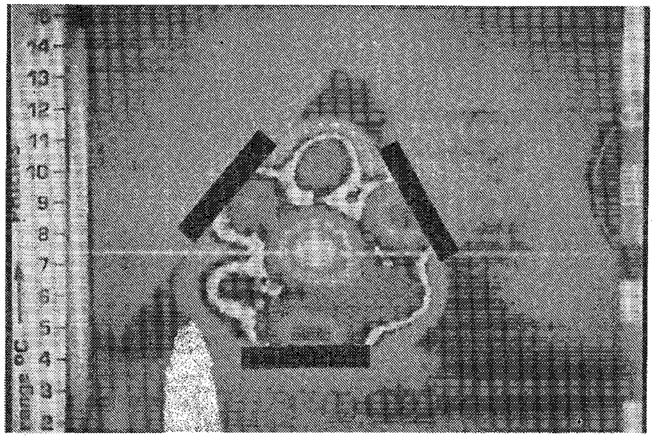


Fig. 5. Heating of adipose tissue with three applicators. The phantom is of equilateral triangular shape with 15-mm length of side and 80-mm thickness. Central focus at 40-mm depth is from interference of three radiators. Adjoining hot spots reflect interaction between two adjacent applicators due to beam spreading. Same scale as before. Power: 20 W/applicator, 3 min.

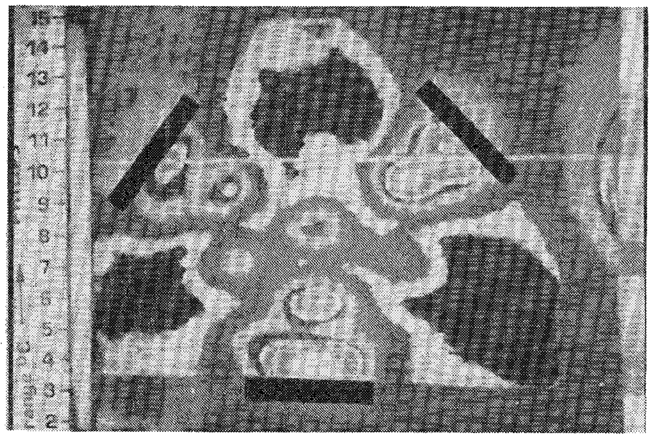


Fig. 6. Heating of a piece of adipose tissue with the same shape as shown in Fig. 5. Length of side is 250 mm. Three applicators are used. A variety of hot spots is demonstrated to appear besides the central spot at 65 mm. Same scale as before. Power: 20 W/applicator, 3 min.

Applicators were driven with 10 W each for 2.5 min. The shapes of the temperature distributions of Fig. 2(a) indicate that nearly independent heating is obtained when two applicators, as depicted by the white bars, are separated by a dielectric slab of 30-mm thickness. Both radiators are excited in co-phase and constructive interference should occur. Reducing the slab thickness of 21 mm (Fig. 2(b)) results in a melting together of both hot spots. Further reduction of applicator spacing to 17 mm (Fig. 3(a)) finally yields a focal point, i.e., as measured from one applicator, the heating depth is 8.5 mm, a result consistent with (3). At the surface, no temperature maximum can be observed because direct contact of the applicators causes surface cooling by heat conduction. To prove the true interference of the two beams, the excitation was switched to antiphase. Fig. 3(b) then shows a temperature minimum in the middle and splitting up in two hot spots for that case.

Performing similar experiments with three applicators of the same size in an equilateral triangle of phantom tissue (50 mm thick) finally leads to the heating patterns of Fig. 4(a) and (b). Power was 5 W/applicator for 3 min. Co-phase driving of the radiators produced a focal point in 12-mm depth, whereas antiphase driving (0°, 120°, 240° phase lag) demonstrated cancellation of fields in the center. Again (3) is verified with good accuracy.

Figs. 5 and 6 illustrate the heating patterns in adipose tissue exposed to three applicators with co-phase driving. Applicators were rectangular waveguides (34×72 mm) filled with teflon dielectric ($\epsilon_r = 2.2$). The measurement specimen was an equilateral triangle with about 150-mm length on a side cut out of adipose tissue of 80-mm thickness. Power was 20 W/applicator and heating time was 3 min.

A focal point is induced in the center, corresponding to a 40-mm heating depth. The presence of further hot spots in the vicinity of the applicators due to superficial heating is also visible. However, there are additional hot spots, although lower in temperature. These maxima of the second order essentially reflect interaction between pairs of adjacent radiators and beam-spreading effects. A check was made, where one radiator was switched off, leading to the joining of the hot spots on the symmetry line between the remaining applicators. A further increase in the number of hot spots takes place when the length of side of the irradiated triangular piece of adipose tissue is enlarged to about 250 mm, as shown in Fig. 6. Beside the superficial heating points and the central focus, now placed at a depth of 65 mm, six adjoining maxima may be detected. Of course, this is partly due to the chosen geometry. However, as a matter of fact, it can be seen that constructive interference of several applicator fields in fat tissue may give rise to the existence of unwanted hot spots in the vicinity of the central focus. This will not occur for randomly phased irradiation, because then no superposition of electromagnetic fields but of thermal heating patterns is used, showing no interference.

Considering the location of the main focus, and bearing in mind that naturally grown pork adipose tissue was used as a measuring specimen with the consequence of somewhat undefined dielectric properties, (3) has been confirmed with fair accuracy. A four-element array has been used in a similar experiment on adipose tissue, using applicators with the same dimensions, arranged around a square block of tissue with a side length of 150 mm. Fair agreement with (3) was observed for the location of the main focus.

IV. Conclusions

For effective deep heating of tissue, multi-applicator arrays may be envisioned. The array elements may either be driven by a variety of randomly phased sources or act in a co-phase mode to obtain constructive interference at a focal point. Physical constraints, due to the absorbant nature of tissue and the presence of layers with high and low permittivity, limit the achievable heating depth severely:

1) The depth of a focal point is proportional to the penetration depth of the wave and the natural logarithm of the number of applicators in the radiating array.

2) A randomly phased array yields only half the penetration depth of an array utilizing constructive interference.

3) However, in contrast to the randomly phased approach, the constructive interference scheme may give rise to unwanted and uncontrolled hot spots in the vicinity of the wanted focal point.

4) Following the design requirements of [1], namely, that the diameter of an aperture must be about one wavelength in the medium which is irradiated, leads to the consequence that a focal point in muscle can only be obtained when no significant fat layer is present. Example: At 2450 MHz ($\lambda = 12.25$ cm in air), following [1], leads to an applicator diameter of about 1.8 cm when muscle is irradiated directly, and to about 5-cm applicator diameter when a fat layer is present. These sizes should be compared to the attainable depth of a focal point, which is approximately 10 mm in muscle.

5) The feasibility of a focal point appears to be only realistic when adipose tissue (breast cancer) is irradiated. Because of the large lateral dimensions of the applicators, only two or three radiators can be grouped with reasonable size. As already mentioned, the applicator size is approximately 5 cm at 2450 MHz. At 915 MHz and irradiating fat, the applicator diameter would be about 13 cm (corresponding to a penetration depth of 14 cm [1]). Making the applicator dimensions smaller than these values would result in a rapid decrease of penetration depth, thus losing the benefits of the lower frequency. The same remarks are valid for the frequency of 433 MHz, where the required applicator diameter would be 30 cm.

These results have been checked with fair accuracy by experiments using muscle phantom and pork adipose tissue. Deviations from the above statements are likely to occur for the *in vivo* situation, because of the presence of the dynamics of a living system, like, for instance, blood flow. However, it is assumed that this could only affect the effective penetration depth of the used frequency by surface cooling effects of the blood stream and heat convection, and heat conduction to the interior. The principal features of the various systems should remain unchanged.

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Field Theory of Planar Helix Traveling-Wave Tube

DEVI CHADHA, SHEEL ADITYA, MEMBER, IEEE, AND RAJENDRA K. ARORA, SENIOR MEMBER, IEEE

***Abstract*—A pair of unidirectionally conducting screens, conducting in different directions, constitute a planar helix. The planar helix is proposed as a slow-wave structure for application in a traveling-wave tube (TWT). Field theory is applied to analyze the behavior of the planar helix in the presence of a flat electron beam present between the two screens. Results indicate the presence of three modes, with one mode having a negative attenuation constant, as in the case of the usual helix-type TWT. Curves are shown for a typical proposed planar TWT. Also, the effect of beam current is indicated.**

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The authors are with the Department of Electrical Engineering, Indian Institute of Technology, Hauz Khas, New Delhi, 110016, India.

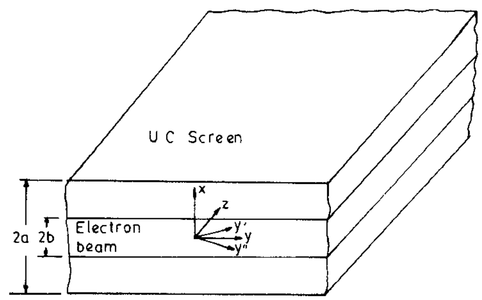

Fig. 1. The planar traveling-wave tube: z—direction of propagation; y' and y''—directions of conduction of top and bottom UC screens, respectively; θ—helix angle.

I. Introduction

The analysis of a helix-type traveling-wave tube (TWT) has been carried out by Pierce [1], Chu and Jackson [2], and others [3], [4]. In a TWT, a slow-wave structure such as a helix or coupled cavity is used to slow down the electromagnetic wave to the velocity of the electron beam so that a strong interaction between the two can take place. A slow-wave structure in planar geometry, consisting of a pair of parallel unidirectionally conducting (UC) screens conducting in different directions and separated by some distance, has been studied by Arora [5], Aditya [6], and Aditya and Arora [7]. It appears that this planar slow-wave structure can be used in a TWT. In view of this, a planar helical TWT is analyzed in the present text. Field equations are derived and the modal solution of the problem is obtained. The variation of the complex propagation constants of different modes with beam voltage is studied.

II. Configuration

The planar TWT considered here for analysis consists of a pair of UC screens (Fig. 1) separated by a distance $2a$ in the x-direction. As shown, the top and the bottom screens conduct in directions y' and y'' which, respectively, make angles θ and $-\theta$ with the y-axis. An electron beam of thickness $2b$ is symmetrically located between the two screens. Both the screens and the beam are assumed to be of infinite extent in the y-direction. As pointed out in [5], limiting the structure in the transverse direction will not disturb the wave in regions remote from the ends, and results obtained here will be applicable to a structure several wavelengths wide. A practical structure can be terminated in the y-direction by closing the "helix" by conducting wires. The beam current density is constant over the cross section and has only a z-component. In practice, this assumption is very nearly realized by means of the focusing action of a strong dc magnetic field applied in the z-direction.

III. Wave Equations and Electron Dynamics

The boundary conditions require both TE and TM modes to be present. If the fields are assumed to be varying as $\exp(j\omega t - \Gamma z)$, and are constant in the y-direction, i.e., $\partial/\partial y \equiv 0$, then the field equations can be written in the following form.