Phased-Array Design Considerations for Deep Hyperthermia Through Layered Tissue

P. A. CUDD, A. P. ANDERSON, M. S. HAWLEY, AND J. CONWAY

Abstract —Results are presented which demonstrate localized heating at depth, by a phased array in a homogeneous thorax phantom and the problems caused by a more realistic case of a layered tissue equivalent phantom. A phased array of contacting radiators is proposed for overcoming the difficulty of selective heating within the body cavity caused primarily by the muscle layer. The field from one aperture radiator in contact with layered tissue is predicted by a planar spectral diffraction algorithm incorporating transmission and reflection operations on the plane wave spectrum. This prediction process is validated by experimental results. The algorithm enables the prediction of a minimum number of phased contacting radiators required for selective heating within lung tissue through fat and muscle layers at 2.45 GHz, and provides a guide for the design requirements of a multiapplicator system.

I. INTRODUCTION

THE cytotoxic effects of heat on tumor cells raised to temperatures above the body norm have been published in the literature [1], [2]. The reduced malignant cell mortality for temperatures below 43°C may be critical for the efficacy of heating. An important consideration in hyperthermia is damage to healthy tissues which must be contained. For the safety of the patient, regional and whole body hyperthermia are subject to a temperature limit which is lower than that necessary for localized hyperthermia. Consequently the latter approach is preferable for maximizing tumor cell toxicity. Prognosis is improved by taking advantage of synergistic effects when hyperthermia is used in combination with radiotherapy or chemotherapy [2], [3].

Microwave applicators are the natural choice for selective heating near the surface, but there is now wide agreement that deep heating from a single antenna applicator is unlikely to be successful [4], [5]. The phased-array approach [6], [7] has been shown to offer the possibility of overcoming this limitation and its feasibility has been demonstrated by field measurements in a homogeneous phantom filled with a lung-equivalent dielectric [7]. The array antenna/thorax phantom system used in [7] is shown diagrammatically in Fig. 1(a) where the homogeneous phantom consisted of wet sand ($\epsilon_r = 22$, $\tan\delta = 0.19$) with 20-mm spaced probing holes. A sleeve antenna encapsulated in expoxy resin and titanium dioxide to match the phantom material was used to probe the field. Fig. 1(b) shows a typical field pattern obtained from this configuration. Such a result is unthinkable for single applicators. Investigations elsewhere have endorsed the favorable prospects for using phased-array radiators in one form or another [8], [9]. The focused applicator design study described in [10] considers focused beams and is not strictly a phased array, nevertheless, the results are encouraging in terms of penetration. Results from a two-dipole array [11] are also encouraging although the low frequency used (~ 400 MHz) results in fairly uniform heating of the whole torso. Not all investigations have been so encouraging. Results presented for a muscle phantom [12] seem to have limited practical value given the size of the phantom although the result for fatty tissue may have practical application to hyperthermia of the breast. The flaw in all the above investigations is the consideration of homogeneous phantoms. Results obtained from simple phantom models are not representative of the reality of different tissues, their associated shapes and boundaries, which will result in the electromagnetic propagation and power deposition rate being quite different in each tissue type. However, tissue equivalent phantoms with detailed structures simulating the human anatomy are difficult to construct and have limited value for hyperthermic evaluation, due to physical variations in normal and tumor tissue structures and also because blood flow effects are absent.

Tissue profiles are also a major problem for theoretical predictions since they have no simple geometrical shape. Yet boundaries between tissues of divergent dielectric properties may produce localized hot spots and cannot be ignored. Analytical solution of a layered cylindrical structure has recently been given [13] where the aperture elements are considered as one-dimensional. Extension of this theory to arbitrary sections with two-dimensional apertures would appear to be formidable.

Theoretical studies of selective heating within a typical body cross section must consider two very difficult problems. The first is the determination of the electromagnetic power deposition within the heterogeneous tissue volume, and the second involves the bio-heat equation which will give the actual temperature distribution within this structure. Attempts are being made to solve some of these problems by using mesh techniques [11], [14]. Such tech-

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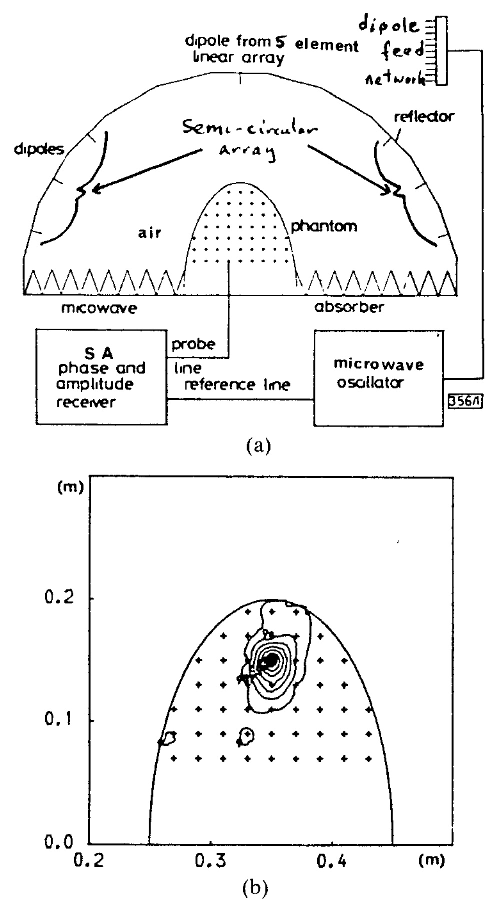

Fig 1. (a) Schematic diagram of the arrays of printed dipoles around the thorax phantom. (b) Measured electric field at one plane within the phantom. Horizontal and vertical spacing between sample points: 20 mm; + sample point.

niques are based on the validity of small sampling intervals ($\sim \lambda/10$), since larger sampling intervals may leave hot spots unpredicted. The CPU time necessary to find the two-dimensional temperature distribution using a mesh technique [15] is already considerable with a coarse sampling interval of approximately $\lambda/4$, thus serious limitations could arise when applied to the clinical situation. Moreover, these techniques have not provided the "inverse solution" for the excitations required by a multi-element array to achieve a certain heat distribution. Deep hyperthermia will be restricted by the capability to penetrate the layers constituting the body wall, and it is this fundamental limitation on selective heating which must be overcome.

The approach proposed in this paper is to calculate and superimpose the fields from a number of contacting radiators strategically located to achieve the best localized deep heating. Before describing the spectral prediction technique for determining the requirements from individual radiators for heating within the body cavity, we present the experimental results pertaining to this major problem of surrounding layers.

II. The Shielding Effect of Muscle Layers

The concept of using a number of remote radiators to concentrate heat in a given region within the body is clinically appealing. However, muscle tissue presents a difficult propagation path, and, in the form of a surrounding layer as in the thorax or abdomen, poses an insuperable problem for remote radiators. With the same configuration as in Fig. 1(a) temperature measurements were made using thermocouples and calibrated graph plotters instead of the sleeve antenna and receiver. Any perturbing effect of the thermocouples on the microwave field was avoided by removing them during the heating process. Measurements were taken from the homogeneous phantom and a second phantom of the same overall size containing a 10-mm covering layer of fat[1] ($\epsilon_r = 5.5$, $\tan\delta = 0.2$) over 10-mm of

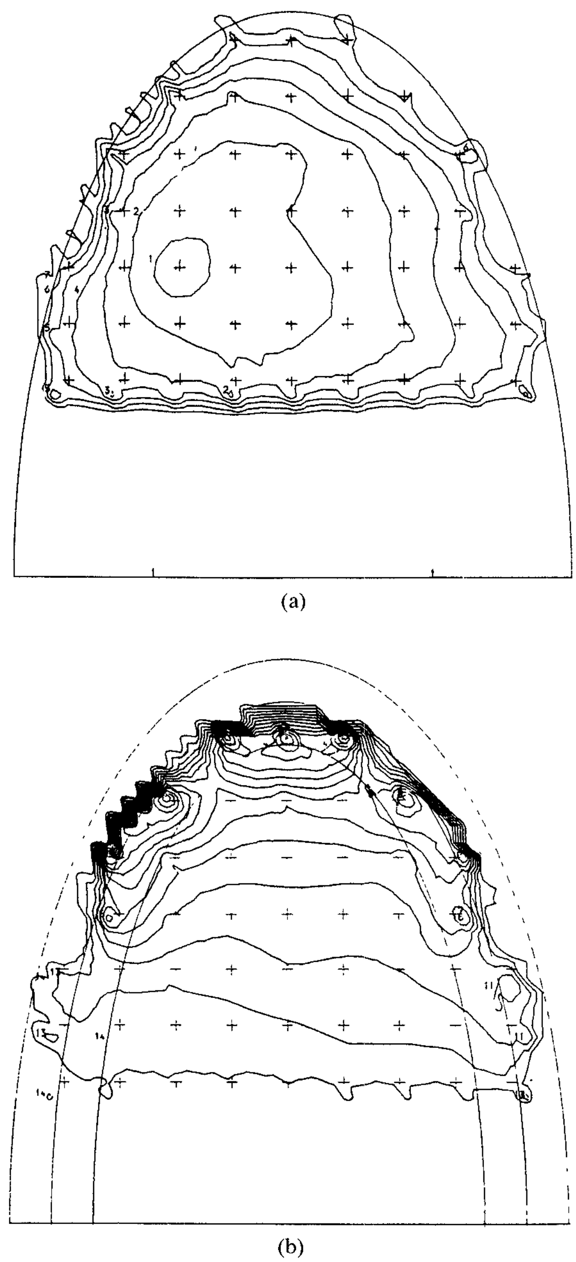

Fig. 2 (a) Measured thermal pattern in a homogeneous phantom obtained from 11 radiators (Freq. = 2.45 GHz) $T_{max} = 33.1°C$; 0.5°C contours; spacing between sampling points as in Fig 1(b). (b) Measured thermal pattern in an inhomogeneous phantom obtained from 11 radiators (Freq. = 2.45 GHz) $T_{max} = 39.1°C$; 0.5°C contours; spacing between sampling points as in Fig. 1(b).

[1]Guy recipe for fat and muscle.

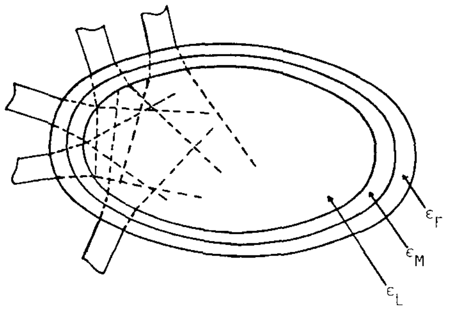

Fig. 3. Schematic diagram indicating the radiation from contacting waveguide elements into a layered thorax section.

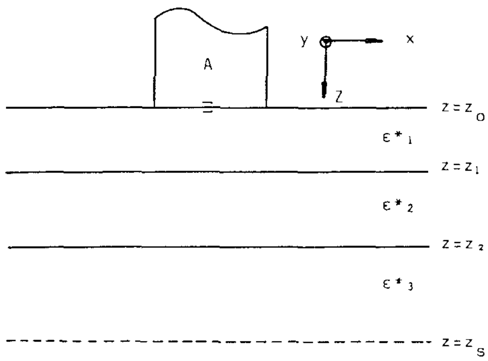

Fig. 4. Schematic showing the coordinate system used for the propagation process in a layered structure. A—large dimension of aperture. ϵ_i^*—complex permittivity of ith medium; origin of coordinate system.

muscle ($\epsilon_r = 47.0$, $\tan\delta = 0.35$), on the lung equivalent[2] core ($\epsilon_r = 22.0$, $\tan\delta = 0.19$). Figs. 2(a) and (b) show the measured isotherms for the homogeneous and layered phantoms interpolated from the 20-mm sampling grid. It was found impossible to achieve a focal region in the core material of the layered model and maximum temperatures were observed in the muscle layer.

The need to overcome surface layer problems has been approached by considering arrays of interstitial implants [16]. Although this idea should be effective, it would appear to be clinically inappropriate in many cases. The failure of the array of remote radiators to produce deep heating provoked the consideration of an array of small contacting rectangular apertures (dielectrically loaded waveguide) to reduce superposition of array element field contributions in the muscle layer. To assess the efficacy of a phased contacting array, the performance of a small aperture element radiating into layered media must be specified. It should be pointed out that the elements we envisage have quite different objectives to those considered in [4]. In addition to the requirement of a confined heating pattern in the muscle layer, the element should contribute a field component capable of deep focusing from a sufficient number of aperture locations on the surface. The situation is shown schematically in Fig. 3 for fat and muscle layers surrounding a homogeneous interior. The use of small apertures to achieve this goal incurs the acceptance of some additional attenuation that would be avoided by single aperture optimization [4].

III. Single Radiator in Contact with Stratified Tissue

With reference to the coordinate system shown in Fig. 4, we first consider a waveguide aperture in contact with a homogeneous dielectric half-space ($\epsilon_1^* = \epsilon_2^* = \epsilon_3^*$) and the propagation of a wavefront from the aperture plane $z = z_0$ to a parallel plane $z = z_s$ ($z_s > z_0$). The aperture excitation is considered to be the dominant rectangular waveguide mode, i.e., TE_{10}, propagating the y-polarized electric field distribution E_{y0}.

Denoting the two dimensional aperture at the plane $z = z_0$ by $E_y(x, y; z_0)$ and the resulting field distribution on the plane $z = z_s$ by $E_y(x, y; z_s)$ then the forward propagation process ($z_0 \rightarrow z_s$) can be expressed in the Fourier domain as follows:

$$\tilde{E}_y(x, y; z_s) = \exp(jkm|z_s - z_0|)\tilde{E}_y(x, y; z_0) \quad (1)$$

where, for a lossless dielectric

$$m = \exp jk\sqrt{1-(\lambda S_x)^2-(\lambda S_y)^2} \quad (2)$$

$$k = \frac{2\pi}{\lambda}. \quad (3)$$

S_x, S_y are the coordinates in the spectral domain, and ~ indicates the Fourier transform. For the case of a lossy medium, propagation is dependent on and calculated from the complex propagation constant:

$$k = \frac{2\pi}{\lambda}\left(1 + j\frac{\epsilon''}{\epsilon'}\right)^{1/2} \quad (4)$$

where $\epsilon_i^* = \epsilon' - j\epsilon''$ is the complex permittivity in each medium. The calculation procedure is simplified by the approximation that the spectral decomposition for lossless media ((1) and (2)) applies.

For tissue containing layers of different dielectric, as shown in Fig. 4, the propagation filter in the Fourier domain may be applied incrementally [17] and modified at each boundary. We assume that, as in free space, the wavefront can be decomposed into an angular spectrum of plane waves. Each plane wave can be associated with a Fourier component, for which the wave normal makes an angle Θ with the z-axis, where

$$\cos\Theta = \sqrt{1-(\lambda S_x)^2-(\lambda S_y)^2}.$$

At any boundary the constants in the propagation filter which characterize the dielectric are changed to account for propagation in the new medium. This modification also includes refraction. The transmission coefficient for each plane wave component can be calculated for the appropriate Fresnel expression which then multiplies $\tilde{E}_y(x, y; z_s)$. The appropriate reflection coefficient is used to include the reflected field contributions. However, contributions must be added in the space domain.

In a finite, layered dielectric, multiple reflections and transmissions occur at each boundary. Since plane wave

[2] Water-saturated sand.

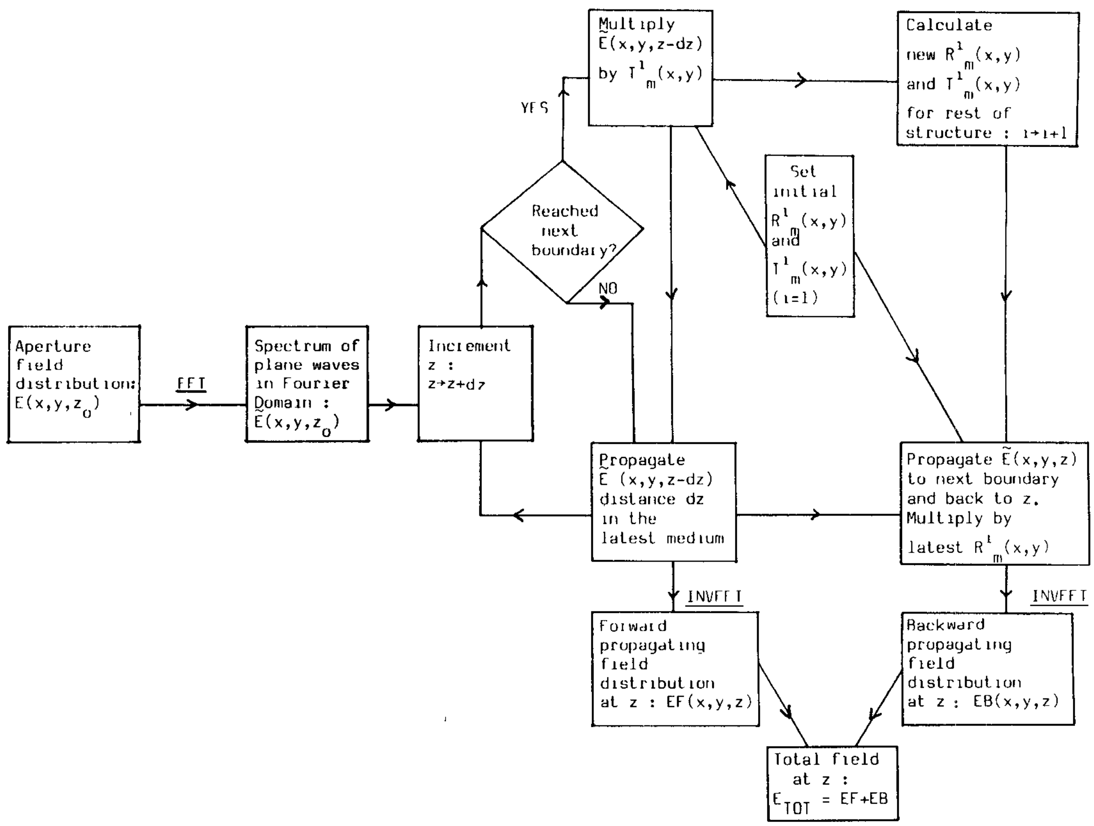

Fig. 5. Flow diagram of the spectral prediction process.

components in the Fourier domain are available, it is possible to apply the cascade transmission line analog for plane waves to account for the multiple reflections and transmissions. The cascade technique provides a rapid calculation of the reflection coefficient R^1_m at the first boundary which includes any contributions from all the subsequent boundaries. It is necessary to calculate the R^i_m for each successive boundary and each angle of incidence. The transmission coefficients which correspond to the R^i_m can be obtained by applying the boundary conditions. A flow diagram of the whole process is given in Fig. 5. This technique, which is readily applicable to several boundaries, has been compared to previous results for a single boundary [4] and found to be compatible.

Since a diffraction model for a probe in contact with tissue was already under consideration by the authors for a 5-GHz radiometer [18], the experimental assessment of the theoretical prediction was obtained using a 5-GHz dielectrically loaded waveguide probe in water. Radiated field measurements were made with a monopole mounted on an automatic scanner and connected to an HP network analyzer. The results in Fig. 6 relate to a homogeneous water phantom ($\epsilon' = 74.0$, $\epsilon'' = 22.8$) showing the predicted and measured E_y field components at three distances from the aperture. Fig. 7 shows the spreading effect on the field pattern produced by a 1-cm fat equivalent layer ($\epsilon' = 5.0$, $\epsilon'' = 0.8$) introduced between the aperture and the water. It should be noted that the three distances at which E_y is determined in Fig. 7 are measured from the fat-water boundary.

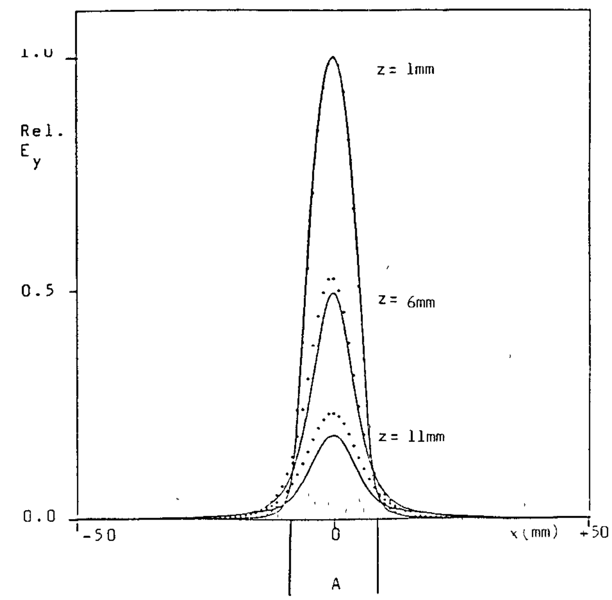

Fig. 6. E_y amplitude (at 5 GHz) in a water phantom ($y = 0$ plane). Values normalized at $z = 1$ mm. —— predicted values. • measured values. A—large dimension of waveguide.

The attenuation in the different media and the size of the heated volume obtainable from a phased array system suggest that an optimum radiation frequency other than one of the ISM standards may be desirable in some cases. The objective of selective heating in the lung will require values in the region of 2 GHz. For the nearest ISM frequency, 2.45 GHz, Fig. 8 shows the predicted field

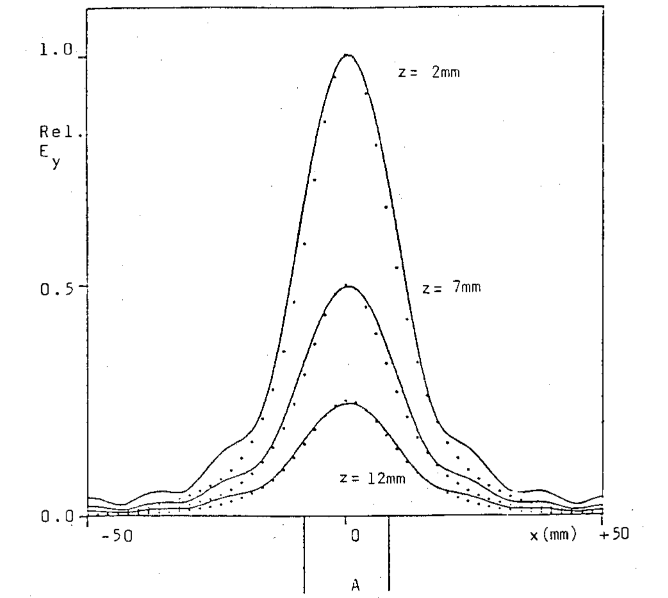

Fig. 7. E_y amplitude (at 5 GHz) in water for a fat/water phantom ($y = 0$ plane). Values normalized at $z = 2$ mm. —— predicted values. • measured values. A—large dimension of waveguide.

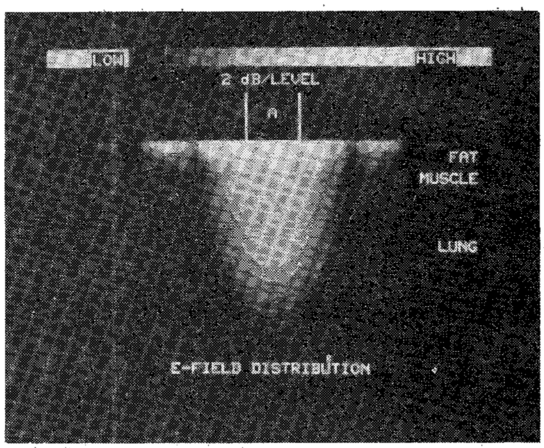

Fig. 8. Predicted E_y amplitude distribution ($y = 0$ plane) at 2.45 GHz in a layered tissue model.

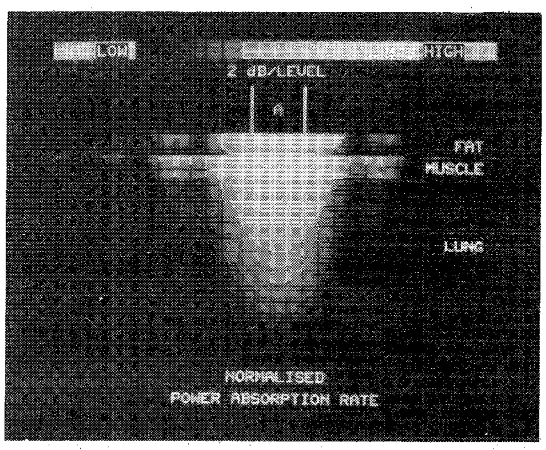

Fig. 9. Relative power absorption rate pattern for a single contacting aperture. Scale as Fig. 8.

distribution resulting from a single 2.5×1.5-cm aperture in contact with a tissue equivalent layered dielectric (i.e., 1 cm of fat, 1 cm of muscle, and 8 cm of lung). The field values obtained for each region of Fig. 8 are converted into a

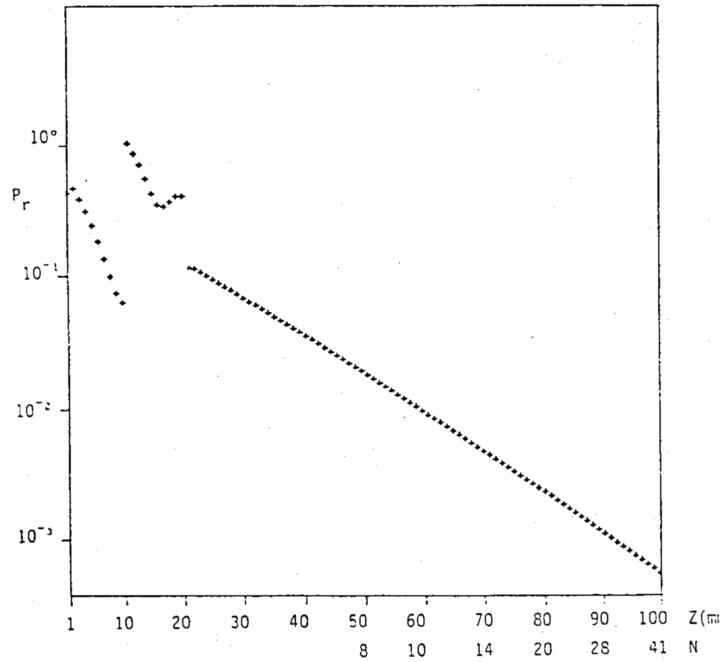

Fig. 10. Profile of on axis normalized P_r versus distance into tissue and N.

relative power absorption rate P_r (where power absorption rate $P = \omega\epsilon'' E^2$). The result of this process is shown in Fig. 9 which indicates the localized high dissipation region in the muscle layer. The quantified pattern for P_r, therefore, provides essential information for the design, configuration, and excitation of an ensemble of contacting elements.

IV. Deep Heating Capability of a Multielement Phased Array

The success of the array approach for localized deep heating is dependent on a sufficient number of coherent contributions from suitably phased elements superimposed in the specified region. In the present case, the permissible contribution from one element is limited by each muscle region "hot spot." Fig. 10 shows the profile through the central region of the P_r pattern from which an indication of the number of radiators required to achieve focal heating equivalent to the maximum P_r beneath each radiator in the muscle layer can be obtained. On the basis of a spherical layered phantom (equal contributions from each element) it is possible to calculate the number of (equiphase) sources N necessary for a given focal depth within the lung tissue. Power absorption at the focus is proportional to N^2. If the minimum requirement is focal P_r equals maximum P_r beneath a radiator, then for this simple example only, the z-axis is calibrated in terms of N.

The practical implementation of this phased-array approach will generate problems in specific cases. Hot spots in other regions due, for example, to bone-tissue interfaces may occur and require investigation. Surface heating will occur due to the high-power deposition in the skin, therefore, surface cooling will be necessary. If circulated water bags are used to provide skin cooling, the water layer will also attenuate the z-components of the E-field near the aperture (which the present model excludes). Hence, the problems of near-field heating associated with small apertures will be avoided in the practical case. By adding a 1-cm-thick layer of water and 1-mm-thick layer of skin to the model represented in Figs. 8 and 9, the appropriate values of element contributions and N_s can be determined from the prediction algorithm.

V. Conclusions

The layered tissue of the body wall, particularly the muscle layers, present an impediment to deep heating of the interior. Our experiments have shown that a phased array of remote radiators produces maximum heating in the muscle wall of a thorax phantom and cannot produce focal heating in the interior for any phased excitation of the array elements. The high rate of absorption in the muscle layer is a limiting factor which can be mitigated by our array of surface contacting applicators designed to provide coherent superposition of field in the interior. It has been shown that, for our fat/muscle/lung model, a minimum of eleven 2.45-GHz applications are required for the focal power absorption rate in the lung to equal that in the muscle under one applicator. This number will be strongly influenced by muscle thickness. The required contribution from each applicator is calculated by a spectral diffraction/cascade algorithm for a radiating aperture in contact with stratified tissue. In a specific case the contribution would be calculated for each chosen site on the surface. These investigations support the proposition that deep heating within the body cavity is feasible provided there is sufficient interelement spacing to prevent excessive muscle heating.

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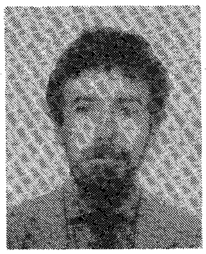

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