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An Ultrasonic Phased Array Applicator for Hyperthermia

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Abstract—**An ultrasonic phased array applicator for hyperthermia provides electronic steering of the sound beam rather than mechanical movement of the transducer assembly. An applicator consisting of a stack of linear phased arrays is examined. The effects of various design parameters, including individual array height and length, are discussed.**

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I. Introduction

One of the major limitations of clinical hyperthermia systems today is inadequate control of the energy deposition used for maintaining a therapeutic temperature. Ultrasound, in the 0.2 to 3.0 MHz frequency range, has several properties which favor its use in clinical hyperthermia. Due to its short wavelength and favorable absorption characteristics, ultrasonic energy may be focused into a small region while achieving a greater depth of penetration than electromagnetic energy [1].

Conventional systems for deep localized ultrasound hyperthermia include focused transducers or several unfocused transducers arranged such that their fields overlap within the tumor volume. None of the systems developed to date are capable of three-dimensional focal region placement without some type of mechanical movement of the transducer(s) [2]. The use of an ultrasound phased array applicator for deep localized hyperthermia is an alternative approach with several advantages over those employing mechanical scanning. A phased array allows

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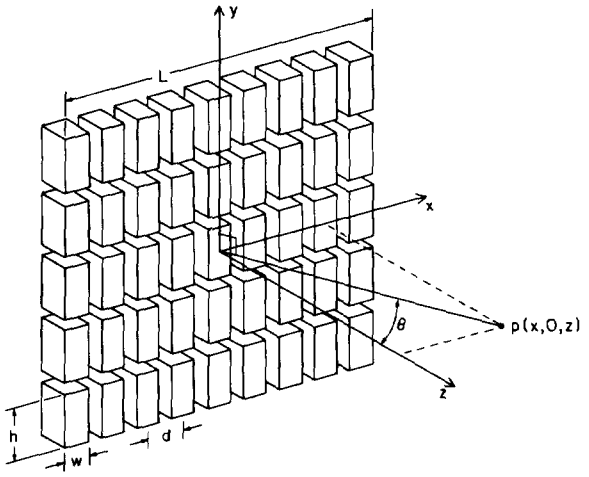

Fig. 1. Stack of five linear phased arrays. Each array has height h along y-coordinate direction and length L along x-coordinate direction. Elements of each array have width w and center-to-center spacing d. Point P represents the focal point for the third array, which is steered off axis by the angle θ.

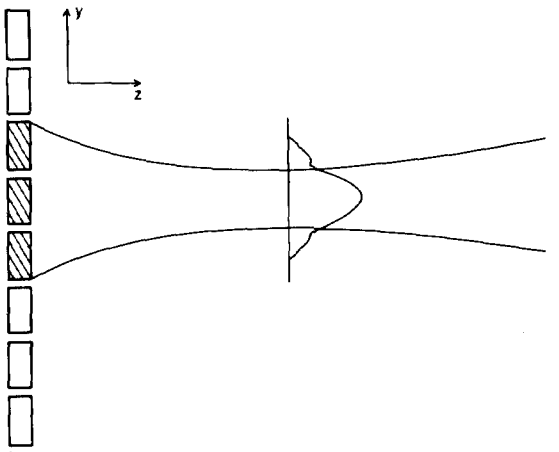

Fig. 2. Intensity profile from a group of three linear arrays excited simultaneously.

electronic scanning of a focused beam without movement of the transducer assembly. The focal region of a phased array can be swept much faster than that of a mechanically scanned system. This enables the phased array to provide more precise control over heating in the treatment field since its scan path can be changed quickly to maintain the desired normal and tumor tissue temperatures.

The disadvantages of using phased arrays in general, and a two-dimensional array of elements in particular, are the complexities of phase control, the large number of amplifiers and associated phasing circuits required, and the need for complex fabrication techniques to produce many small transducer elements. These complications can be reduced if, instead of a complete two-dimensional array of elements, a series of stacked linear arrays is employed.

II. Analysis of Stacked Linear Array

An ultrasonic hyperthermia applicator consisting of a system of stacked linear phased arrays is shown in Fig. 1. The system achieves focal placement in three dimensions by using the linear phased arrays to control placement in the x-z plane and by switching among arrays to control placement in the y direction.

In the development of a stacked linear array the narrowing of the ultrasonic beam in the near-field–far-field transition region must be considered. For a rectangular ultrasonic source, the minimum 3-dB beam width in this region is approximately one third the size of the source in the same dimension. Thus, if a group of three of the stacked arrays are excited at one time, each with the same set of continuous wave (CW) drivers, the 3-dB beam width in the y direction at the near-field–far-field transition would be approximately equal to the height of one array, as shown in Fig. 2.

Scanning in the y direction can be achieved by turning off an array at one side of the group of three excited arrays and exciting the adjacent array on the other side. Since the 3 dB points overlap, the time-averaged intensity between the scan positions is approximately the same as at the center of each of the scan positions so that a uniform time-averaged intensity is achieved. Alternatively, the excitation could be shifted as rapidly to any other group of three arrays.

The design of a stacked linear array system requires the specification of the desired treatment volume (range of depths and cross-sectional dimensions), the desired treatment depth, and the desired system gain in time-averaged intensity at the tumor relative to the body surface. The specification of these parameters, along with the choice of an appropriate operating frequency, define the major dimensions of a stacked linear array system.

The system operating frequency is chosen so that the rate of heating in the tumor region is maximized relative to heating in normal tissue surrounding the tumor, including the body surface. The optimum choice of frequency is dependent upon the tumor size and depth and must consider heating behind, as well as in front of, the tumor.

The height h of each array in the stack is dependent upon the desired treatment depth and operating frequency. Choosing a stack of three arrays to have its near-field–far-field transition distance equal to the distance of the center of the tumor d_T gives the realation

$$d_T = (3h)^2/4\lambda, \tag{1}$$

where λ is the wavelength in the tissue.

The desired treatment volume, the time-averaged intensity gain, and the operating frequency determine the length of the arrays. The system of stacked linear phased arrays must provide a time-averaged intensity gain such that the intensity averaged over a complete scan of the tumor is greater at the tumor than at the body surface and other locations within normal tissues. This gain G_{TA} is defined as the ratio of the time-averaged intensity at the tumor center to the time-averaged intensity at the body surface. This assumes, for this preliminary study, that the time-averaged intensities at the body surface and at the tumor are uniform in the x direction. A G_{TA} greater than one is obtained by making the stack of arrays long compared to the tumor width, as illustrated in Fig. 3.

A simple approach to calculate the length of the stack of arrays necessary to selectively heat a tumor of a given size is to consider the desired value for G_{TA} and to approximate the loss over the irradiation path due to attenuation. The attenuation coefficient of the tissue A (expressed in dB per cm) is multiplied by the path length to the center of the tumor d_T to find the loss due to attenuation. This loss is added to the G_{TA} (expressed in dB) to yield the required time-averaged system gain G.

$$G = G_{TA} + Ad_T. \tag{2}$$

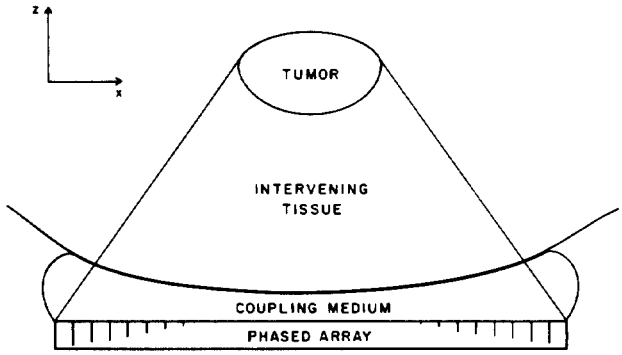

Fig. 3. Linear array configured for tumor treatment.

The array length L is determined from the following relation

$$L = D \cdot 10^{(0.1\,G)}, \tag{3}$$

where D is the diameter of the tumor.

This approximate calculation does not consider the increased path length from the ends of the array to the tumor, the directionality of the elements, nor the phase quantization, all of which affect the system performance. These factors can be considered by theoretically calculating the actual intensity field produced by the array. For this purpose, a computer program was developed which divides the elements of the array into subelements that are small enough so that their fields can be represented, in the region of interest, by the far-field approximation. The contribution to the acoustic field from an individual subelement of height Δh and width Δw centered at (x', y') is given by

$$dp = U_0 \frac{j\rho c k \Delta h \Delta w}{2\pi R} e^{j\beta} e^{-(A+jk)R} \operatorname{sinc}\left(\frac{k(y-y')\Delta h}{2R}\right) \cdot \operatorname{sinc}\left(\frac{k(x-x')\Delta w}{2R}\right), \tag{4}$$

where k is the propagation constant, ρ is the density of the medium, A is the attenuation of the medium, U_0 is the surface velocity amplitude of the subelement, β is the phase of the surface velocity, and R is the distance from the subelement to the point where the pressure is determined [3]. The total acoustic pressure at a specific point is determined by summing the contributions at that point from each of the subelements. This approach was found most efficient for the rectangular element geometry employed.

Phase quantization errors are introduced into the system if digital electronics are used to produce the phase shifts necessary for focusing. Phase quantization errors result in increased sidelobe intensity and decreased main lobe intensity. For example, the focal intensity for a four-bit quantized phase shift (maximum possible phase error of 22.5 degrees) would be reduced by 0.06 dB [4]. The model used in this study includes the effect of this phase quantization error.

Using the model described above, the maximum treatable tumor diameter can be determined when the other parameters including frequency, array length, and G_{TA} are specified. The model provides the field distributions generated by a set of three linear arrays, as shown in Fig. 2. From a plot of intensity versus distance from the array the ratio of the intensity at the focus to that near the surface of the array G_F is determined. This gain is large since such a plot does not consider the time-averaging resulting from scanning the beam in the x direction. This effect is considered by multiplying G_F (expressed as a ratio) by the 3-dB beam width in the x direction. However, G_F must first be adjusted for the following effects: a) the desired G_{TA}, defined previously; b) the effect of increased heating at the surface resulting from each array being used in three different array groups, i.e., three y positions; and c) the increased heating of the tumor resulting from energy outside the 3-dB beam width in the y direction. The last effect is significant for the y direction, but not for the x direction, because the field is unfocused in the y direction, and the intensity decreases much more slowly with distance from the center of the ultrasonic beam. The first two effects will decrease G_F and the third will increase G_F so that an overall corrected gain G' can be defined, based on the field plots from the model, given by

$$G' = G_F - G_{TA} - G_S + G_T, \tag{5}$$

where G_S and G_T represent the increased surface and tumor heating in decibels, respectively. The maximum treatable tumor diameter D is given by

$$D = B \cdot 10^{(0.1\,G')}, \tag{6}$$

where B is the 3-dB beam width in the x direction. This procedure is illustrated by the example provided in the next section.

III. Array Design Illustration

The array parameters will be defined to treat a tumor 7 cm in diameter centered at a depth of 10 cm. The desired G_{TA} will be 3 dB. Additionally, it will be assumed that the center-to-center spacing of the elements in the arrays is equal to one half a wavelength, so that grating lobes are not present. An operating frequency of 250 kHz was chosen to illustrate the design approach. The attenuation coefficient and the velocity will be assumed to have values of 0.15 dB/cm and 1500 m/s, respectively, in the normal and tumor tissue. The array height is determined from (1) to be approximately 1.6 cm.

The results of the application of the simple approach to the determination of the array length are as follows. The time-averaged system gain G and the required array length L, as calculated from (2) and (3), respectively are 4.5 dB and 19.7 cm. This calculation of array length can be checked by applying the more rigorous approach based on field calculations.

The calculated field intensity versus distance from a group of three arrays 20 cm in length and with four-bit phase quantization error is plotted in Fig. 4. For all points close to the array the intensity is at least 17 dB below the intensity at the focus resulting in a G_F of 17 dB. A corrected gain G' of 12 dB is calculated from (5), based on a G_{TA} of 3 dB and a combined effect of G_S and G_T of -2 dB. The 3-dB beam width of the

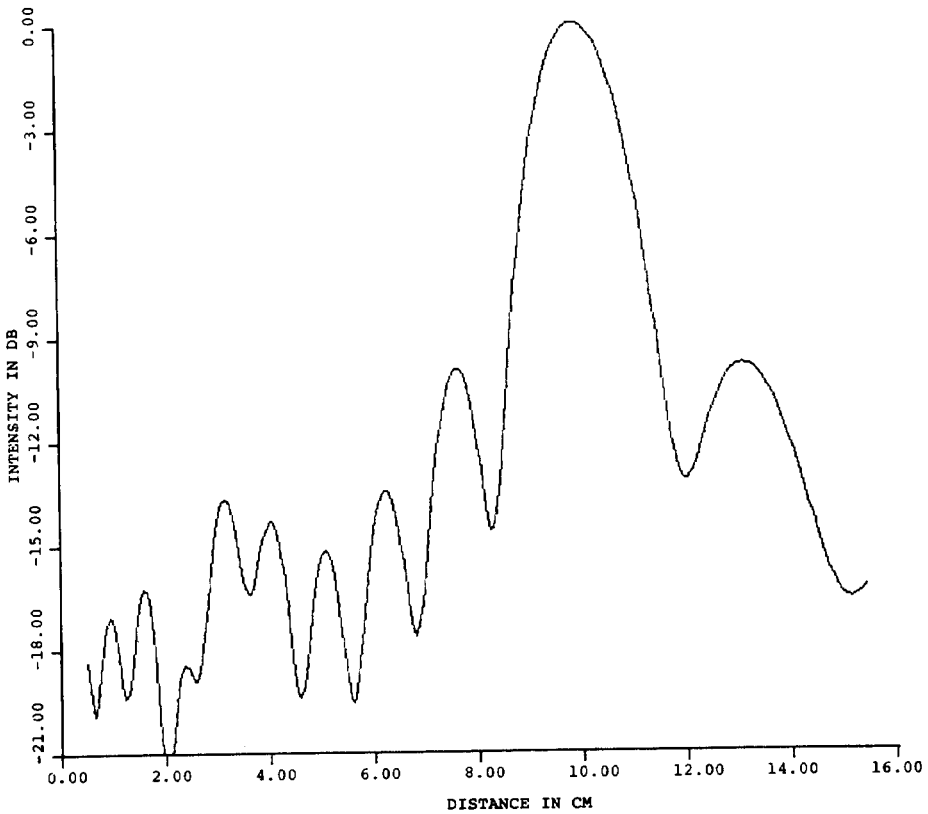

Fig. 4. Relative intensity versus distance from the source in tissue with an attenuation coefficient of 0.15 dB/cm. The source is a group of three linear arrays, each with h = 1.6 cm and L = 20 cm.

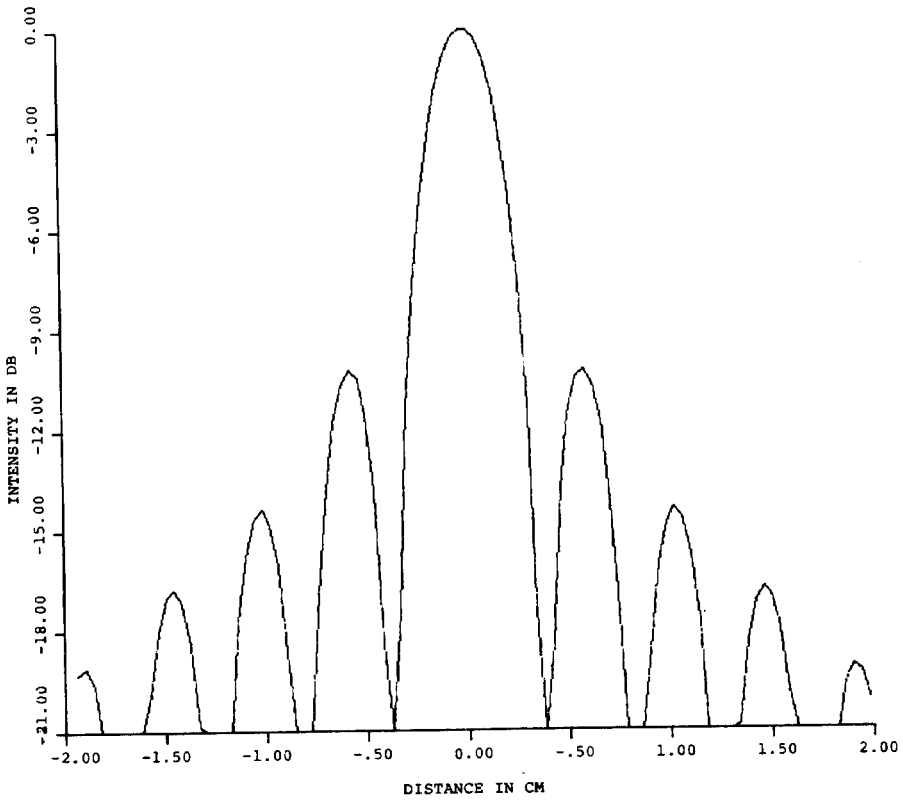

Fig. 5. Relative intensity versus x at z = 10 cm in tissue. Source parameters are the same as for Fig. 4.

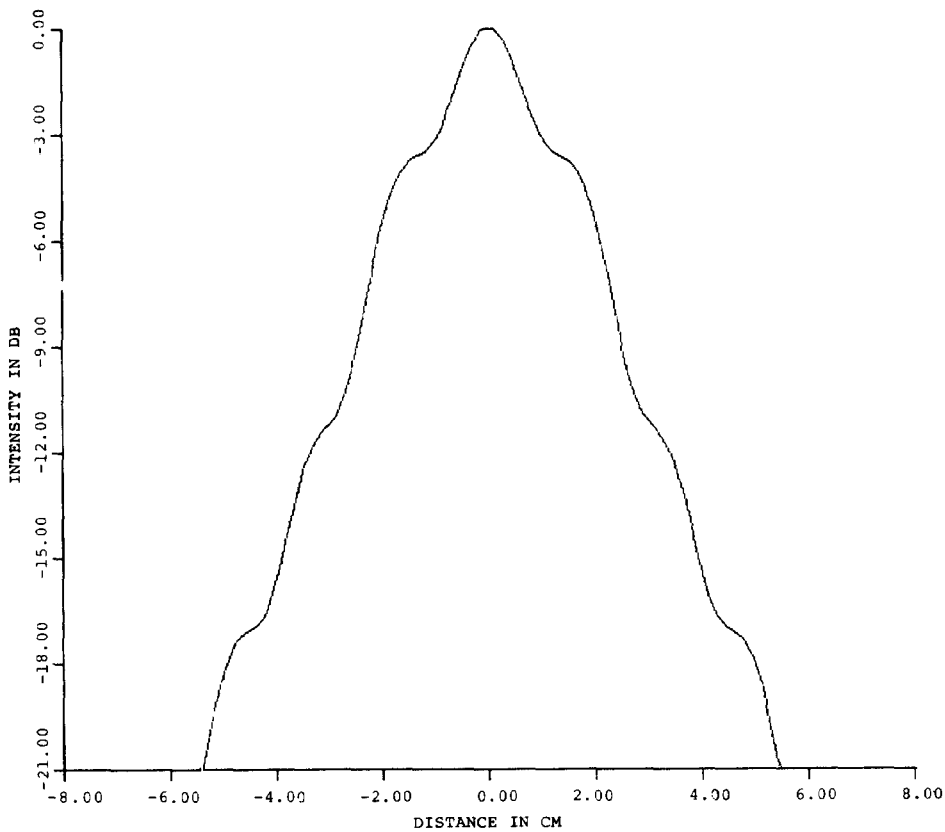

Fig. 6. Relative intensity versus y at $z = 10$ cm in tissue. Source parameters are the same as for Fig. 4.

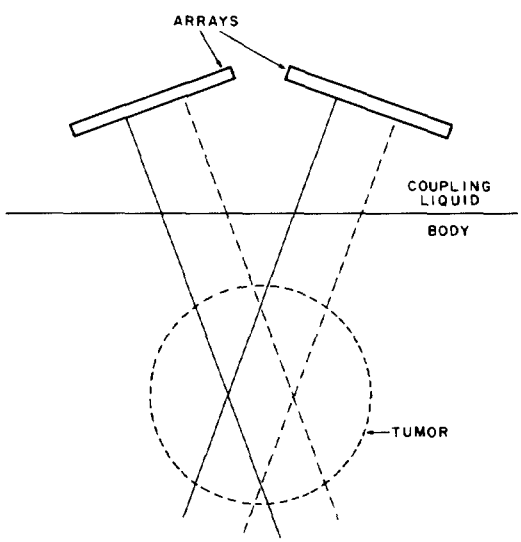

Fig. 7. Superposition of fields in the tumor from two stacks of linear arrays. Excited groups of linear arrays can be changed in unison for each stack to move the central axes of the beams, for example, from solid to dashed lines.

focus is 0.34 cm as measured from the plot of field intensity versus x for this group of arrays (see Fig. 5). The maximum treatable tumor diameter D is 5.4 cm, as calculated from (6), which is 23 percent less than the 7 cm diameter indicated by the simple approach.

Fig. 6 shows the field intensity versus y at a distance of 10 cm from the group of three arrays. The 3-dB beam width from Fig. 6 is approximately 1.8 cm which is very close to the 1.6 cm height of an individual array and confirms this design specification.

IV. Discussion

The stacked linear phased array requires far fewer amplifiers and associated phasing circuits than a two-dimensional array hyperthermia applicator. Since a stacked linear phased array provides lateral gain in only one dimension, there is a trade-off between the size and depth of the tumor to be treated for each specified linear array length. As indicated in the previous section, a stacked linear array operating at 250 kHz with array lengths of 20 cm will provide enough gain to adequately heat a tumor at a depth of 10 cm only if the tumor were less than 5.4 cm in diameter. For a tumor located closer to the surface the maximum lateral dimensions of a treatable tumor are greater.

It may be desirable to treat tumors larger than 5 cm in diameter at depths greater than 10 cm. This appears to be impractical using a single stacked linear array applicator. However, by using two of these applicators with their fields superimposed as illustrated in Fig. 7, one can achieve an additional gain of nearly 3 dB. In this configuration the superposition of fields will be optimum at one depth only. However, this is not an additional restriction as a single stacked array required that the tumor be located at the near-field–far-field transition distance for the y dimension of the array. It should also be noted that the two stacked arrays need not be separately controlled. In Fig. 7, it is readily apparent that the two arrays may be driven in parallel and that the same shifting of linear array excitation can be applied to each. In this approach, no additional amplifiers would be required, but the focal regions would be at different depths since the distances to the tumor are different for corresponding portions of each array. This may actually be a benefit as one array would be focused towards the front of the tumor and one towards the back, providing more uniform heating with depth.

An ultrasonic hyperthermia applicator consisting of a stack of linear phased arrays, based on the work reported here, does appear to provide the electronic scanning of the sound beam

with sufficient system gain while employing a reasonable number of amplifiers and phasing circuits.

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