Analysis of a Multielement Ultrasound Hyperthermia Applicator

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Abstract—**An unfocused multielement ultrasound applicator was developed for hyperthermic treatment of superficial tumors. The applicator contains sixteen 3.8-cm-square individually controllable elements on a 15.2-cm-square piezoelectric ceramic plate. The acoustical power output of each element can be independently applied to facilitate uniform heating throughout the treatment area while minimizing undesired heating in normal tissues. The performance of the applicator was examined by measuring acoustical power output and beam profiles. The results of this analysis indicated that the applicator is capable of producing required therapeutic output levels with excellent localization and control of the power deposition.**

I. Introduction

Although unfocused ultrasound applicators have been used to heat superficial tumors [1]–[3], the beam size of such applicators often does not match the area to be treated. This results in either undesired temperature elevation in the normal tissue surrounding the tumor, or the necessity to physically move the applicator to treat the entire tumor volume. Moreover, such applicators are inflexible in that the spatial distribution of ultrasonic energy within the treatment field cannot be modified to adapt to local changes in blood perfusion that might occur during the course of tumor treatment.

A better approach to treating superficial tumors involves the use of an applicator with individual, independently controllable sections or elements. Superficial tumors can be heated, without excess heating of normal tissues surrounding the area of interest, by controlling the acoustical power output of the individual applicator elements. Multipoint temperature monitoring throughout the treatment field can be used to supply the necessary information to control the spatial extent and magnitude of applicator power, making possible the generation of more uniform temperature distributions across the treatment field.

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An applicator has been developed that comprises a 4×4 array of square elements. Some initial results were previously reported by the authors, which demonstrated that the applicator produced a relatively uniform field and that it could effectively heat a perfused tissue phantom [4]. Since then another group of authors measured and reported additional field measurements for the same applicator [5]. In this paper previously unreported details are provided regarding the original design and testing of the applicator and results of field measurements at 1.0 and 3.4 MHz.

II. Design Considerations

A. Frequency

The design of an ultrasound hyperthermia applicator requires consideration of a number of specific criteria and goals for optimal utility in a specific application. These design criteria include: 1) optimal depth of penetration; 2) size of overall treatment area; 3) flexibility to alter the spatial extent and local power level of the ultrasound beam; and 4) adaptability to many treatment locations. Each of these criteria were considered in the design of the multielement applicator.

The heat generation as a function of depth associated with the passage of plane-wave ultrasound through tissue is dependent on the absorption properties of the tissue that in turn vary approximately linearly with frequency. The intensity of an ultrasonic plane wave propagating in the positive z direction is given by

$$\boldsymbol{I}(z) = I_0 e^{-2Az}\hat{z} \tag{1}$$

where $\boldsymbol{I}(z)$ is the acoustic intensity vector (note that all vectors are in bold type) at a distance z into the tissue, I_0 is the magnitude of the intensity incident on the surface of the tissue (i.e., $z = 0$), $\hat{z}$ is a unit vector in the z direction and A is the attenuation coefficient of the tissue. The rate that energy is dissipated as the wave travels through the tissue is equal to the absorbed power per unit volume P_A that may be derived from the general expression

$$P_A = -\nabla \boldsymbol{I} \tag{2}$$

yielding

$$P_A = 2AI_0 e^{-2Az} \tag{3}$$

for the case where the absorption and attenuation coefficients are equal.

Assuming that the frequency dependence of the attenuation can be expressed approximately as

$$A = A_0 f \tag{4}$$

where A_0 is the attenuation coefficient in tissue at 1 MHz and f is the frequency in MHz [6], (3) can be written as

$$\frac{P_A}{I_0} = 2A_0 f e^{-2A_0 f z}. \tag{5}$$

Fig. 1 is a plot of P_A/I_0 as a function of frequency at three depths; 2, 5, and 10 cm ($A_0 = 0.1 \text{ cm}^{-1}$). The plot shows that an optimal frequency can be chosen for maximum heat generation at specific tissue depths. The optimal frequency is given by

$$f = \frac{1}{2Az} \tag{6}$$

which was determined by differentiating (5) with respect to frequency and setting the result equal to zero. It is evident from these data that the optimal frequency for heating by plane waves in the 2- to 5-cm range is between approximately 1 and 2 MHz.

The above analysis does not address the temperature rise that would be expected as a function of depth. Such a computation depends upon blood perfusion and boundary conditions at the surface. The blood perfusion will vary among clinical situations and the surface temperature can be adjusted by applying surface cooling. Thus, for the initial design only the above analysis was used to provide the approximate frequency for a desired treatment depth.

B. Element Dimensions

The overall treatment area is determined largely by the size of the applicator aperture. This is due in large part to the directive nature of ultrasound beams from sources with apertures of dimensions greater than a few millimeters at low megahertz frequencies in tissue. The directivity of ultrasound under these conditions permits specification of the treatment region, and offers the possibility of adapting the shape of the ultrasound beam to the shape of the area to be heated. Further, by using multiple independent elements the power level applied to a specific region could be adjusted such that local changes in blood flow, and thus in required power deposition, could be countered with local modifications in ultrasound intensity.

In order to utilize the aforementioned features of ultrasound transduction and propagation in tissue, square applicators of two sizes were constructed, one consisting of a square array of 16 elements and the other a square array of four elements. For each applicator the piezoelectric ceramic elements (PZT-8) were 3.8 cm × 3.8 cm with a thickness of 0.2 cm. Thus, similar control of the treatment region is provided for both the x and y dimensions, where the z direction is perpendicular to the surfaces of the elements. Besides the obvious advantages of geometry, square sources produce fields that have no on-axis nulls and are more uniform in the near field than fields generated by circular sources [7], [8].

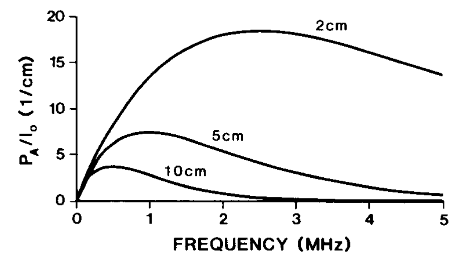

Fig. 1. Calculated relative power deposition rate in tissue as function of frequency.

The width of each element was chosen such that the field of each element would be well collimated throughout the working depth of the applicator. Considering a patient coupling bolus of degassed water approximately 5 cm in depth affixed to the face of the applicator, and a maximum penetration depth of 6 cm at 1 MHz, the minimum length of the near field must be 11 cm. The minimum width of the elements may then be approximately expressed in terms of the near-field length L_{nf} as

$$d_{min} \simeq (4\lambda L_{nf})^{1/2} \tag{7}$$

where λ is the acoustic wavelength in the tissue. Using (7), the minimum element width for this case is approximately 2.6 cm. Larger elements were chosen to provide a larger treatment area without increasing the number of elements and to insure that the near field extended well beyond the working region.

III. Fabrication

Three 1-MHz 16-element applicators, each employing a different element mounting design, were fabricated. The first design, ME1, consisted of four adjacently mounted piezoelectric ceramic 7.6-cm-square plates. Four 3.8-cm-square elements were defined on each of the plates. The four plates were mounted in a 17 cm × 17 cm × 12 cm water-tight anodized aluminum housing on a Plexiglas rear support frame that maintained the air backing for the 16 transducer elements (see Fig. 2). The plates were seated in the frame such that a 3.2-mm-wide strip of Plexiglas separated the four plates. An anodized aluminum front face frame was used to secure and provide a common ground contact for the ceramic plates.

A second design, ME2, replaced the Plexiglas rear support frame with a brass frame that reduced the spacing between the four plates to 0.79 mm. The front face frame was eliminated in this design by securing the plates in the brass frame with conducting epoxy around the front edges of the plates. The plates and supporting frame were mounted in the same housing that was employed for ME1.

The third design, ME3, used only a single 15.2-cm-square plate, as shown in Fig. 3, instead of four separate plates. This design required that the plate be supported only around its edges, thus eliminating the need for the crossbars in the rear support frame. Conducting epoxy was

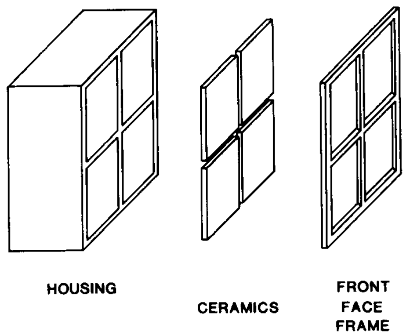

Fig. 2. Exploded view of ME1.

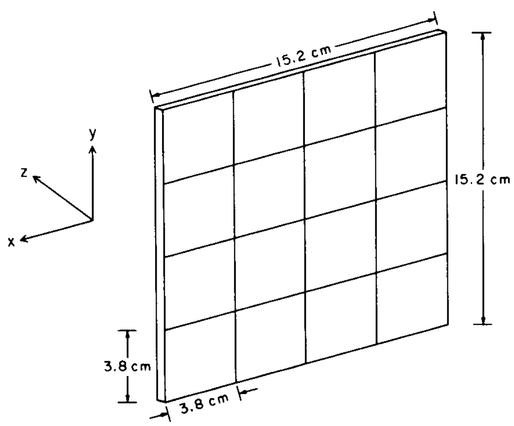

Fig. 3. Element configuration on back surface of piezoelectric ceramic of ME3.

again used both to secure the plate in the housing and to provide the ground connection. Later on, a water coupling bolus was fabricated and attached to this applicator so that it could be used for animal testing. The bolus consisted of an aluminum frame holding a 0.5-mm latex rubber membrane, approximately 6.4 cm from the transducer face. The membrane was flexible and extended beyond the bolus frame such that it could conform to the body contour.

The elements of each applicator were designed to be individually excited using sixteen independent RF sources (four in the small applicator). Each applicator element was matched to its 50-Ω radio frequency (RF) source using a high-Q electrical impedance matching network mounted inside the transducer housing.

IV. Methods

The acoustical field distributions produced by the applicator were measured using a 1-mm diameter PZT-5A hydrophone probe. The applicator was placed in a 50-l Plexiglas tank filled with degassed water and lined on the inside walls with sound absorbing material. Microcomputer controlled stepping motors swept the probe across the field of the applicator. The acoustic pressure detected by the probe was plotted as both a function of lateral (across the face of the applicator) and longitudinal (perpendicular to the face of the applicator) position.

The theoretical field distributions were calculated using the rectangular radiator method [8], [9]. This method divides the element of the applicator into subelements that are small enough so that their field can be represented, in the region of interest, by the far field approximation. The total acoustic pressure at a given field point is calculated by summing the contribution at that point from each of the subelements.

The acoustical power output of the applicator was determined by measuring the radiation force exerted on a reflecting target. The target was 7.6 cm × 5.1 cm × 0.3 cm and consisted of approximately 200-μm-thick brass plates bonded to each side of an open rectangular Plexiglas frame. The air gap formed between the plates made the target virtually a perfect reflector. The target was suspended at an angle of 45 degrees to the incident field by small diameter nylon threads and enclosed by a Plexiglas frame with acoustically transparent windows (25-μm-thick polyethylene) in the front and rear to prevent convection and acoustic streaming from affecting the response of the target.

The total acoustic power incident on the target, W, was determined from

$$W = \frac{mgcd}{(L^2 - d^2)^{1/2}} \tag{12}$$

where m is the mass of the target, corrected for buoyancy, g is the acceleration of gravity, L is the length of the suspension, c is the speed of sound in water, and d is the horizontal deflection of the target that occurs when the sound is present [10]. The deflection, d, was measured with a cathetometer to an accuracy of 0.2 mm.

V. Results

A. Field Intensity Distributions

Field intensity distributions were determined at 1.0 MHz for the three 16-element transducers designed as a part of this study, and at 1.0 and 3.4 MHz for the final design with coupling bolus attached. The transverse field distribution plot for ME1 showed significant variations in intensity at locations along one of the bars of the face frame securing the four ceramic plates. The minima did not appear to correlate with the position of the edges of the elements, perhaps because the scan was along a bar. However, no large minima were observed in the transverse field when scanned across the boundary between any two adjacent elements on the same plate. Initially the nonuniformity occurring between plates was thought to be due primarily to the 6.4-mm-wide aluminum frame that covered the joints between plates on the front face of the transducer. However, transverse field intensity distributions for ME2, which had no face plate, were measured and the results, (as illustrated in a typical field plot in Fig. 4), showed that this design also exhibited an unacceptable nonuniformity, primarily a large minimum at the joint between the two plates. This suggests that the decreased intensity at the joints was due in part to the damping effects

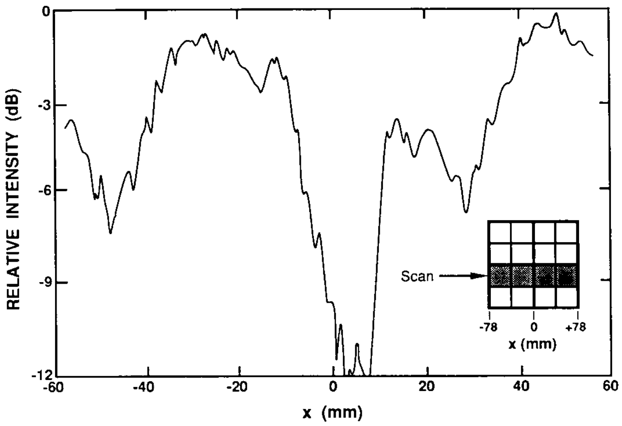

Fig. 4. Relative field intensity of ME2 versus x position ($z = 7.6$ cm). Data points were taken at such small interval that curve is presented as continuum.

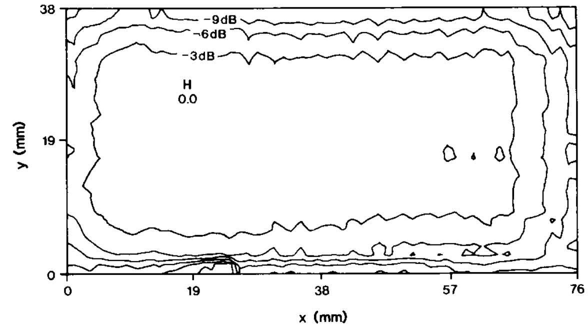

Fig. 5. Contour plot of the relative field intensity of ME3, with two adjacent elements excited, versus x and y ($z = 7.6$ cm).

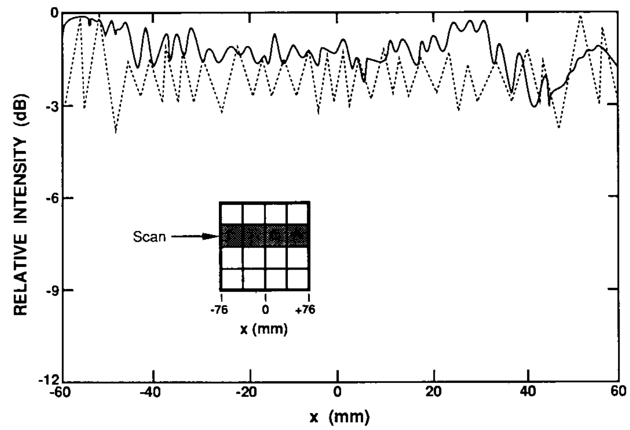

Fig. 6. Comparison of theoretical (dotted line) and experimental (solid line) relative field intensity of ME3, with four linearly adjacent elements excited, versus x position ($z = 7.6$ cm). Experimental data points were taken at such small interval that experimental curve is presented as continuum.

of the underlying supporting frame and the epoxy used to hold the plates in place.

Indeed, it was found that an applicator made from a single plate produced the desired uniform field. Transverse field intensity distributions for ME3, as shown in the examples of Figs. 5 and 6 and in [5], exhibit no large minima in the field between adjacent elements. The beams appear to be well collimated, and when adjacent elements are excited, a smooth, notchless transition between beams is present with a typical lateral variation in acoustic pressure of less than 2 dB. These results are consistent with the results from field profiles between adjacent elements on a single plate for ME1 and ME2.

Fig. 6 shows a comparison of a typical experimental and the theoretical transverse field intensity distributions, with four linearly adjacent elements excited, at a distance of 7.6 cm from the face of the transducer. This distance corresponds to the approximate distance from the transducer to a superficial tumor with a degassed water coupling bolus in place. The theoretical plot shows variations in the field intensity similar to those observed experimentally; however, the experimental data consistently showed more uniform intensity profiles than predicted by theory. This is expected, since the theory does not include the effects of the damping of the elements that occurs near their edges. The damping results from the clamping of the element's edges either by the epoxy holding the ceramic plate in the housing or by an unexcited adjacent element. This decreases the acoustical output of the elements near their edges, which tends to smooth out the field intensity variations.

Fig. 7 shows transverse field intensity distributions, with the coupling bolus in place on ME3 and four linearly adjacent elements excited as for Fig. 6, at several distances from the source at both 1.0 and 3.4 MHz. These results demonstrate the relative uniformity of the fields at the fundamental and at the third harmonic. It is desirable to use higher frequencies for shallow tumors or for cases where bone lies closely behind the tumor. This applicator can be used at 1.0 and 3.4 MHz and potentially at higher odd harmonics of the fundamental when still higher frequencies are desirable.

Fig. 8 shows the relative intensity profile, in the longitudinal direction, with one element excited first at 1.0 and then at 3.4 MHz. The increasing intensity with depth at 1.0 MHz and the relatively uniform intensity profile at 3.4 MHz are consistent with the theory for a square source [7].

B. Acoustical Power Output

The acoustical power output of several of the elements of ME3 operating at 1.0 MHz was measured using the reflecting radiation force target. Fig. 9 shows a typical result for a measure of acoustical power output versus the input electrical power for one of these elements. The maximum total acoustic power output per element with the RF power amplifiers (Labthermics Technologies, Inc.) used in this study is approximately 150 W, leading to a maximum acoustic intensity per element of approximately 10 W/cm^2. The average electroacoustic efficiency of the elements was determined to be approximately 89 percent,

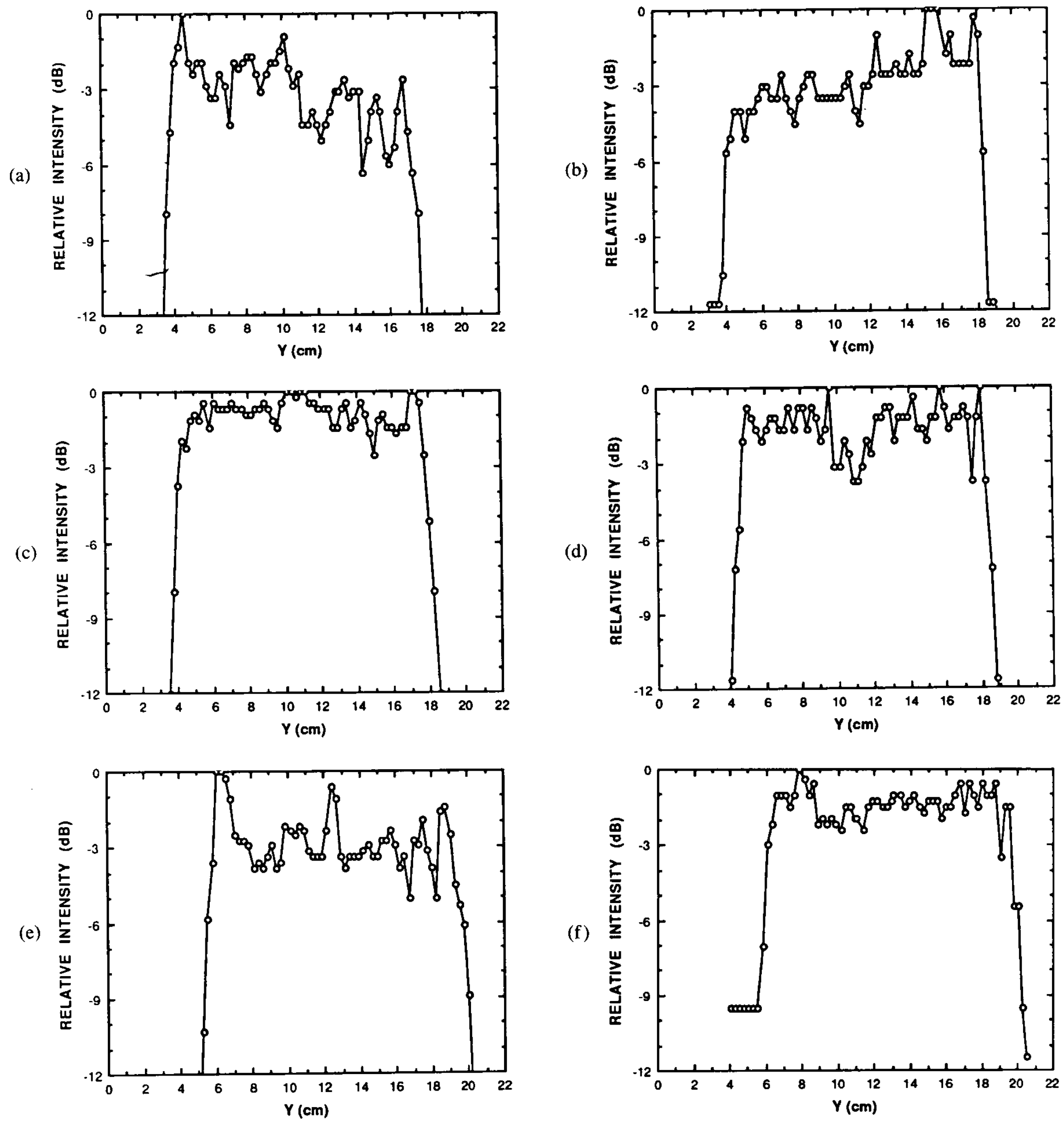

Fig. 7. Relative intensity of ME3, with coupling bolus mounted and four linearly adjacent elements excited (as in Fig. 6), versus y position at 1.0 and 3.4 MHz as function of z position. (a) f = 1.0 MHz and z = 8.2 cm. (b) f = 3.4 MHz and z = 8.2 cm. (c) f = 1.0 MHz and z = 10.2 cm. (d) f = 3.4 MHz and z = 10.2 cm. (e) f = 1.0 MHz and z = 12.2 cm. (f) f = 3.4 MHz and z = 12.2 cm.

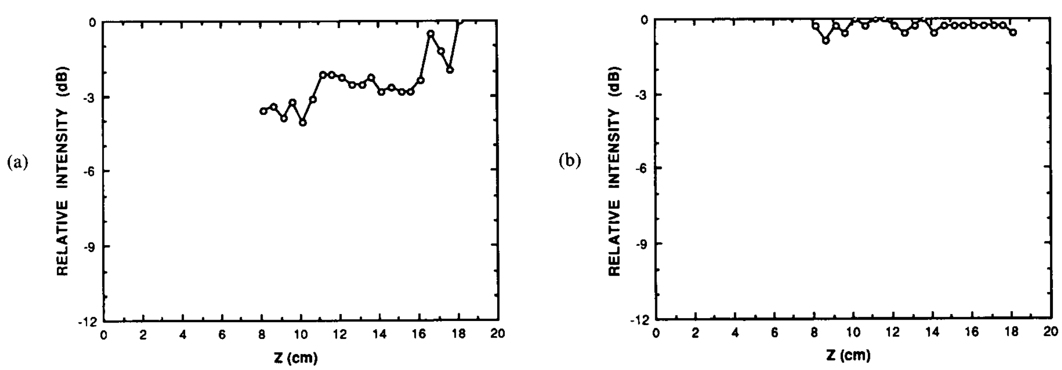

Fig. 8. Relative intensity of ME3, with coupling bolus mounted, versus z position at (a) 1.0 MHz and (b) 3.4 MHz with one element excited.

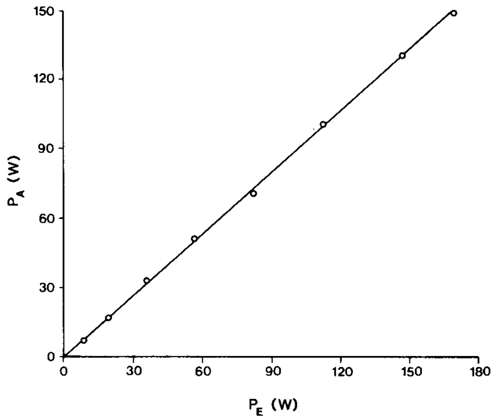

Fig. 9. Acoustical power output, P_A, versus electrical power input, P_E, for element of ME3.

and thus approximately 11 percent of the input power is lost to heat in the ceramic.

VI. Conclusion

A multielement ultrasonic hyperthermia applicator for the treatment of superficial tumors was designed and constructed. A simple analysis of the relative power deposition rate in tissue as a function of frequency indicated that an applicator frequency between approximately 1 and 2 MHz would be optimal for treating tumors in the 2- to 5-cm range of depths. The applicator was also evaluated at its third harmonic frequency that will be useful for treating shallower tumors where lesser penetration is desirable to avoid heating underlying normal tissues.

The use of a single piezoelectric ceramic plate not only simplified the design of the applicator, but it also eliminated the large transverse variations in the acoustic intensity that occurred with applicators using multiple plates. In the longitudinal direction, the relative intensity for each element increased with depth at 1.0 MHz and was very uniform at 3.4 MHz. Of course, the size of the applicator is governed not only by acoustical considerations, but also by the intended clinical application. For example, smaller applicators, such as a 2 $\times$ 2 element applicator, can be fabricated to provide better access to restricted regions such as the neck.

Acoustical power measurements demonstrated that the applicator was capable of producing intensities of at least 9 W/cm^2. Reports in the literature indicate that no more than approximately 2 W/cm^2 should be required for treatment of superficial tumors [1], [3]. Thus, it appears that this applicator supplies more than sufficient power to heat tumors. Since each element can be controlled independently, the area to be heated can be altered to match the size of the tumor, and the power to each element can be varied to maintain a uniform temperature distribution in the presence of inhomogeneous or variably perfused tissue. This system is a significant improvement over previous generations of ultrasonic hyperthermia applicators.

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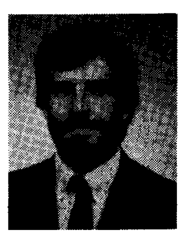

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