HEATING OF A PERFUSED TISSUE PHANTOM USING A MULTIELEMENT ULTRASONIC HYPERTHERMIA APPLICATOR

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ABSTRACT

An ultrasonic multielement hyperthermia applicator capable of treating surface tumors of various dimensions was designed and constructed. The applicator contained sixteen 1.5 inch square elements on a large ceramic transducer, whose acoustical power outputs can be independently controlled to minimize undesired heating in normal tissues. The performance of the applicator was examined by heating a perfused pig kidney "phantom". The kidney was heated with the applicator at several different perfusion rates, controlled by a roller pump, and the temperatures within the kidney were monitored. The results of these experiments show that the applicator is capable of producing significant temperature rises in this phantom and that the perfused kidney is a good model to study the effects of perfusion rates on ultrasonically induced hyperthermia.

I. INTRODUCTION

Ultrasound has two properties that make its use in inducing local hyperthermia in superficial tumors desireable: 1) since the wavelength is small in the therapeutic frequency range, ultrasound beams can be well-collimated, enabling localization of energy deposition; and 2) since the absorption in tissue is proportional to frequency, the depth of penetration can be controlled by changing the frequency of the ultrasound [1].

The disadvantages of using ultrasound in hyperthermia are primarily related to impedance mismatches that are present at tissue-air or tissue-bone interfaces, and the high absorption of ultrasound in bone. Ultrasound is almost totally reflected at tissue-air interfaces, and can cause local hot spots at tissue-bone interfaces [2].

A major problem with current ultrasound hyperthermia systems is the inability to accurately control the deposition of the energy used to heat tumors in the body. Single unfocused ultrasonic transducers have been used to heat superficial tumors, but often these applicators are too small to heat the entire tumor volume without moving the applicator. A single transducer applicator larger than the surface area of the tumor solves the problem of heating the entire tumor volume simultaneously, but produces excess heating in normal tissues surrounding the tumor.

A better approach to heating superficial tumors would be to use an applicator that consists of many unfocused transducer elements whose acoustical power outputs can be independently controlled. The extent of this applicator's treatment field should be greater than the equivalent surface area of most surface tumors so that movement of the applicator during treatment would not be required to heat the entire tumor volume. Superficial tumors can be heated, without excess heating of normal tissues, by controlling the acoustical power output of the individual elements of the applicator. The temperature in normal and tumor tissues would be monitored with thermocouples. Should the temperature of a particular region in the treatment field become excessive, the acoustic power of the element(s) supplying energy to that region would be reduced. This would make it possible maintain the normal and tumor tissue temperatures in the treatment field at their desired levels.

This paper studies the design considerations and testing of such a multielement applicator. A recently developed perfused tissue phantom was used to test the applicator's ability to heat perfused tissue. Using this phantom, the performance of the applicator was evaluated with several perfusion rates. The field intensity of the applicator in a transverse plane and temperature data associated with heating of the phantom are presented.

II. DESIGN CONSIDERATIONS

One of the most important design considerations was the choice of the operating frequency of the applicator. Since the attenuation coefficient of ultrasound increases approximately linearly with frequency, there is a trade off between depth of penetration of the ultrasound and rate of heat generation. The intensity of an ultrasonic plane wave propagating in the positive z direction is given by:

$$\vec{I}(z) = I_o e^{-2Az}\ \hat{z} \qquad (1)$$

where $\vec{I}(z)$ is the intensity at a distance z into the tissue, Io is the intensity incident on the surface of the tissue, $\hat{z}$ is a unit vector in the positive 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 power loss P_L which may be derived from the general expression

$$P_L = -\nabla\cdot\vec{I} \qquad (2)$$

which yields

$$P_L = 2AI_o e^{-2Az} \qquad (3)$$

for the case under consideration. For the purposes of this paper the absorption and attenuation coefficients are considered equivalent. From Equation (3), it is clear that the attenuation coefficient A affects the power deposition in two ways. A larger value for A will increase the power deposition, but will decrease the depth of penetration. Figure 1 shows typical attenuation curves for muscle at several frequencies [3]. It is clear that as frequency increases, the depth of penetration decreases since the attenuation is approximately linearly dependent on frequency. Figure 2 illustrates the power deposition curves in muscle at the same frequencies as shown in Fig. 1. Clearly the choice of an optimum frequency depends on the desired depth of the treatment field. An operating frequency of 1 MHz was chosen for this applicator to provide adequate power deposition to tumors 2 to 5 cm in depth [4]. The attenuation coefficient at this frequency is large enough to produce adequate heating with minimal incident power while minimizing heating in normal tissue behind the tumor.

The multielement applicator design consists of a square array of 16 elements mounted in an aluminum housing. The housing contains 16 matching networks for the elements. Since the elements were large (3.8 cm square), the driving electric field was expected to be confined to the area of the elements [5]. For use in clinical hyperthermia treatment, a bolus filled with circulating degassed water would be used to couple the ultrasonic energy to the patient and to provide some cooling to the surface of the patient.

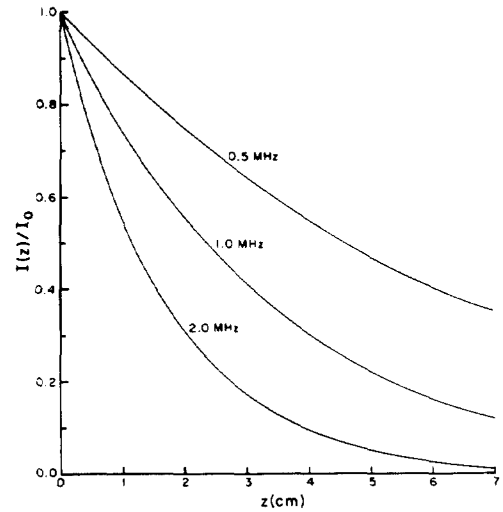

Fig. 1 Attenuation of ultrasound in muscle

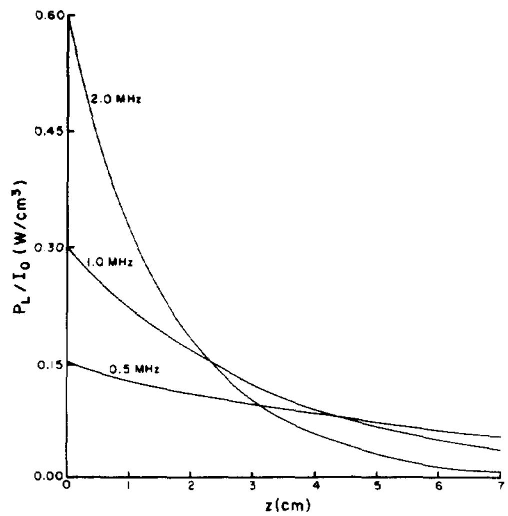

Fig. 2 Ultrasound power deposition curves in muscle

III. METHODS AND RESULTS

To measure the field intensity profiles, the applicator was placed in a tank filled with degassed water. A computer controlled millbase allowed a hydrophone probe (a 1 mm diameter, 20 MHz PZT-5A ceramic disk mounted on the tip of a semi-rigid coaxial rod) to be swept across the field of the applicator in the desired manner. The output of the probe was amplified, then digitized and stored in a minicomputer. The data was plotted as relative intensity versus position.

Figure 3 is a surface plot of the field intensity, with two adjacent elements excited, in a plane parallel to and 7.6 cm from the face of the applicator. This profile shows that the field intensity is quite uniform over a large surface area. This was of interest since these two elements served as the source for the phantom studies and the field plotted in Fig. 3 was for a plane at the surface of the phantom.

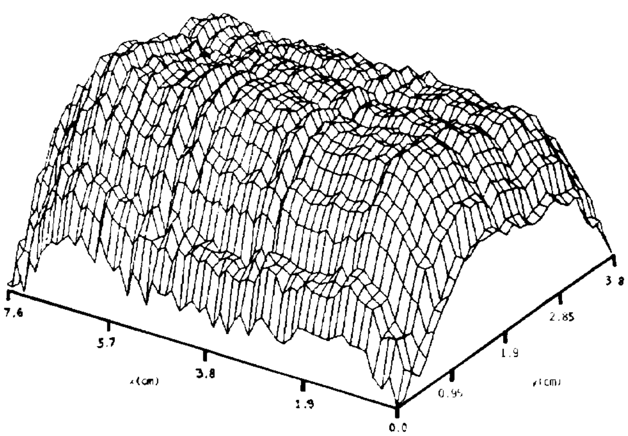

Fig. 3 Surface plot of field intensity of two adjacent elements (z = 7.6cm)

The perfused tissue phantom employed in this study was a pig kidney, which was prepared by ethyl alcohol (ROH) fixation after excision, stored in alcohol, and rehydrated prior to use [6]. The fixed kidney has patent vascular channels whose architecture possesses all of the subtle complexities and sizes which might be observed in living tissues. The intrinsic thermal conductivity (k) of the ROH fixed kidney cortex increased approximately 10%, and the k of the medulla was essentially unchanged when compared to the respective values found in the freshly excised organ [7]. The absorption coefficient at 1 MHz in the rehydrated fixed kidney was measured as 0.014 Np/cm which is approximately half that for freshly excised kidney [3].

The phantom was perfused with degassed water at a rate controlled by a roller pump which could vary the flow and/or pressure to simulate various tissue perfusion conditions. The kidney weighed approximately 44 grams. Because of its relatively small dimensions (7 cm x 3.5 cm x 2.5 cm) the kidney was irradiated with only two of the applicator's elements, in a tank of degassed water. Nine thermocouples were placed at various depths (d) and lateral positions throughout the kidney and the temperature data was collected using a 16 channel temperature data aquisition system consisting of a 16 channel thermometry system (TX-100, URI Therm-X, Inc.) and a microcomputer (Apple IIe) for data storage.

The affect of perfusion rate on the temperature versus time response of the kidney is shown in Fig. 4. As expected, an increase in perfusion results in a decrease in both the rate of temperature increase and the steady state temperature within the phantom. Figure 5 is a comparison of the temperature versus time response at three different depths. The temperature at 3 mm depth was consistantly lower than that at greater depths. This is likely due to the combination of two factors: 1) greater heat loss by conduction to the nearby surface; and 2) greater perfusion in the kidney cortex relative to the medulla. The temperature at the other two depths is consistent with the decrease in heating rate with depth.

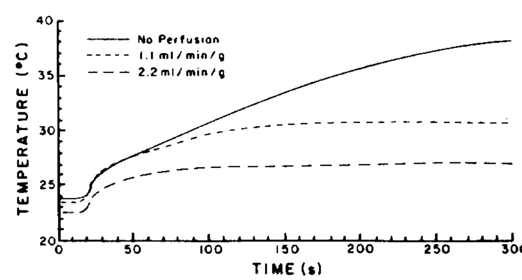

Fig. 4 Plot of temperature versus time for several perfusion rates (d=11mm, $I_0 \doteq 5$ W/cm^2)

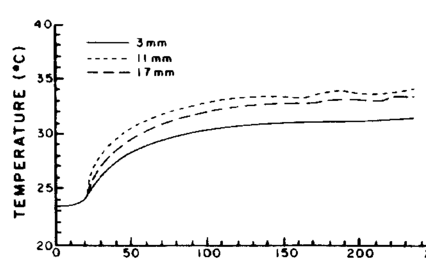

Fig. 5 Plot of temperature versus time for three different depths (1.1ml/min/g, $I_0 \doteq 10$W/cm^2)

IV. CONCLUSIONS

A multielement ultrasonic hyperthermia applicator for the treatment of superficial tumors was designed and constructed. Initial field measurements suggest that the applicator will provide relatively uniform intensity over a transverse plane. Initial results using a perfused kidney phantom show that the applicator can produce adequate heating with perfusion levels comparable to those that could be found in tumors.

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