DESIGN AND CHARACTERIZATION OF A 10 cm ANNULAR ARRAY TRANSDUCER FOR HIGH INTENSITY FOCUSED ULTRASOUND (HIFU) APPLICATIONS

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ABSTRACT

A large transducer to generate high intensity focused ultrasound (HIFU) for surgery via tissue ablation is described. Designed to address the need for therapeutic applications into deep tissue, the device operates at 1.25 MHz. In order to increase focus-to-entrance intensity ratios, the design was optimized to have a natural focal distance of 10 cm at f/1.0. Variable focus is achieved by phasing the array, and particular care was exercised to maximize the range of focal distances that can be achieved with a limited number of amplifiers and elements, using a constant-phase delay design. Design approach and tradeoffs are discussed, with emphasis in comparing theoretical calculations with test performance.

INTRODUCTION

The use of high intensity focused ultrasound (HIFU) has been growing in the last few years, and there is abundant literature on the method itself[1-4]. One of the limitations to delivering high doses deep into tissue lies in the fact that attenuation partly cancels the effect of preferential dosage at the focus relative to the region where the beam enters the tissue.

Another limitation is that, to treat a volume, the transducer must be moved along a plane parallel to the skin, and then towards and away from the body in order to cover all three dimensions. But varying the distance between the transducer and the body has a drawback: the dosage at the entrance surface increases as the transducer is retracted (and also does the dose at the focus, although at a different rate, depending on tissue attenuation).

With the exception of pioneering work done in treating brain tumors[1] with large transducers, the method has been developed mainly to generate lesions in the 2-4 cm range of depths. Extending the HIFU technique to treat tissue deep into the body would bring the benefits of this "bloodless scalpel" to new therapeutic possibilities, such as breast, liver, and other large organs.

Using the annular phased array concept previously used both in imaging[5] and therapy[4], the present work explores the possibility of a design to treat at depths between 6 and 10 cm. The design goals and approach used are discussed, as well as a description of the computational tools used in the design.

METHODS

The ideal design, when attempting to sweep a focal zone in three dimensions, is a two-dimensional array. However, the complexities of 2-D transducer design, amplifiers and interconnections is beyond the scope of this work. Since sweeping in two dimensions parallel to the entrance plane is a relatively simple motion control task, we have addressed the axial sweep issue with a variable focus annular array design. The array was chosen with a natural focus (radius of curvature) of 10 cm, and an aperture f/1.0 (see figure 1). This geometry would allow about 13 mm for standoff, and maintain a large focus-to-skin dose ratio[3].

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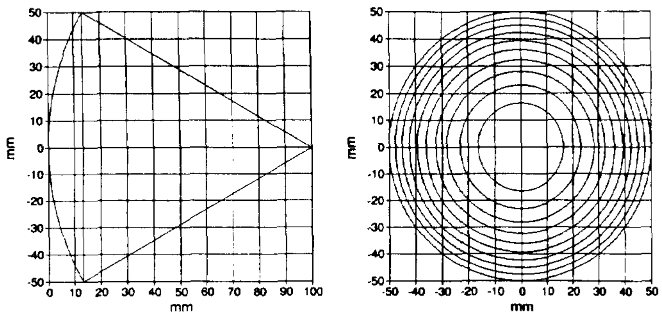

Figure 1. Design of annular array.

The optimal frequency was determined using the method outlined in reference [3], which for an attenuation coefficient of 0.7 dB/MHz-cm and 10 cm in tissue is 1.25 MHz.

As discussed in [4], the number of elements is dependent on wavelength and geometry. Two basic parameters determine the number of elements: the maximum intended phase difference between elements, and the intended focal range. It is easy to prove that the smallest number of elements required for a given focal range occurs when their weighted mean radii correspond to equal phase shifts (see figure 2).

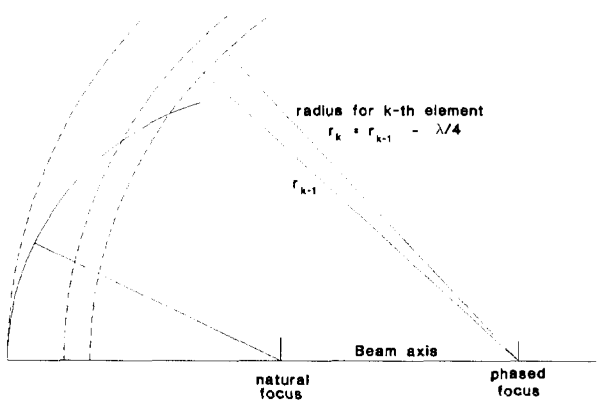

Figure 2. Geometry for determining annuli dimensions.

This differs from the standard approach of making all elements equal in area, which results in equal load for the amplifiers but not necessarily fewest elements. The equal phase design, as it turns out for this geometry, results in annuli that have areas that decrease only slightly with increasing diameter - almost an equi-area design. However, this is not necessarily the case for larger apertures or larger focusing ranges.

We chose to divide the device into sub-elements that would differ at most by about 90° (radial spacing of $\lambda/4$) from their neighbors, when driven to focus the sound at its extreme far end of the range (about 128mm), resulting in a minimum number of 10 elements.

SIMULATION

In order to simulate the beam in water, the surface of the device was divided into cells 1 mm per side (see figure 3), and a straight-forward Rayleigh-Sommerfeld integration was performed numerically for sets of points either at a plane orthogonal to the beam axis (figures 4 and 5) or along the axis (figure 6). Because the beams would be ultimately compared with schlieren images, we included the pressure projections in the simulation, which is also shown in figure 5. Although the intensity maps in the plane perpendicular to the beam are similar for various focusing conditions, the beam profiles along the axis change dramatically as the phases are varied (see figures 6, 7 and 8). The secondary peak visible in figure 8, when the focus is shifted away from the transducer, increases significantly in amplitude as the focus is moved beyond 120mm.

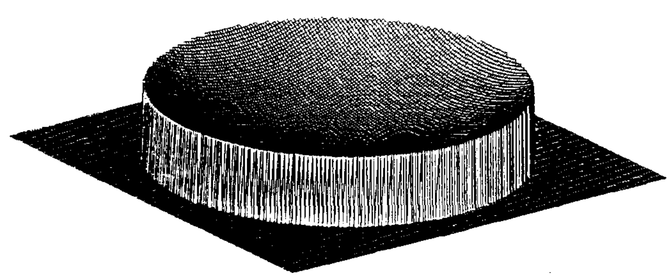

Figure 3. Transducer surface divided into mesh used for beam simulation.

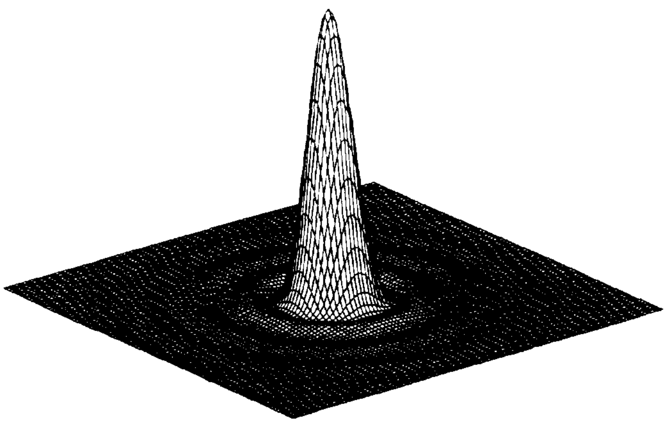

Figure 4. Computed intensity at the focal plane. Area shown is 10mm by 10mm.

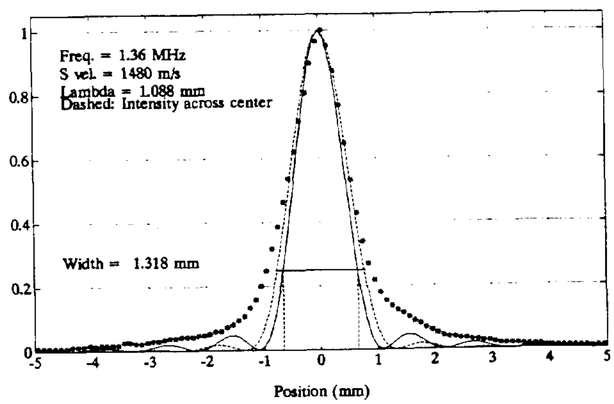

*Figure 5. Computed beam profile across the beam axis, at the focal plane. Measured pressure projection (schlieren) from measurements shown as *.*

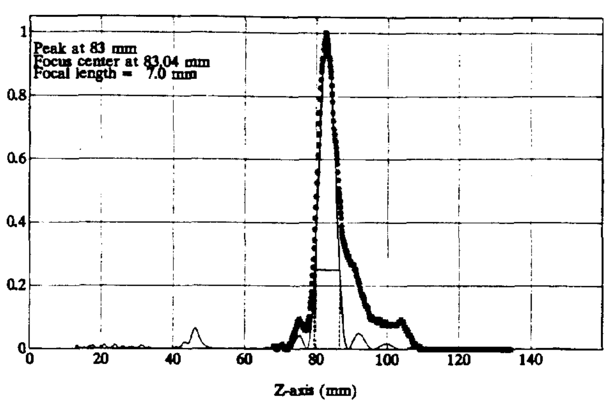

Figure 6. Computed and measured axial beam profile for 80mm focus.

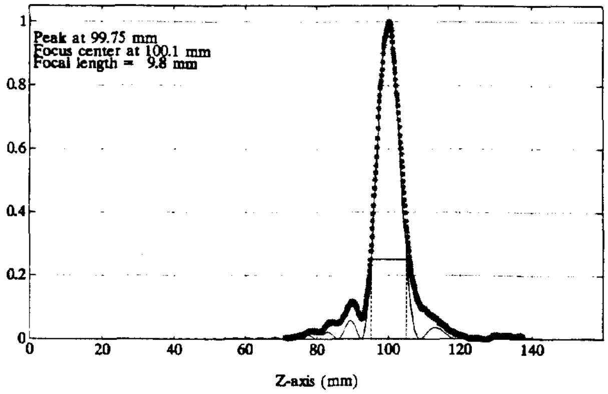

Figure 7. Computed and measured beam profile along the axis, all elements in phase.

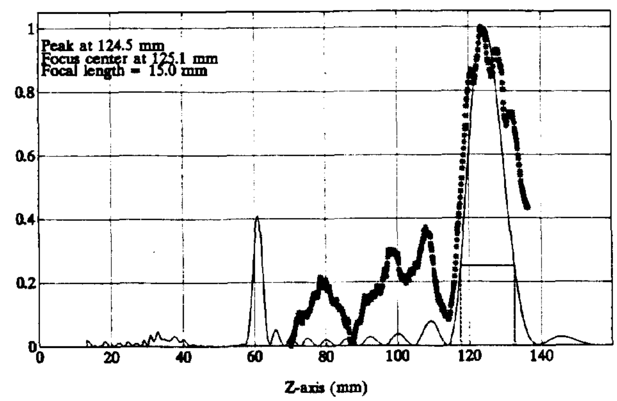

Figure 8. Computed and measured axial profile for 90° shifts among consecutive elements. Note secondary focus at 60 mm in simulation.

PERFORMANCE

The first device built differs somewhat from the design. The device exhibits a large acoustical coupling among neighboring elements, most certainly due to the monolithic construction, in which the elements are defined by removing a ring of plated electrode. This large cross-talk undoubtedly has some effect on the focusing ability, as verified by measurements. Figures 9, 10 and 11 show the beam when focused at the nearest end of the range, at its natural focus, and at the farthest end of the range.

The individual elements were connected to amplifiers driven by generators in which the phase and magnitude can be adjusted independently.

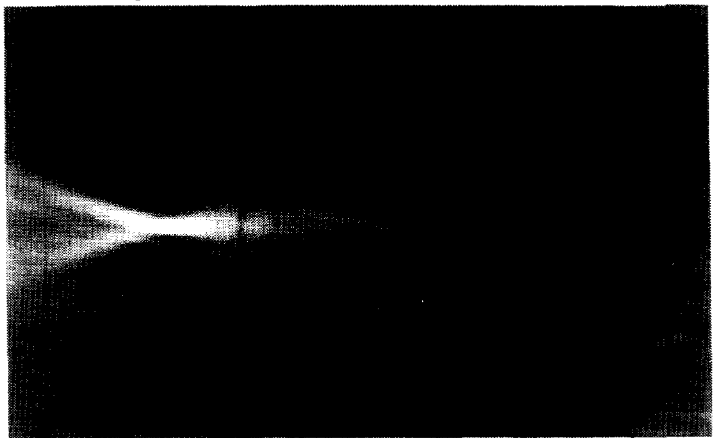

Figure 9. Beam focused closest to the transducer (82mm). Transducer is to the left of the schlieren image.

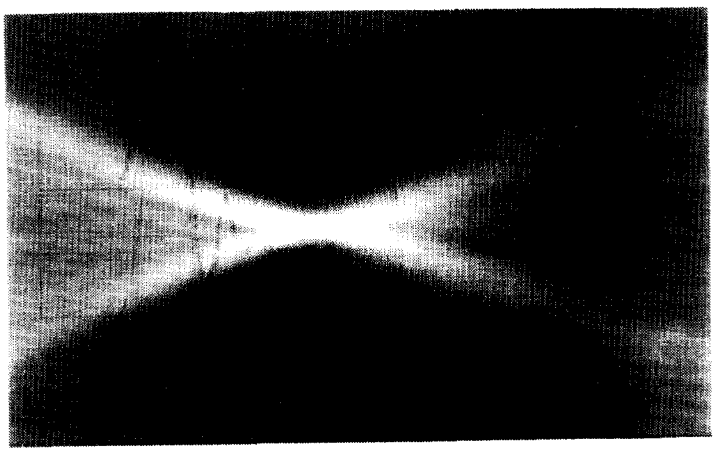

Figure 10. Beam created with all elements in phase, at 100mm.

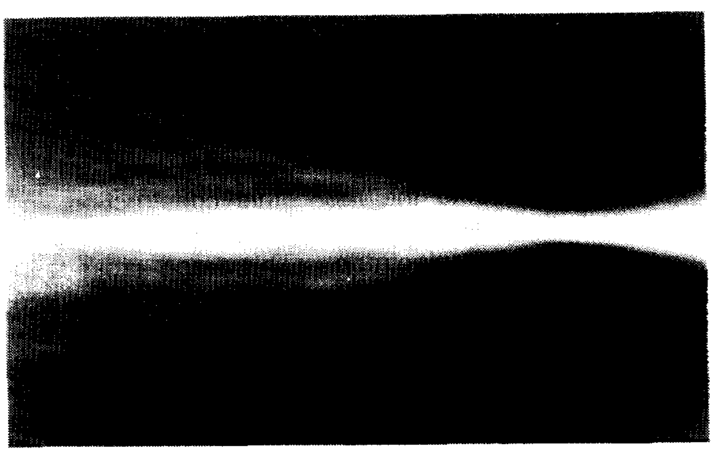

Figure 11. Beam focused farthest from the transaucer, 128mm.

Despite the large cross-talk, the measured focal distances are within 2 mm of the predicted values at each end of the range, and the focal zones are only slightly longer (10%) than we would expect from the simulations assuming acoustic isolation.

Possibly the lack of acoustic isolation is responsible for blurring the near field peak that was predicted for deep focusing (figure 8), as the simulation indicates that this sharp peak is strongly dependent on coherent summation. The high intensity near the beam axis (figure 11) may also be due to cross-talk, but it may be reduced by fine-tuning the amplitudes over the array.

CONCLUSIONS

This transducer is capable of focusing between 8 and 12 cm, and because of its high efficiency it is capable of delivering intensities suitable for extracorporeal HIFU applications. Reducing the cross-talk remains a challenge.

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