Sparse Random Ultrasound Array for Focal Surgery

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Abstract - The feasibility of using novel ultrasound phased arrays consisting of array elements larger than a wavelength and randomly located on a spherical shell have been investigated, both theoretically and experimentally, for surgical applications. Random placement of the sources minimizes grating lobes associated with the array. The size of the treatment volume, i.e., the region over which the focus of the array can be electronically scanned, is determined by the directive, overlapping beams from the circular sources that constitute the array elements. The treatment volume and the intensity at the focus have been determined for a range of frequencies and source dimensions. The source size and frequency determine the number of elements that are needed to generate the intensities required for tissue ablation. Gaussian sources were also evaluated for use in such an array. Results are presented for arrays consisting of 64 elements.

I. INTRODUCTION

High intensity focused ultrasound has been shown to be very useful for surgical treatment (ablation) of tissues [1]-[15]. At frequencies with attenuation coefficients appropriate for surgical treatment, the wavelength is sufficiently small to allow adequate control of the treatment field. All of the early systems, and many current systems, have been based on mechanical movement of a fixed focus beam. Recent research has addressed the development of phased arrays which allow electronic focusing and steering without mechanical movement of the transducer assembly [16]-[21]. Several groups have taken a number of approaches to develop systems which employ phased array technology, but all with a limited number of elements. Phased array systems have been proposed for treatment of the heart [19], the prostate [20] and other regions of the body [18],[21].

The authors recently proposed the use of a sparse random phased array using elements that are large compared to a wavelength, mounted on a spherical shell [21]. Such an array exhibits a treatment volume that is defined by the region where the beams from each of the sources overlap. Within that treatment volume it is possible to steer the focus of the array to treat the desired volume of tissue. Grating lobes are minimized by using a random placement of the sources. The results from the initial study [21] showed good agreement between theory and experiment, as summarized in Section II. The remainder of this paper provides additional theoretical results showing the variation in steering width and peak focal intensity with radius of the sources, frequency and tissue thickness.

II. BACKGROUND

In a previous study [21] the authors examined the use of a phased array for surgical treatment consisting of circular 2.1-MHz sources mounted on a spherical shell of 5-cm radius, and 10-cm radius of curvature. Two arrays were examined. The first, the hexagonal array, consisted of 108 elements, 8 mm in diameter, densely mounted in a hexagonal pattern on the spherical shell. The acoustic field of this array was examined both theoretically and experimentally with 64 pseudorandomly chosen elements excited. The second, the random array, consisting of 64 sources pseudorandomly located on the spherical shell, was examined theoretically.

The theoretical examination of the hexagonal sparsely filled array showed the presence of six significant, hexagonally arranged, grating lobes regardless of the particular subset of 64 elements excited. The presence of such undesirable grating lobes was confirmed experimentally for one representative array of randomly activated elements. The formation of these grating lobes was due to the regular arrangement of sources in a repeated hexagonal pattern. The focus decreased in intensity when steered away from the geometric focus of the array, while any grating lobe that was steered toward the geometric focus increased in intensity. Thus, when the main beam was steered 5 mm off the geometric focus at an intensity above the lesion threshold, one grating lobe was of larger amplitude than the focus and would produce a lesion at the site of the grating lobe. It was clear that the presence of these grating lobes degraded the performance of the hexagonal array and resulted in a beam pattern that was inadequate for surgical treatment which involved steering of the focus. It is desirable to have the intensity for any of the grating lobes on the order of 10 dB lower than the intensity at the focus for the full range of beam steering, so that surgical treatment can be well controlled, with no spurious heating of normal tissues.

In order to more fully investigate the origin of the unacceptable grating lobes observed during the theoretical and experimental examination of the hexagonal array, another array was examined theoretically. This array was "constructed" in exactly the same manner as the hexagonal array, with the important exception that each array element was randomly positioned on the spherical shell. Such an arrangement eliminated the repeated hexagonal pattern hypothesized to be the source of the grating lobes previously observed. The results of the theoretical analysis of this truly random array showed very low amplitude grating lobes with no steering (approximately 14 dB lower than the peak focal intensity) which increased to approximately 8 dB lower than the peak focal intensity for steering of 5 mm, but were within acceptable limits for the designed steering range of $\pm$ 5 mm. The steering range was defined as that for which the intensity at the focus was within 3 dB of the unsteered peak focal intensity and the grating lobes were approximately 10 dB below the peak focal intensity throughout the entire range. Thus, the random array examined should provide adequate performance when treating regions 1 cm or less in diameter. It was clear that the random placement of the elements yielded an array configuration that produced fields without significant grating lobes, for surgical applications. Further, the use of spherical geometry, and random, asymmetric location of elements, minimized grating lobes even when array element dimensions and spacing significantly exceeded λ.

There were several important points noted. It is important to distribute sources randomly over the entire array to maintain the full aperture and its associated focal dimensions. The study suggested that filling the 10-cm aperture spherical shell to approximately 50% of the available active area did not seriously degrade the ability to obtain a sharp focus at intensity levels sufficient to ablate tissue at 2.1 MHz. An increase in sparseness (area of sources excited) would decrease the intensity gain of the array, defined as the ratio of intensity at the focus to average intensity at the surface of the sources. Of course, the choice of the size of each array element would also affect the performance of the array, and the degree of sparseness that could be achieved. The size of elements used in the study (8-mm diameter) limited the distance that the focus could be steered and consequently the size of the region that could be treated to approximately 1 cm in diameter, but showed that this approach could achieve the desired goals for surgical application. Larger steering distances and treatment volumes are possible, simply by decreasing the element size, so that the elements are less directive. This, in turn, would result in less power from each element for a given applied voltage. Thus, for a constant aperture size, either the applied voltage must be increased to compensate, within limits that maintain reliability, or the number of elements must be increased, or both. For some applications it might be desirable to treat regions as large as 2 - 3 cm in diameter.

This earlier study demonstrated that spherical segment phased arrays using large (greater than λ), directive elements randomly spaced on the array surface can provide fields that are free of undesirable grating lobes while providing the necessary power to ablate tissue. Such systems offer the advantages of electronic steering of the focal region without moving the transducer assembly. In addition, the focus can be modified electronically to produce a larger focus if desired. These types of phased array systems offer great flexibility for treating tissues in various locations with widely varying volumes. The strongly focused field produced by a large aperture array can be used when it is critical to avoid damage to nearby structures. However, the focus can be modified to make it larger to facilitate the efficient ablation of large volumes.

III. METHODS

Most of the methods have been discussed in detail previously [21] and will only be outlined here. The non-invasive surgical ultrasound array examined in this study consisted of circular ultrasonic elements mounted on a spherical shell, with a radius of 5 cm, and a radius of curvature of 10 cm. Theoretical calculations of the field pattern of the entire array were made by first determining the field pattern from one of the circular elements in the array and storing the results in a lookup table. The results from this calculation were then used to compute the field from the entire array by a superposition of the fields generated by each element, with appropriate magnitude and phasing applied. The position of the 64 elements of the array were determined pseudorandomly by computer. All results shown here are for one particular pseudorandom set of locations. The field of the array was calculated at a density of 5 points per millimeter in all directions.

The steering width was determined by steering the focus from the geometric focus of the array until the peak intensity at the focus decreased to one-half its value at the geometric focus. All intensity calculations were based on a source velocity amplitude of 0.1 m/s (corresponding to an intensity at the source of approximately 0.77 W/cm^2). Thus, the variation of intensity with various parameters is shown, but the absolute value for peak focal intensity will depend upon the value of source amplitude used.

The effect of tissue attenuation was determined using the arrangement shown in Fig. 1. Degassed water coupling was assumed to be present between the transducer and the plane surface of the tissue. The tissue thickness was defined as the distance from the tissue surface to the location of the geometric focus of the array. For each element a pressure attenuation coefficient of $\alpha = 0.05\ f^{1.1}$ cm^{-1} was applied for the length of its path that was through tissue, where f is the frequency of sound in megahertz.

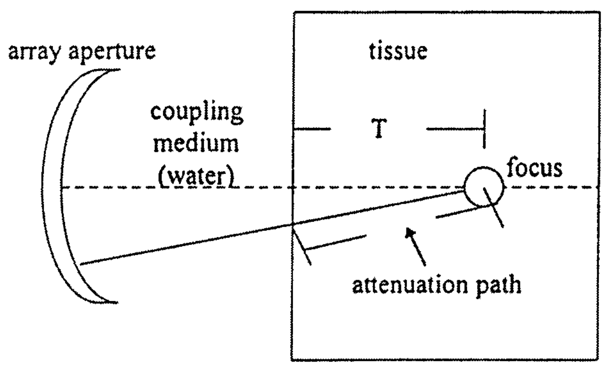

Figure 1. Schematic diagram of array and tissue arrangement.

IV. RESULTS

The effect of various array parameters on lateral steering width and intensity were examined. Figure 2 shows the variation in steering width with the product ka, where k is the propagation constant and a is the radius of each of the array source elements. The steering width decreases with either an increase in frequency or radius of the sources, either of which has the same effect of increasing the directivity. The steering width was independent of the tissue attenuation, and was approximately equal to the beam width of a source at the focal distance for the array (10 cm), for the conditions examined here.

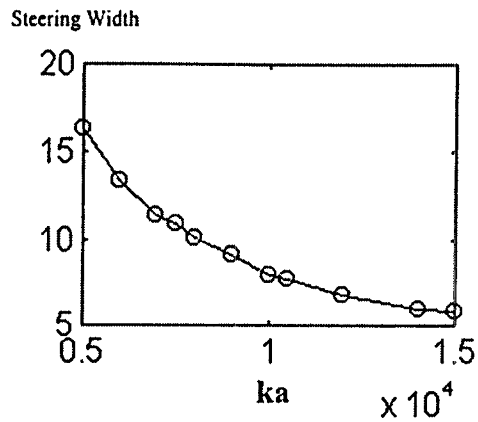

Figure 2. Lateral steering width (mm) versus ka. The steering width is independent of tissue attenuation.

The lateral steering width versus position along the axis of the array is shown in Fig. 3. Although the steering width was greater as the focus was moved away from the geometric focus, the peak focal intensity decreased as shown in Fig. 4. This decreased intensity tends to counter any advantage of increased steering associated with movement along the axis of the array.

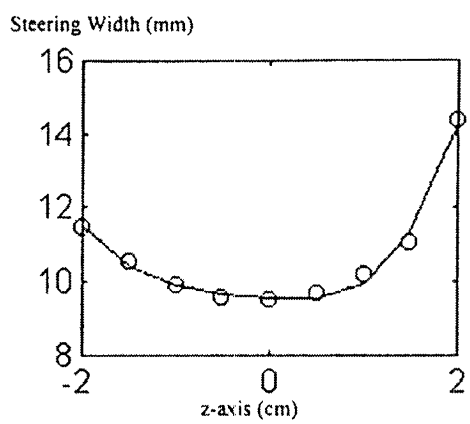

Figure 3. Lateral steering width versus axial position relative to the geometric focus (z = 0 cm).

The peak focal intensity increases with radius of the sources as shown in Fig. 5 for different tissue thicknesses, T. This is again the result of the increased directionality of larger sources. The effect of frequency is demonstrated in Fig. 6 where the intensity is plotted versus frequency for various tissue thicknesses. This plot clearly demonstrates the fact that there is an optimum frequency for treating tissue at a given depth, and that the optimum frequency decreases with increasing depth.

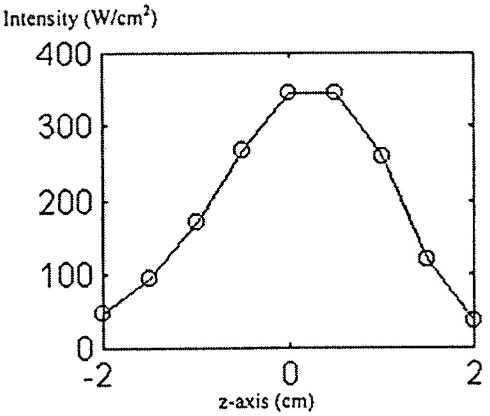

Figure 4. Peak focal intensity versus axial position relative to the geometric focus (z = 0 cm).

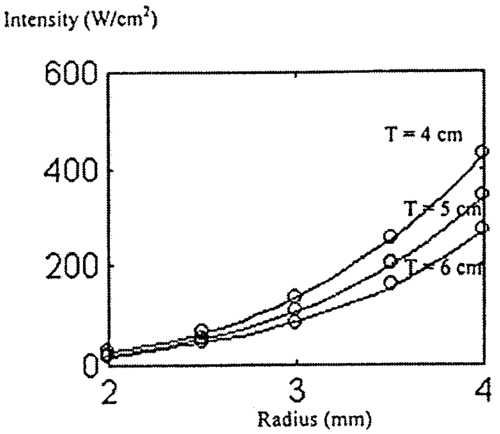

Figure 5. Intensity versus source radius for different tissue thicknesses. The frequency was 2 MHz and the intensity at the face of the sources was approximately 0.77 W/cm^2.

The effect of using a Gaussian source for each array element was examined to determine if such a source might provide an increased steering width over the piston source used in the computations above. Results showed that the steering width was increased slightly, but at the expense of decreased peak focal intensity.

V. CONCLUSIONS

The results of this study show that the treatment volume can be increased from the approximately 1 cm diameter associated with the random array examined

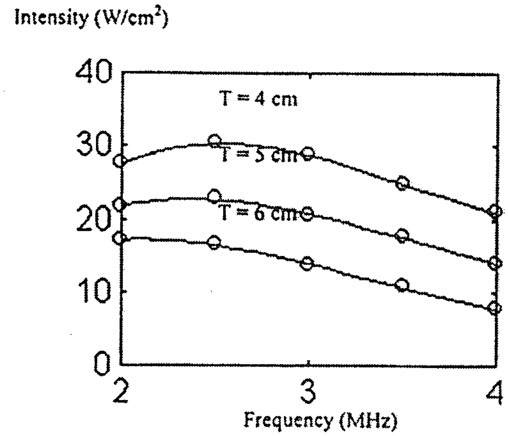

Figure 6. Intensity versus frequency for different tissue thicknesses. The source radius was 2 mm and the intensity at the face of the sources was approximately 0.77 W/cm^2.

previously [21] by decreasing the radius of the sources. However, to compensate for the resultant decrease in peak focal intensity it would be necessary to increase the number of elements. Thus, there is a tradeoff between the size of the treatment volume and the number of sources with associated amplifiers and phasing circuits. The array examined in this study was approximately 50% filled, which seems to be a good goal for future arrays so that the source amplitude required does not exceed the limits for reliable operation of the sources employed.

Future studies should examine the production of multiple focal spots simultaneously to effectively enlarge the focal region. This is desirable to reduce the number of lesion sites required to cover the entire treatment volume, so that treatment can be effected in a reasonable time frame.

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