Efficient Wideband Linear Arrays For Imaging And Therapy

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Abstract— Focused ultrasound surgery has typically been investigated with a large aperture single element dish, or narrowband transducer array, both of which preclude good ultrasound imaging. However, guidance is essential for effective therapy. The goal of this work was to develop linear arrays that have both the high transmit efficiency needed for therapy, yet the wide bandwidth associated with transducers for modern imaging systems. We present analysis and measurements on linear array transducers from 3 to 6 MHz, with measured transmit efficiencies up to 90 percent and pulse-echo fractional bandwidths of 45 to 65 percent. Power measurements on a 3.2 MHz, 5×5.5 mm linear array revealed unfocused continuous acoustic output in excess of 30 watts at greater than 80% efficiency while retaining the array's performance. Human scanning results and ultrasound imaging of phantoms is shown, as well as *in vitro* lesion formation with the 32 to 64 element arrays. By combining these features into one transducer, several benefits result, including real time electronic imaging, therapy and monitoring, lack of spatial misregistration, low cost, multiple-frequency imaging/therapy and the power of acoustic beamforming techniques. The high efficiency allows the transducer to be small in size, enables endoscopic and catheter-based procedures, provides close access to the region of interest with less intervening tissue layers, supports higher frequencies, and yields better spatial resolution for imaging and therapy. The results of the study can help guide the design of such systems.

INTRODUCTION

Ultrasound surgery has historically developed with the use of a large aperture, low-frequency, spherically focused transducer to achieve a large intensity gain and deep penetration [1,2]. The fixed focused single element is mechanically moved in three dimensions to treat a volume, or alternatively in two dimensions such as with annular arrays [3,4]. Imaging or monitoring performance with such transducers has been limited due to their narrow bandwidth, low frequency, and/or need for mechanical scanning. More recently, linear and phased arrays have been investigated for focused ultrasound surgery because of their advantages in spatial and temporal control of the focal spot size [5 – 10]. During the development of therapeutic ultrasound it became obvious that image guidance and treatment monitoring are essential for effective therapy. Several approaches have been investigated such as magnetic resonance imaging and separate transducers for diagnostic imaging and therapy [11, 9]. However this has resulted in various limitations including varied spatial resolution, spatial misregistration of treatment and monitoring, non real-time effects, and additional size and cost. The linear array configuration is well suited for combining functions into a single transducer. A multilayer diagnostic/therapy transducer has been attempted, and a spherically focused 1 MHz 30% fractional bandwidth array has yielded synthetic aperture images about its focus [5, 10].

This work presents the results of the development of a single transducer effective for therapy and conventional imaging. The scope of the work is focused on the transducer performance as opposed to the acoustic configuration of the arrays for a given application. Described below are the methods used to design such transducers, the means of testing them, and imaging and lesioning results.

METHODS

1. Transducer Design

A therapeutic transducer must be capable of large acoustic power outputs. Ultrasound surgery is based upon creating a focal lesion at a desired depth via a preferential acoustic intensity gain. A thermal mechanism is evident for exposure times from 0.1 to 10 seconds at a lesioning threshold of the form, $IT^{0.5} = C$, where I is the spatial peak delivered intensity, T is the exposure duration, and C is a constant typically at or below 500 W/cm^2·s$^{0.5}$ for tissues such as the mammalian liver [12]. Focal intensities of 1500 W/cm^2 are commonly employed for rapid thermal lesioning. As ultrasound waves propagate to the focus they are attenuated on the order of 0.5 dB/cm/MHz = 0.077 dB/wavelength. Thus the required acoustic power output at the source requires an efficient radiator. Issues of azimuthal and elevation focusing, optimum frequency, size of the array, number of elements, intended focal depths, tissue characteristics and other application dependent factors are discussed in the literature [13,14,5–10]. A fundamental trade-off in ultrasound transducer design has been sensitivity or efficiency versus bandwidth [15]. A large acoustic output, such as required for therapy, typically entails low fractional bandwidths. However simulations revealed that good fractional bandwidths (45 – 65%) were possible in an efficient array. Several arrays were fabricated taking into account the power output, heat loss, fractional band-

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width, and pulse-shape requirements for imaging. PZT-4 and PZT-8 or equivalent materials were used as the active transduction material.

2. Transducer Testing

The prototypes were tested for electrical impedance and electrical crosstalk on a network analyzer (Hewlett Packard HP8175A). Pulse-echo sensitivity, center frequency, fractional bandwidth and pulse ringdown time were measured via pulse reflection off of a flat stainless steel plate in a water tank, using a pulser/receiver (Panametrics 5900PR) and digital oscilloscope (LeCroy 9450A) interfaced to a computer. Electric impedance matching transformers or shunting of elements was used to optimize the pulse shape and sensitivity. Acoustic power measurements were made with a reflecting cone type device, (Ohmic Instruments UPM-30). The cone-target was suspended from a 1 mg precision / 1 mg accuracy electronic balance (Ohaus LS120) and interfaced to a computer for automated tests. Electric power was measured via a digital oscilloscope and measurement packages (LeCroy 9354A). Efficiency readings were done after 8 seconds of stabilization and with temporal averaging. First, a frequency sweep was performed at a nominal peak acoustic output of 1 watt. Next, an optimal output frequency was chosen, namely a point at or near peak efficiency. Then high-power amplitude tests were conducted at that frequency until either the efficiency dropped off or the upper measurement limit of the power meter was reached (30W).

3. Imaging Tests

Transducers with 64 elements were interfaced to a real-time ultrasound scanner. Live diagnostic scanning of human volunteers and phantom imaging tests were conducted. The phantom used (Gammex/RMI 404GS LE) was a precision small parts unit with 0.1 mm nylon fibers, target spacing of 0.25, 0.5, 1 and 2 mm, and small gray scale cysts. The phantom had an attenuation characteristic of 0.5 dB/cm/MHz.

4. Lesioning Tests

Lesioning tests were performed with the transducers driven by a 16 discrete-level, phase-focused, FET-based driver circuit. Phases were chosen to place lesions on-axis several millimeters from the transducer. The drive level going to the transducers was monitored to assess power input levels. The transducer was mounted in a support gimbal such that it could transmit face down into a $6 \times 6 \times 14$ cm tank. The tank was filled with degassed, distilled water at 25°C. Electric power was applied from the driver to transducer for 1 to 12 seconds until a lesion was formed in tissue supported in the tank. The tissue was held from 2 to 5 mm from the surface of the transducer. Bovine muscle tissue, turkey breast, and chicken breast tissue were all successfully lesioned. A knife (X-Acto No. 11) was used to section the tissue after a lesion was applied. The lesion size and location were recorded using calipers and a digital photograph taken.

Table 1: Array Parameters.

Type	Number of Elements	Azimuth [mm]	Elevation [mm]
3 MHz	32	30	5
3.5 MHz	21	5.5	5
5 MHz	64	30.5	5
6 MHz	32	20	3

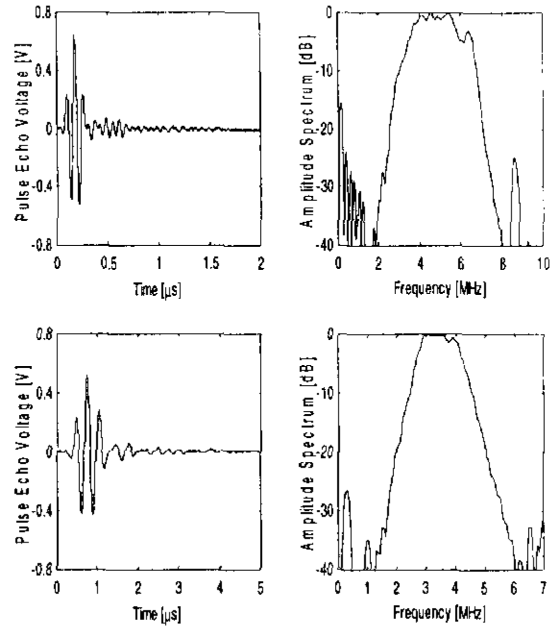

Figure 1: Pulse-echo test results for 5 MHz, 65% fractional bandwidth transducer (top) and 3.5 MHz, 48% fractional bandwidth transducer (bottom).

RESULTS

Fabricated array parameters are shown in Table 1.

Pulse-echo waveform and spectrum for the 5 MHz and 3.5 MHz imaging/therapy probes is shown in Fig. 1. Array crosstalk was below -32.8 dB.

A typical measured power and efficiency response is shown in Fig. 2 for the case of the 3.5 MHz transducer.

Test results for the various transducers are summarized in Table 2.

A phantom image from the 5 MHz, 64 element array is illustrated in Fig. 3.

A lesion was created with the same 5 MHz transducer, and is shown in Fig. 4

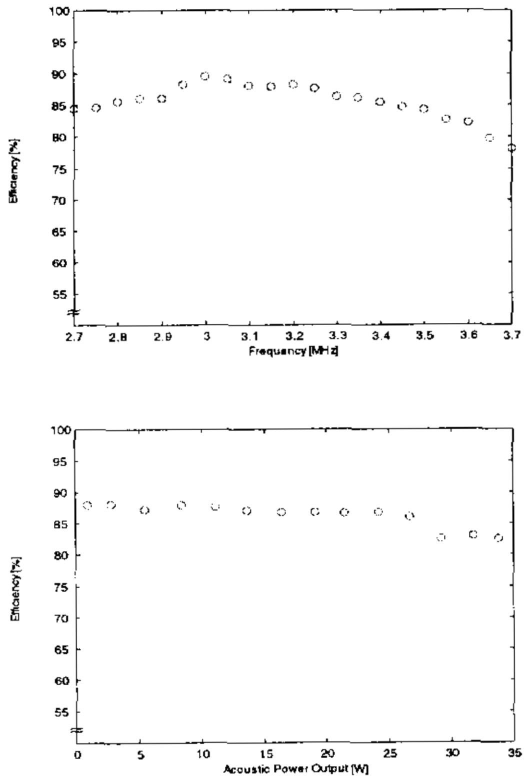

Figure 2: Measured 3.5 MHz imaging/therapy transducer power output and efficiency response. Top: frequency response of efficiency at 1W nominal acoustic output; bottom: efficiency versus acoustic power output at 3.15 MHz.

Table 2: Imaging/Therapy Transducer Bandwidth And Efficiency Test Results. Abbreviations: F_C = pulse-echo center frequency, FBW = pulse-echo fractional bandwidth, η_{PK} = peak transmit efficiency

Array Type	F_C MHz	FBW [%]	η_{PK} [%]
3 MHz	3.32	62	67.0
3.5 MHz	3.47	48	89.7
5 MHz	4.98	65	80.4
6 MHz	5.64	46	83.0

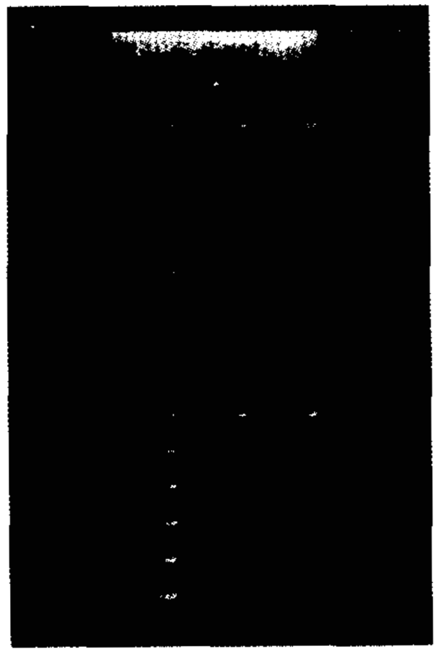

Figure 3: Phantom image created by 5 MHz imaging/therapy transducer without electrical matching and depth setting approximately 9 cm.

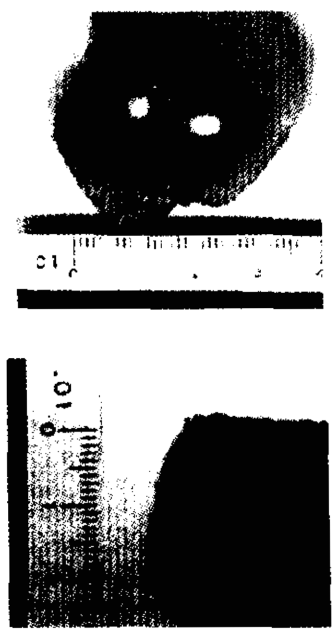

Figure 4: Cross sectioned focal lesion produced by 5 MHz imaging/therapy transducer showing two halves of lesion (top). Deep seated lesion produced with 3 MHz transducer (bottom).

DISCUSSION AND CONCLUSIONS

We have described the results of combined wideband/efficient linear arrays with high performance for both imaging and therapy. Numerous benefits result.

Various types of array configurations can be used to achieve the desired cost/performance objective with minimal or no mechanical movement of the array, resulting in a real time, compact system. Beamforming techniques can be employed, such as apodization to drop sidelobe levels, and aperture and f-number control with depth, i.e. adjustment of the focal spot size and position. Dynamic control of the array in response to feedback such as treatment monitoring can compensate for non-ideal propagation and heating effects including thermal conduction and perfusion. Spatial misregistration and varied resolution among the various modalities (imaging/therapy/monitoring) is avoided. Powerful digital signal processing techniques can be applied to treatment monitoring algorithms with high temporal and spatial quality acoustic signals.

The wide bandwidth makes the arrays less sensitive to the temperature elevation in the array. Therapy can be achieved in the lower (or higher) frequency band depending on the objective, with lower frequencies allowing good beam steering. Imaging detail, contrast, and axial resolution are improved allowing conventional ultrasound imaging with the same transducer and industry evolution to therapy and treatment monitoring. Less narrowband ringdown allows fast switching between therapy and imaging modes on the same transducer element.

The high efficiency allows a smaller transducer and/or less electronic drive to be used to deliver the same acoustic power without thermal fatigue of the transducer or excessive cable transmission loss. Small size transducers can be used in endoscopic and catheter-based procedures, which will allow close access to the tissue of interest with less intervening tissue layers. This yields better acoustic images, and allows the use of higher frequencies with improved spatial resolution and higher cavitation threholds.

Future work will include acoustic configuration of the arrays for various applications (size, frequency), with analysis and measurement of the acoustic beamprofiles and heating patterns.

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