

Design and Evaluation of a Feedback Based Phased Array System for Ultrasound Surgery

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Abstract—A driving system has been designed for phased array ultrasound applicators. The system is designed to operate in the bandwidth 1.2 to 1.8 MHz with independent channel power control up to 60 W (8 bit resolution) for each array element. To reduce power variation between elements, the system utilizes switching regulators in a feedback loop to automatically adjust the DC supply of a class D/E power converter. This feedback reduces the RF electrical power variation from 20% to 1% into a 16 element array. DC-to-RF efficiencies close to 70% for all power levels eliminates the need for large heat sinks. In addition to power control, each channel may be phase shifted 360° with a minimum of 8 bit resolution. To ensure proper operation while driving ultrasound arrays with varying element sizes, each RF driving channel implements phase feedback such that proper phase of the driving signal is produced either at the amplifier output before the matching circuitry or after the matching circuitry at the transducer face. This feedback has been experimentally shown to increase the focal intensities by 20 to 25% of two tested phased arrays without array calibration using a hydrophone.

I. INTRODUCTION

HIGH POWER ultrasound phased arrays have potential in several therapeutic applications [1]–[7]. These arrays can increase the focal necrosis volume through multiple focus patterns [8], [9] and electronically steer foci to reduce the reliance on mechanical positioning systems [4], [10]. The drawback of these arrays is the increased complexity and cost of the driving hardware. Although there is a scarcity of published work on the design of ultrasound phased array driving systems, most designs have used a switching amplifier with duty cycle control of power [11], [12] and the use of counters or delay circuitry to adjust the phase [11], [13]. These systems have performed well in that they are efficient and fairly simple. However, new advances in transducer design have reached a point where the hardware has become a limiting factor for precise field generation. For example, several array designs have recently been investigated which have elements of different sizes and impedances [10], [14], [15]. Switching amplifiers cannot properly drive these arrays without array specific hydrophone calibration [12] because the output power is load

dependent and the electronic phase of these systems depends on output power level. The purpose of this paper is to present a system architecture that can accurately drive a therapeutic ultrasound phased array with various element dimensions such that the need for hydrophone calibration is reduced. Specifically, this paper will discuss the importance of distributed control, electronic element matching, power feedback with a class D/E power converter, and phase feedback to ensure proper electrical phase at the transducer surface.

II. MATERIALS AND METHODS

A. Specifications for a Therapeutic Phased Array System

Therapeutic Transducer Array Description: Phased array transducers of various shapes have been suggested as applicators for both low and high power therapeutic modes. These configurations include annular or concentric ring arrays [3], [16], stacked linear arrays [17], tapered linear arrays [5], cylindrical sectioned arrays [2], and square element spherical sectioned arrays [18]. Although several of these arrays contain elements which are relatively uniform in size and function, new arrays such as the aperiodic linear array by Hutchinson *et al.* [14] purposely use elements of multiple dimensions to decrease undesirable transmission grating lobes. The multiple element sizes have an important impact in the design of an ultrasound driving system—the hardware must be capable of properly controlling the electronic phase and power across transducer loads which have magnitudes in the range of 10 to 10,000 Ω and varying capacitive phases.

Frequency and Power Levels: Most therapeutic ultrasound transducers range in frequency from 0.5 to 10.0 MHz. The precise frequency and power level is determined by the application. Unfortunately, a 0.5 to 10.0 MHz frequency bandwidth can only be implemented using less efficient linear amplifier designs (classes A, B, and AB). These amplifier classes have poor efficiency and high power dissipation, and therefore require large heat sinks and increased system weight and bulk. For a large scale array, the system size becomes unreasonable. By narrowing the specified bandwidth, more efficient amplifiers can be implemented to make the system more manageable. In this application, the frequency range was chosen to be 1.2 to 1.8 MHz, and the output power was specified to be 0 to 60 W per channel with 8 bit resolution.

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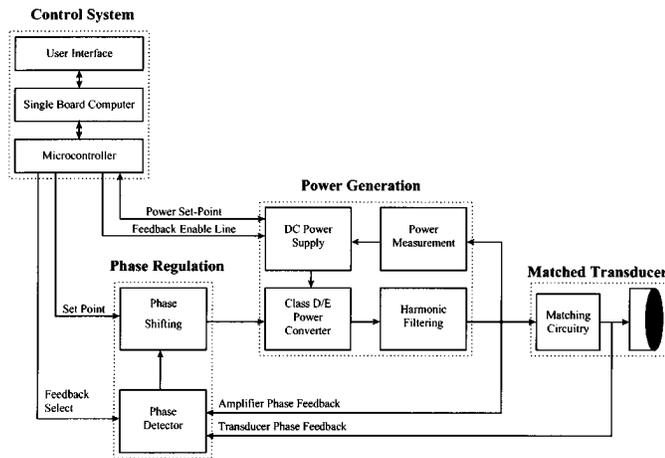


Fig. 1. Block diagram of the phased array ultrasound driving system.

Phase Resolution: The amount of phase resolution needed for therapeutic ultrasound arrays is the topic of some debate. Wang *et al.* [19] suggests that four bit resolution is sufficient for ultrasonic phased arrays due to the phase differences presented in nonhomogeneous tissue. Fan [20], however, has shown that higher resolution is preferable for large scale arrays with complex focal patterns. For this reason, the system design will implement a minimum of 8-bit resolution.

Control: The control subunit of a phased array system performs three essential tasks: it monitors system performance, it modifies system output, and it sets safety interlocks. These tasks are interrelated. For example, the system must be able to monitor the output powers on all of its channels in real time and detect erroneous power levels to ensure patient safety. If a single array element should fail, the system should be able to quickly turn off power to that element without disturbing the rest of the array. For arrays with a small number of elements, this can be done with a simple centralized control system. Monitoring large scale arrays with a single processor, on the other hand, leads to long communication times and slower response. Similarly, electronic scanning of single or multiple foci requires that a system be able to rapidly change the output phase and/or power for all of its channels simultaneously. A centralized control architecture can accomplish this for a small array, but the amount of data bandwidth needed to rapidly communicate with a large number of elements can become unreasonable. For this reason, this design implements a distributed control architecture.

B. Overview of Array Driving System

A block diagram of the array driving system is shown in Fig. 1. The system may be divided into four main units: control and system monitoring, electrical transducer impedance matching, phase regulation, and power conversion. Although this system is not unlike most phased array systems, the implementation of feedback to ensure proper phase and power regulation is previously unpublished for therapeutic ultrasound hardware.

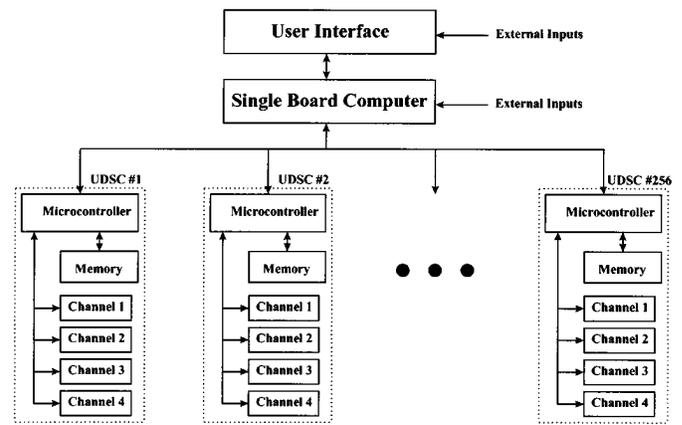


Fig. 2. Distributed control architecture of the phased array driving system.

Control Strategy: This system utilizes a distributed control strategy (Fig. 2). The basic control block is the Ultrasonic Driving System Card (UDSC). This 15.2 cm \times 27.9 cm printed circuit board contains all the hardware to drive four matched transducers. It has a Motorola 68HC11 microcontroller which controls power and phase for the analog hardware, monitors output signals to trigger safety interlocks, and holds individual calibration data for each channel on the UDSC. The cards also contain local read/write memory for a phase and power stack. If it were necessary for the phase and/or power to be changed rapidly during a sonication, such as if the focus were to be scanned, the phase and power data can be downloaded directly to the amplifier's local memory prior to the sonication. A single pulse can then trigger a step in the stack index, and therefore change the power and phase for the entire array. This dramatically reduces the communication overhead during sonication, which in turn allows the microcontrollers to more closely monitor the amplifiers.

All the microcontrollers interface over a single bus with a x486 based single board computer to report operational status and to receive operational commands from the user interface. This single board computer is dedicated strictly to communicating with the microcontrollers and interpreting commands from the user interface. This strategy is necessary because the user interface may be occupied with external interfaces such as a magnetic resonance imager. In addition, the dedicated single board computer allows more timely communication with a larger number of UDSC (the system is designed to implement up to 256 UDSC corresponding to 1024 channels). All of the control architecture is based on the principle of modularity so that the same hardware may be used for arrays with different numbers of elements (i.e., individual UDSC may be added or removed from a given system).

Electrical Transducer Impedance Matching: Electrical impedance matching for individual transducers is advantageous for three reasons. First, matching increases the maximum power transfer from the amplifier into the transducer. Second, the power delivered into a matched load can

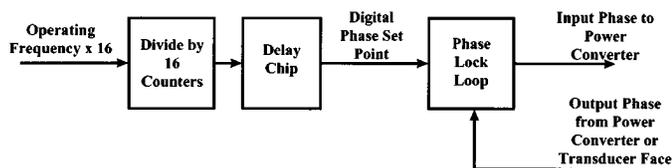


Fig. 3. Phase regulation system.

be measured using simple circuitry (see Appendix). Third, matching elements of varying impedances ensures that the same range of power can be delivered to each individual element in the array. Since system control and treatment monitoring are essential aspects of this design, the small increase in circuitry (two passive elements) is justified and easily implemented.

Power Conversion: This system implements a separate class D/E power converter for each amplifier channel to convert a digital input signal and a DC source to a high power, high frequency sinusoid. The class D [21] and class E [22] switching amplifiers are based on the same principle: use active switching devices (FETs) to drive a resonant circuit while avoiding appreciable current flow through the device when there exists a voltage drop across it. The theoretical maximum efficiency for each of these converters is 100% [23], although the efficiency decreases as a function of bandwidth and load variation. To reduce the extraneous harmonic content of the output signal, a low pass filter is added as an output stage. It is this filter which determines the bandwidth for the system and a modification of this filter can change the operating range of a given power converter.

Feedback is used to compensate for nonlinearities inherent in class D and class E amplifiers. The power feedback signal is obtained from a dual directional coupler [11], [24] which measures the forward and reflected power accurately for a $50\ \Omega$ load (see Appendix). The forward feedback signal is then fed to a voltage switching regulator which adjusts the DC supply to the class D/E converter such that the desired RF power is achieved. The feedback signal also is used to trigger microcontroller interlocks which monitor unreasonably high reflected power (as in the case of a failed transducer element). More efficient power conversion using a duty cycle controlled class D amplifier was rejected due to its inherent increase of undesirable harmonics leading to a decrease in power measurement accuracy.

Phase Control: Several methods have been proposed to phase shift the output signal [13], [25], [26]. The simplest method uses preloadable counters similar to those used by Ngo [27]. Unfortunately, as pointed out in Lovejoy *et al.* [13], 8-bit phase resolution using this technique requires a master clock frequency 256 times the ultrasound frequency, increasing complexity and decreasing reliability. Lovejoy *et al.* [13], therefore, recommends the use of a discrete delay based system. This system implements a combination of both counters and delay circuitry. Fig. 3 is a diagram of the phase regulation unit. The input master clock operates at 16 times the frequency of the transducer (typically

24 MHz for a 1.5 MHz transducer). This clock is applied to simple preloadable four bit counter to create phase steps of 22.5 degrees. The other four bits of resolution are created using a delay chip (8 bits of 0.5 ns steps). This combination of counters and delay circuitry is effective because it increases phase resolution while avoiding ultra high frequency master clock signals and a significant increase in chip count.

Like power control, feedback is necessary to ensure proper phasing of a class D/E amplifier. This method uses a phase locked loop (PLL) based feedback loop to adjust the input digital clock of the power stage to regulate the phase of the high power output sine wave [28], [29]. The feedback signal can be obtained from either the matching circuitry of the transducer or directly from the transducer face so that the matching delay is eliminated [12]. The system also can be modified to receive its feedback from an external source such as a hydrophone.

C. System Characterization and Measurement Techniques

Measurements into a $50\ \Omega$ Load: A Bird $50\ \Omega$, 200 W dummy load was used to characterize the system. Individual channel efficiencies were calculated as the RF power delivered to the load divided by the DC power to that power converter channel. In all cases, the RF power was measured using a Hewlett Packard 438A Power Meter with a Werlatone (C1373) coupler. The system frequency response was measured using a Hewlett Packard 8590A Spectrum Analyzer, and waveform measurements were recorded using a Tektronix TDS 380 Oscilloscope. Transducer impedances were measured using a Hewlett Packard 4193A Vector Impedance Meter.

Measurement in Transducer Loads: The ultrasound driving system was experimentally tested using several transducer arrays (see Table I for descriptions of arrays). The arrays contained between 14 and 62 elements with multiple element sizes in each array. The variety of transducer elements was used to demonstrate the capability of the system to control power and phase with several element sizes and shapes. The unmatched transducer impedance values ranged between 20 and $1000\ \Omega$ in magnitude and were always capacitive. Acoustic measurements were made with a 0.5 mm hydrophone (Precision Acoustics, LTD) or with a radiation force technique [30]. The same hydrophone was used to obtain a phase calibration signal to compare the system response both with and without acoustic feedback. This was accomplished by placing the hydrophone at a specified focal location of an array and calibrating the phase of each element of that array such that the measured acoustic signals were coherent.

III. RESULTS

A. Class D/E Converter Efficiency

At 1.5 MHz the average DC-to-RF efficiency into a dummy load was measured to be 78% at 60 W, and

TABLE I
ARRAYS USED TO TEST THE ULTRASOUND DRIVING HARDWARE.

Array Design	Number of Elements	Frequency	Reference
Spherically Sectioned	16	1.64	[31]
Sector/Concentric	52	1.5	[15]
Concentric Ring	14	1.5	[10]
Aperiodic	62	1.07	[1]

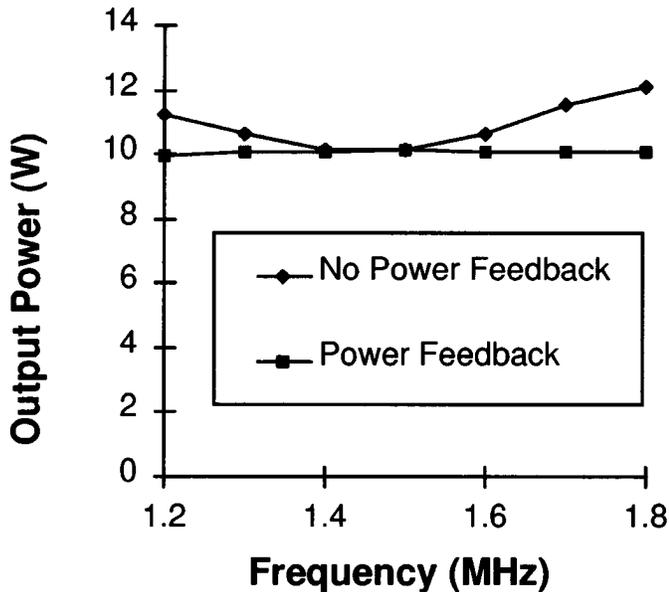


Fig. 4. Output power into a 50 Ω dummy load with and without power feedback. When feedback was not used, the DC supply to the power converter was set such that at 1.5 MHz the output power would be 10 W.

dropped to 68% for output levels below 2 W. The main losses occur in the voltage switching regulator and the ferromagnetics of the filter of the class D/E converter. To further improve efficiency, larger magnetics could be used, but the added bulk of the magnetics would be greater than the decrease in required heat sinking. Similar efficiencies are found throughout the operating bandwidth.

B. Power Output/Regulation of the Class D/E Power Converter

The effect of power regulation is illustrated in Fig. 4. For a desired output of 10 W into a dummy load, the power regulation feedback lowers the maximum error from 20 to 1% in the specified amplifier bandwidth (1.2 to 1.8 MHz). For frequencies below the system bandwidth (1.2 MHz) the regulation yields larger errors due to the higher harmonic content of the output signal. The high frequency limit of the amplifier is determined by the maximum specified output power. This power level drops rapidly when operating above 1.8 MHz due to the low pass filter cutoff.

Although transducers are matched to 50 Ω , a tuned amplifier such as class D or class E will still suffer a variation in power due to different transducer impedances off reso-

nance. For example, each element of a 16 square element array [31] was matched to 50 Ω at the array's resonant frequency (1.64 MHz). When each of its elements was driven individually with the same amplifier with a fixed supply voltage (no power feedback), the measured output power varied 20% (4.75 to 5.80 W). By implementing power feedback, the output power variation decreased to less than 1% (5 ± 0.08 W).

C. Harmonic Content of Output Sinusoid

In the frequency band 1.2 to 1.8 MHz, the highest harmonic measured while driving a 50 Ω dummy load is 36 dB lower than the primary signal (this occurs at 1.2 MHz). When the harmonics are greater than -30 dB (at frequencies below 1.2 MHz), the power measurement capability of the system is decreased and the power regulation has decreased efficacy.

D. Power Measurement Dependence on Transducer Matching

To correctly measure and regulate power using a dual directional coupler [24], the transducer must be matched using LC circuitry such that its impedance at the operating frequency is 50 Ω (the standard impedance of a dual directional coupler). The coupler yields two output signals representing the forward and reflected power delivered to the load. If the impedance is exactly 50 Ω , there is no reflected signal and the forward power accurately measures the power delivered to the load. If the impedance varies from 50 Ω , the measured forward power will be greater than the actual power delivered to the transducer. By regulating the output power using the measured forward power, the system will never deliver more than the specified power. To test this, the matching circuit of a transducer was varied such that the load impedance differed from the ideal 50 Ω . The acoustic output power of the mismatched transducer was then measured using radiation force measurements for a constant regulated power level. The results are plotted in Fig. 5 with the center contour indicating the theoretical point where the actual acoustic output power is 10% less than the desired power (see Appendix for theoretical details). Power regulation, therefore, guarantees that the power delivered to a mismatched transducer will not exceed the programmed output power.

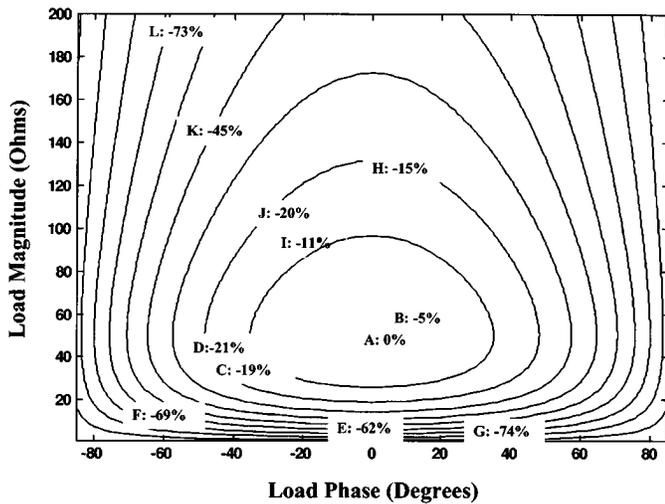


Fig. 5. Acoustic power regulation dependency on proper transducer matching. The contour lines mark 10% increments where the actual power output is lower (by that percentage) than the set point power. The letters plot the mismatched transducer impedances whose output power was measured for a given set point power. The percentage indicates the drop in acoustic power vs. the properly matched 50 Ω transducer load.

E. Output Phase Response

The output phase is characterized by three parameters: range, resolution, and jitter. The output range of the system is 360° for all frequencies in the bandwidth. The phase resolution is 0.5 ns (0.27° at 1.5 MHz) resulting from the delay circuitry. The output phase, however, has some jitter caused by the locking of the PLL. This jitter ranges 3 to 8 ns across the frequency bandwidth.

F. Phase and Power Relationship for a Class D/E Converter

A class D/E power converter does not maintain the phase of the input over the entire output range of the amplifier [28]. This means that the output phase will depend on the output power level. As a typical class D or class E amplifier, the phase varies 48° from 0 to 60 W in this system without feedback. Feedback from either the amplifier output or the transducer face reduces this error to less than 3°.

G. Effect of Phase Feedback on Acoustic Fields

To achieve maximum power transfer and to accurately measure output power in this system, transducer loads must be matched to 50 Ω . Unfortunately, the matching network introduces a phase shift between the amplifier output voltage and the transducer [12]. If all array elements are exactly the same impedance, then this shift is constant and unremarkable. If the elements are different shapes or sizes, this shift will vary [12], [32]. For example, the measured phase shift ranged from 29 to 94° for a concentric ring array [10] and 30° for an aperiodic array [1]. Phase feedback using the transducer voltage as

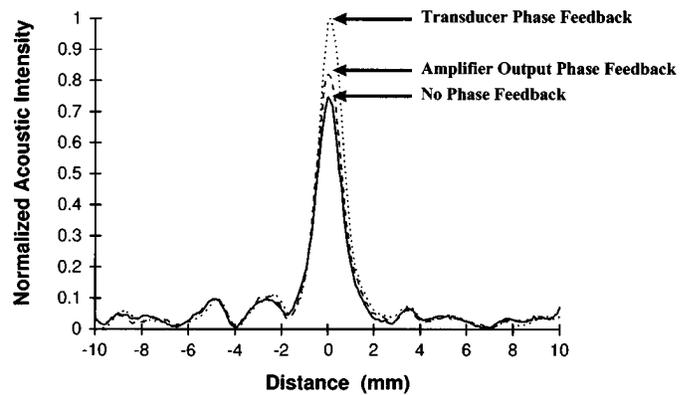


Fig. 6. Hydrophone scan of acoustic intensity across a single focus for the aperiodic array. The focus was located 4 cm from the center of the array and the scan was performed parallel to the array at that depth. The scans were repeated for each type of phase feedback at the same power and phase inputs. Intensities were normalized using the peak measurement of the three scans.

the feedback signal automatically compensates for these shifts. Figs. 6 and 7 illustrate the acoustic effects of applying phase feedback. A line scan across the single focus of the aperiodic array (Fig. 6) demonstrates that phase feedback from the transducer face for a linear array can increase peak focal intensities by 25% compared to no feedback and 18% for feedback that is taken from the amplifier output stage. Fig. 7 contains a contour plot of the acoustic fields measured using a hydrophone for the combined sector vortex/concentric ring array. The implementation of phase feedback decreases undesirable foci and increases the desired peak intensities by an average of 20% indicating an improvement in the control of the acoustic field.

H. Comparison of System Response With and Without Acoustic Feedback

To compare the system response with and without acoustic feedback, a hydrophone was placed 4 cm from the center of the 62 element, aperiodic, linear array [1]. Each element was then individually powered and the acoustic phase at the focus was recorded by the hydrophone. Using these measurements, a new phase distribution for the array was implemented such that all of the acoustic signals were coherent at the focus. Fig. 8 compares the acoustic intensities at the focus when the array is driven without any feedback, with electronic feedback from the transducer surface, and with the hydrophone acoustic calibration data. The data shows that hydrophone feedback can increase the acoustic intensity by 53% over the system without any feedback but that the improvement drops to 32% compared to the system with electronic feedback from the transducer face.

I. System Response Times

As stated previously, this system has memory for each microcontroller (see Fig. 2) which can be preloaded with

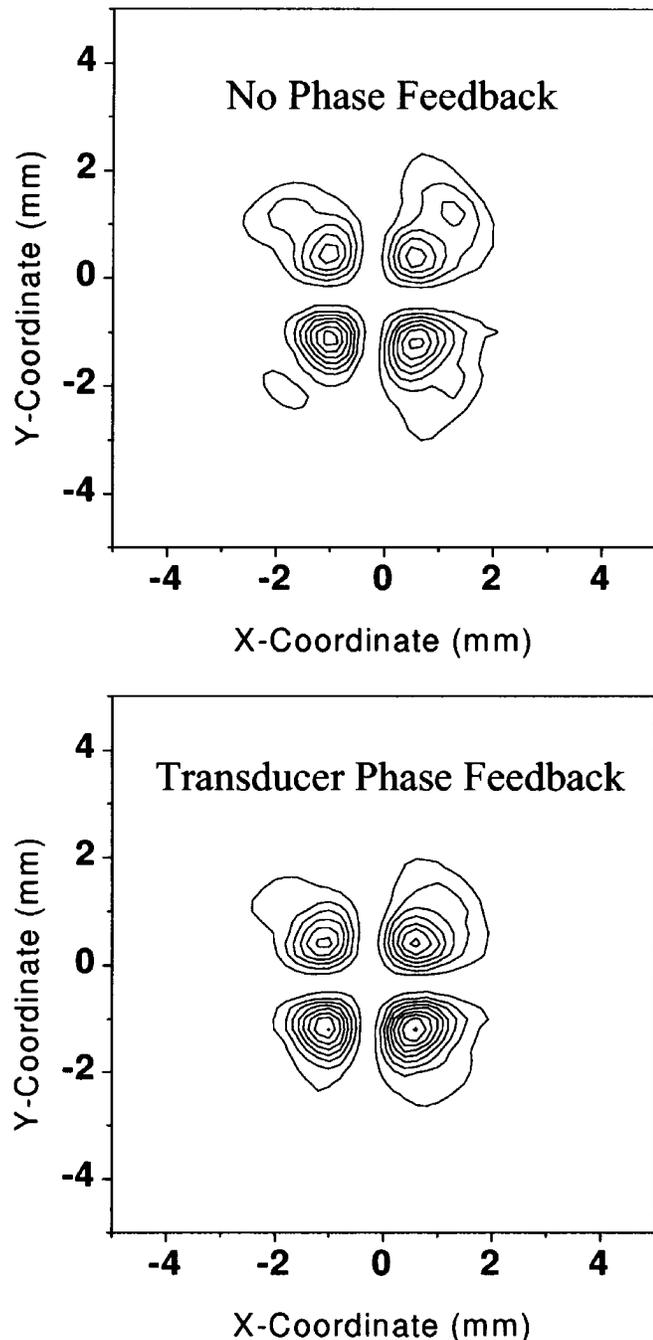


Fig. 7. Multiple focus hydrophone scans across the focus of a concentric ring/sector vortex array without phase feedback and with phase feedback. The acoustic intensity peaks of the feedback scan are 20% higher on average than those without feedback. The contour lines correspond to equivalent acoustic intensity amplitudes.

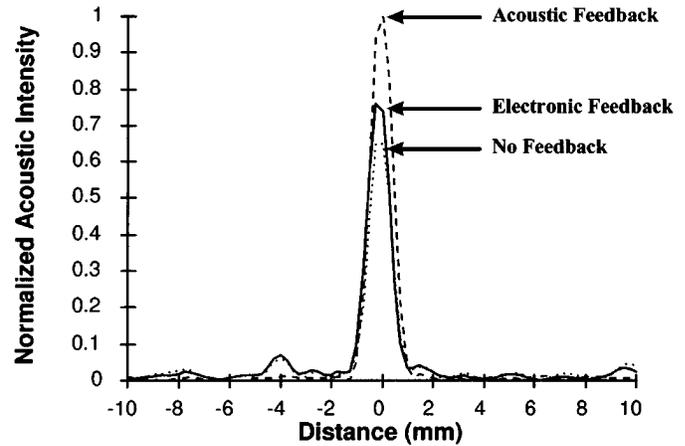


Fig. 8. Hydrophone scan of acoustic intensity across a single focus 4 cm from the center of the aperiodic array [1]. The acoustic feedback signal was obtained by placing the hydrophone at the focus and adjusting the phases of each element individually such that all the acoustic signals were coherent. The resulting signal for the entire array (acoustic feedback) is compared to the array response without hydrophone correction (both with no feedback and with electronic feedback from the transducer surface).

a stack of power and phase settings. A single pulse on the bus triggers a change in output phase and power set by the values of that stack. Following that trigger, output power will settle within 1% of a step input of 1 W and 10 W in 175 μ s and 22 ms, respectively. The power feedback is almost critically damped so there is minimal power overshoot. The output phase with feedback locks within 250 μ s. Therefore, an accurate system output occurs in less than 250 μ s for a 1 W step input and 22 ms for a 10 W step input. The next stack value is available from the microcontrollers within 20 ms such that another trigger pulse may be received.

If a faster response time is needed, then the system can operate with a disabled power feedback loop. The power settling time is then 5.6 μ s and 9 μ s for a 1 W and 10 W step input. This makes the system response time approximately 250 μ s for all output power levels with 20 ms needed for the microcontrollers to update the stack.

IV. DISCUSSION AND CONCLUSION

The ultrasound array driving system described in this paper is able to accurately produce RF signals of appropriate power and phase for arrays of multiple element sizes and frequencies without requiring array specific calibration. This system marks an improvement both in ultrasonic control and in patient safety. By implementing phase and power feedback, the electronic nonlinearities of previous systems can be alleviated without necessitating a change to a less efficient amplifier design. Due to their high efficiency, a 256 channel system is about the size of a medium filing cabinet, making this system suitable for a clinical setting. The distributed control architecture gives the system a fast response time, allowing for proper treat-

ment monitoring and electronic focal scanning for a large scale array.

An important aspect of this design is the measurement of the power delivered to each transducer element. Because tissue necrosis is a logarithmic function of temperature [33], small errors in power generation and/or measurement can greatly effect the tissue response. For that reason, individual power measurement is needed for each array element to accurately control treatment conditions. Simply measuring the total output power for entire arrays cannot offer this critical information.

As a result of improved and individual power measurement, automatic power regulation is now possible. It has been shown that the variance in output power between different transducers driven by class D or E amplifiers can be decreased by implementing a simple feedback loop. This is especially important for phased array fields which rely heavily on destructive interference of same magnitude fields [3], [34]. By measuring the power into a matched 50 Ω transducer load, the variation of acoustic power output is limited to the variation of the electroacoustic efficiency of the elements—a characteristic which is easily measured with the radiation force technique. In addition, by regulating the power to each element individually, nonuniform electrical powers can be delivered to array elements. The concentric ring and combined concentric ring/sector vortex arrays used in this research [10], [15] were driven such that each element had the same acoustic power per unit area although this required that the individual elements would have nonuniform electrical powers (the elements have varying surface areas). Similarly, an array can be driven with nonuniform acoustic powers (per cm^2) if that were desirable to optimize focal patterns.

Power regulation also acts as a safety feature because the output power of a mismatched transducer will be regulated at or below the desired output power level. However, if a load were not perfectly matched to 50 Ω , then it could still be excited close to a desired power level by manually increasing the set-point power of the individual amplifier channel until the difference between the measured forward and reflected power is equal to the desired power level. This compensation can be used for nonideally matched array elements without decreasing significantly the DC-to-RF efficiency of the system.

Phase feedback is also important to the operation of arrays whose elements have varying sizes or output power requirements. Without feedback, the user loses control of accurate phasing for a class D or E amplifier between multiple power levels, and hence the ability to precisely control the acoustic field patterns. An uncalibrated variable delay caused by the matching circuitry for transducers also can reduce array performance. Both of these sources of error are overcome by the implementation of simple feedback circuitry. For the arrays tested in this research, the electronic feedback can improve focal intensities by 20 to 25% over techniques that do not use any feedback.

The electronic feedback system also was compared to an acoustic feedback technique. When a hydrophone was

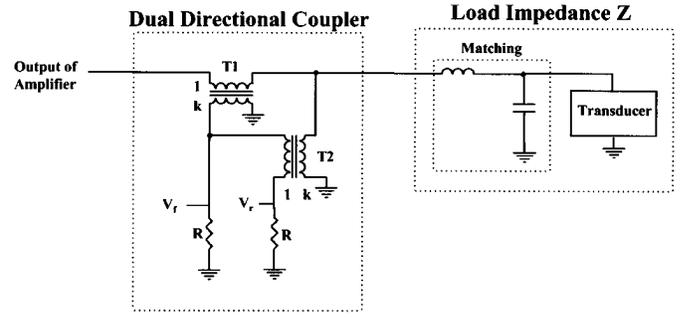


Fig. 9. Dual directional coupler diagram.

placed at a specified focus to calibrate the aperiodic array [1], it was found that the acoustic intensity could be improved 30% over the use of electronic feedback alone (see Fig. 8). The focal diameter between the two techniques, however, is almost identical. It is understood that acoustic feedback can help eliminate variable acoustic phase shifts that electronic feedback cannot. These include shifts caused by the acoustic properties of the transmission medium, by misaligned array elements, and by time delays between the electrical and mechanical oscillations of the transducer material. The main cause of the variable phase shifts, however, could not be accurately determined in this research using the hydrophone since the diameter of the hydrophone (0.5 mm) was over one-third the size of the acoustic wavelength in water (1.4 mm) and positioning errors of only 0.1 mm yielded phase errors of 25°.

In conclusion, it has been shown that, although electronic feedback does not eliminate the utility of acoustic feedback, it can significantly improve the focal intensity patterns of several therapeutic phased array devices. As a note, the system described in this paper can directly implement acoustic feedback by replacing the electronic phase signal of the transducer with the phase signal of a hydrophone preamplifier. However, it is the experience of the authors that adequate power deposition can be obtained in vivo without the use of hydrophone calibration in several cases [1], [10], [15], [31], and that the improvement of focal patterns using electronic feedback further decreases the need for invasive hydrophone feedback.

APPENDIX DUAL DIRECTIONAL COUPLER DESIGN

A circuit diagram of the dual directional coupler used to measure power is found in Fig. 9. Assuming that the transformers are ideal with turns ratio of k and setting the coupler standard impedance R to 50 Ω , then the function of power delivered to the load P_L may be written as

$$P_L = (V_f - V_r) \frac{k^2}{R} \left[V_f^* + \left(1 + \frac{1}{k^2} \right) V_r^* \right] \quad (1)$$

where V_f is the forward power signal and V_r is the reflected power signal (* represents complex conjugation). These

voltages in turn are determined by the load impedance Z_L and driving voltage V_D as written in the following equations:

$$V_f = -V_D \frac{\left(1 + \frac{1}{k^2} \frac{Z}{R} + \frac{Z}{R}\right)}{\frac{1}{k} + \left(\frac{1}{k^3} + \frac{2}{k} + 2k\right) \frac{Z}{R}} \quad (2)$$

$$V_r = -V_D \frac{\left(1 - \frac{Z}{R}\right)}{\frac{1}{k} + \left(\frac{1}{k^3} + \frac{2}{k} + 2k\right) \frac{Z}{R}} \quad (3)$$

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