Abstract—In medical ultrasound imaging, twodimensional (2-D) array transducers are necessary to implement dynamic focusing in two dimensions, phase correction in two dimensions and high speed volumetric imaging. However, the small size of a 2-D array element results in a small clamped capacitance and a large electrical impedance, which decreases the transducer signal-to-noise ratio (SNR). We have previously shown that SNR is improved using transducers made from multi-layer PZT, due to their lower electrical impedance. In this work, we hypothesize that SNR is further increased using a hybrid array configuration: in the transmit mode, a 10 Ω electronic transmitter excites a 10 Ω multi-layer array element; in the receive mode, a single layer element drives a high impedance preamplifier located in the transducer handle. The preamplifier drives the coaxial cable connected to the ultrasound scanner. For comparison, the following control configuration was used: in the transmit mode, a 50 Ω source excites a single layer element, and in the receive mode, a single layer element drives a coaxial cable load. For a 5×102 hybrid array operating at 7.5 MHz, maximum transmit output power was obtained with 9 PZT layers according to the KLM transmission line model. In this case, the simulated pulse-echo SNR was improved by 23.7 dB for the hybrid configuration compared to the control. With such dramatic improvement in pulse-echo SNR, low voltage transmitters can be used. These can be fabricated on integrated circuits and incorporated into the transducer handle.

I. INTRODUCTION

For medical ultrasound imaging, two-dimensional (2-D) array transducers have greater versatility than conventional linear arrays. Using 2-D arrays, one can improve the quality of clinical ultrasound images through several techniques: dynamic focusing in two dimensions [1], phase correction in two dimensions [2],[3], and real-time volumetric imaging [4]. In order to implement these arrays, however, several challenges must be addressed. One problem is that the small 2-D array elements have a low clamped capacitance and a large electrical impedance near the resonant frequency. This results in poor signal-to-noise ratio (SNR) of the transducer.

To address this problem, one can use transducers fabricated from an N layer structure of PZT, with the layers connected acoustically in series and electrically in parallel. For this multi-layer ceramic (MLC), clamped capacitance is multiplied by a factor of N² and the impedance by 1/N² compared to a single layer transducer of the same dimensions.

It has been demonstrated that MLC transducers have greater SNR than single layer transducers, due to their higher capacitance and lower electrical impedance [5][7]. In our previous research, we developed prototype 2-D array elements with 16 active layers [6] and a complete 3 layer, 3×43 array [7] using thick film ceramic technology. For individual array elements, the pulse-echo SNR was improved by 20 dB for the 16 layer elements and up to 7.3 dB for the 3 layer elements, compared to single layer control elements. Simulations with the KLM model [8] were consistent with these measurements. In addition, phased-array images were made in vivo using the 3×43 arrays. Images obtained with the MLC array had improved SNR compared to the control.

We have also developed a method to optimize the multi-layer transducer design for improved SNR [9]. Given the transducer material characteristics and dimensions as well as the source and receiver configuration, one can determine the number of piezoelectric layers for maximum pulse-echo SNR. While the optimization technique used a simplified circuit model, the simulated results agreed with those from the KLM model and experimental measurements. In our previous research, however, several assumptions were made that limited the improvement in pulse-echo SNR: (1) each multi-layer array element was used in both the transmit mode and the receive mode; however, for maximum SNR the ideal transmit element may have a different number of layers than the ideal receive element [9]; (2) the electronic transmitters had a 50 Ω source impedance (the conventional impedance used in analog systems); (3) the preamplifiers were located in the ultrasound scanner; as a result, the receive elements were driving a coaxial cable load that connected the transducer to the scanner.

In this paper, we consider a hybrid array design which eliminates the above constraints. This array incorporates separate designs for transmit and receive elements to further increase the transmit output power and receive SNR. As a result, an individual element can be used for transmit or receive but not both. This is possible for a sparsely sampled array [10],[11] that has transmit and receive apertures with no elements in common.

For a given source voltage, transmit output power is increased when the source impedance approaches zero. Such an ideal source is not practical, but it is reasonable to fabricate a transmitter with a 10 Ω source impedance. In this configuration, maximum output power is obtained by
matching the element impedance to the source impedance with a 10 Ω, multi-layer array element.

In the receive mode, there is no general rule for determining the best configuration for increasing SNR. Instead, modeling techniques can be used to compare the receive SNR for several configurations.

We hypothesize that significant improvements in pulse-echo SNR can be obtained using the hybrid array in the following configuration: in the transmit mode, 10 Ω electronic transmitters exciting 10 Ω multi-layer elements; in the receive mode, single layer elements driving high impedance JFET preamplifiers located in the transducer handle. To test this hypothesis, we designed a 7.5 MHz, 5 × 102 element array. The array was sparsely sampled and the hybrid design incorporated multi-layer transmit elements and single layer receive elements on the same ceramic substrate.

For this work, we used the following control configuration: in the transmit mode, a 50 Ω source exciting a single layer element, and in the receive mode, a single layer element driving a coaxial cable load. This corresponds to a conventional ultrasound scanner configuration.

We used the KLM transmission line model and a circuit analysis technique to compare the performance of the control to several configurations. In the transmit mode, we used a 10 Ω transmitter exciting 10 Ω elements. In the receive mode, we modeled each of the following: bipolar transistor preamplifiers located in the transducer handle, MOSFET amplifiers in the handle, and JFET amplifiers in the handle.

This paper also describes our proposed implementation of the hybrid array. With dramatic improvements in pulse-echo SNR, a conventional high voltage transmitter (i.e., 100 V excitation) located in the ultrasound scanner is no longer necessary. Instead, adequate SNR is obtained using low voltage transmitters (i.e., 10 V) with a low source impedance (i.e., 10 Ω). These could be fabricated on integrated circuits (IC’s) and incorporated into the transducer assembly. This would eliminate the need for a separate coaxial cable between the transducer and scanner for each transmit element.

II. ANALYSIS

A. Transmit Signal

In the transmit mode, a simplified circuit analysis at the resonant frequency is sufficient for accurate simulations. In a previous paper [9], we expanded Kim’s circuit analysis [12] to apply to multi-layer piezoelectric transducers. This analysis is summarized here for clarity. In Fig. 1, a simplified circuit is shown for an N-layer transducer operating at series resonance and transmitting into low acoustic impedances [9]. The electronic transmitter has a source voltage of \( V_{in} \) and a source impedance of \( R_0 \). The transducer has a clamped capacitance of \( N^2C_0 \) and a mechanical resistance of \( R_m/N^2 \).

![Fig. 1. Equivalent circuit for an N layer transducer operating in the transmit mode near series resonance; the source impedance is \( R_0 \) and the transducer impedance is \( N^2C_0 \) in parallel with \( R_m/N^2 \).](image1)

The power dissipated through \( R_m/N^2 \) is proportional to the acoustic output power from the transducer, \( P_{out} \).

\[
P_{out} = \frac{|Z_t|^2}{N^2R_m/R_0 + Z_t/N^2}V_{in}^2
\]

For a given \( R_0 \), maximum power output is obtained for the value of \( N \) that satisfies \( \frac{dP_{out}}{dN} = 0 \). This occurs when \( N = \sqrt{|Z_t|/R_0} \) or the transducer impedance, \( |Z_t/N^2| \), is matched to the source impedance, \( R_0 \). In addition, \( P_{out} \) is further increased by reducing \( R_0 \) toward zero.

B. Receive Signal

In the receive mode, a broadband analysis of signal and noise is performed. The circuit of Fig. 2 shows the receive array element and its electrical loads. The element impedance is represented as \( Z_i/N^2 = (R_i + jX_i)/N^2 \) where \( R_i \) and \( X_i \) are the real and imaginary transducer impedances for a single layer element computed by the KLM model. This impedance is frequency dependent and takes into account the acoustic loads of the transducer as well as its dielectric and mechanical losses. The incident acoustic pressure is modeled as a voltage source with an open circuit voltage of \( V_0/N \). A coaxial cable load was approximated by a single shunt capacitor, \( C_{cable} \). This approximation was reasonable because the cable was very short relative to a wavelength. At the resonant frequencies of our transducers, the preamplifier input impedance is dominated by the input capacitance, \( C_{in} \). The noise
sources are evaluated below. They are represented in this circuit by $e_n$, $i_n$, and $e_n$.

For computing receive signal, it is assumed that the preamplifier input impedance is large compared to the transducer impedance. The receive voltage, $V_r$, at the preamplifier input is related to the open-circuit receive voltage, $V_0/N$, by a voltage divider equation

$$V_r = \frac{N}{jX_{\text{cable}} + Z_t/N^2} \cdot \frac{V_0}{N}$$  \hspace{1cm} (2)

where $X_{\text{cable}}$ is the reactance of the cable capacitance $(-1/\omega C_{\text{cable}})$ and $Z_t/N^2$ is the transducer impedance. In the above equation, $X_{\text{cable}}$, $Z_t$, and $V_0/N$ are all functions of $\omega$ and can be determined by the KLM model.

If $X_{\text{cable}}$ is held constant, then the maximum $V_r$ is computed for the value of $N$ that satisfies $\frac{dV_r}{dN} = 0$. This occurs when $N = \sqrt{|Z_t/LX_{\text{cable}}}$ or the transducer impedance, $|Z_t/N^2$, is matched to the cable impedance, $X_{\text{cable}}$. However, if the preamplifiers are moved to the transducer handle, effectively removing the cable from these computations ($X_{\text{cable}}$ is infinity), then $V_r$ reaches its maximum value of $V_0$ when using a single layer transducer ($N = 1$).

C. Receive Noise

Both the transducer and preamplifier contribute to the receive noise in our system. With the reasonable assumption of statistical independence of these noise sources, we used the principle of superposition to consider each one separately.

The transducer noise is generated thermally from acoustic loading on the front and back faces of the PZT as well as in the dielectric and mechanical losses in the PZT. These loads and losses are represented in the model by the real part of the transducer impedance, $R_t$. As a result Johnson (thermal) noise has a value of

$$\langle \epsilon_n^2 \rangle = 4kT R_t \text{(units of V}^2/\text{Hz})$$  \hspace{1cm} (3)

where $k$ is Boltzmann’s constant ($k = 1.38 \times 10^{-23}$ J/K), and $T$ is the temperature in Kelvin [13]. From analyzing the circuit of Fig. 2, one can calculate the noise spectral density of transducer noise as seen at the preamplifier (units of V$^2$/Hz):

$$\langle \delta n^2 \rangle = \frac{4kTR_tN^2}{(N^2 - \omega(C_{\text{cable}} + C_{\text{in}}) X_t)^2 + (\omega(C_{\text{cable}} + C_{\text{in}}) R_t)^2}$$  \hspace{1cm} (4)

where $R_t$ is the real part of the transducer impedance and $X_t$ is the imaginary part of the transducer impedance as a function of frequency, $\omega$. Also, $N$ is the number of layers in the receive element, $C_{\text{cable}}$ is the cable capacitance and $C_{\text{in}}$ is the preamplifier input capacitance. According to (4), transducers with a lower electrical impedance also have lower noise. In addition, the SNR depends on how efficiently the transducer drives the load.

The preamplifier noise can be represented by a noise voltage, $e_n$, and a noise current, $i_n$, as shown in Fig. 2 [13]. From analyzing the circuit of Fig. 2, one can calculate the contribution of these noise sources as seen at the preamplifier (units of V$^2$/Hz):

$$\langle \delta e_n^2 \rangle = \frac{(e_n^2)(N^2 - \omega X_{\text{cable}} X_t)^2 + (\omega X_{\text{cable}} R_t)^2}{(N^2 - \omega X_{\text{cable}} + C_{\text{in}}) X_t)^2 + (\omega X_{\text{cable}} + C_{\text{in}}) R_t)^2}$$  \hspace{1cm} (5)

$$\langle \delta i_n^2 \rangle = \frac{(i_n^2)(R_t)^2 + (X_t)^2}{(N^2 - \omega X_{\text{cable}} + C_{\text{in}}) X_t)^2 + (\omega X_{\text{cable}} + C_{\text{in}}) R_t)^2}$$  \hspace{1cm} (6)

The total noise spectrum is computed by summing the contributions from the three primary noise sources (units of V$^2$/Hz):

$$\langle \delta^2 \rangle = \langle \delta e_n^2 \rangle + \langle \delta i_n^2 \rangle$$  \hspace{1cm} (7)

Finally, the root-mean-square noise at the input of the amplifier is calculated over the bandwidth of the transducer (units of V):

$$V_N = \sqrt{\frac{1}{2\pi} \int_{BW} \langle \delta^2 \rangle d\omega}$$  \hspace{1cm} (8)

III. METHODS

A. Array Element Design

We have designed a 5 x 102 = 510 element array that is sparsely sampled in the transmit and receive modes. The 1.5-D array was designed to focus and steer in the azimuth dimension, but only focus in the elevation dimension perpendicular to the azimuth plane. The element dimensions are 0.130 x 1.500 x 0.2 mm for an operating frequency of 7.5 MHz. With a saw kerf width of 0.025 mm, the interelement spacing is 0.155 mm in azimuth and 1.525 mm in elevation. Therefore the 1.5-D array has a total aperture size of 15.8 x 7.63 mm.

The array was designed to be fabricated using our standard fabrication techniques for 2-D array transducers [14]. The acoustic backing is light epoxy ($Z = \text{acoustic impedance} = 3.25 \text{ MRays}$). The quarter wave matching layer is 0.7 mm thick and made of conductive epoxy ($Z = 5.1 \text{ MRays}$). A 2 $\mu$m thick sheet of silver foil provides an electrical ground and protects the array elements.

B. Array Element Modeling

Performance of single and multi-layer elements was simulated on the KLM transmission line model (Piezocad, Sonic Concepts, Woodinville, WA). The calculations were made in the frequency domain using 128 data points from 0 to 20 MHz. Table 1 lists the material characteristics of the materials used in the transducer assembly. For modeling, the backing thickness was assumed to be infinite.

Fig. 3 shows the electrical loads used for modeling the arrays. A 2.5 pF shunt capacitance modeled the effect of the wires connected to the array elements. A transmitter of source impedance $R_0$ was used to excite an element. The
TABLE I

Acoustic Characteristics of Materials Used in Transducer Assembly
(Listed from the Front of the Assembly to the Back.)

<table>
<thead>
<tr>
<th>Material</th>
<th>Sound speed V (m/sec)</th>
<th>Acoustic impedance Z (MRays)</th>
<th>Thickness t (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water</td>
<td>1540</td>
<td>1.5</td>
<td>infinite</td>
</tr>
<tr>
<td>Silver foil ground</td>
<td>3650</td>
<td>38.0</td>
<td>0.002</td>
</tr>
<tr>
<td>Silver epoxy matching layer</td>
<td>1900</td>
<td>5.1</td>
<td>0.19</td>
</tr>
<tr>
<td>PZT-5H</td>
<td>3750</td>
<td>30.00</td>
<td>0.61</td>
</tr>
<tr>
<td>Silver epoxy glue bond</td>
<td>1900</td>
<td>5.1</td>
<td>0.01</td>
</tr>
<tr>
<td>Light epoxy backing</td>
<td>2880</td>
<td>3.25</td>
<td>10</td>
</tr>
</tbody>
</table>

**Fig. 3.** The electrical loads used in the KLM model: a voltage source, $V_0$, with an impedance $R_0$ excites the transducer. The transducer includes a 2.5 pF shunt capacitance to model the effect of the wires that connect to individual array elements.

receive elements were driving a high impedance preamplifier. The effects of a coaxial cable load and the input capacitance of the preamplifier were considered in the equivalent circuits discussed above (see Fig. 2 and (2), (4)–(6)).

The following types of amplifiers were modeled: bipolar (Comlinear CLC425), MOSFET (custom design [15]), and JFET (National Semiconductor J310). Table II lists the values for $r_n$ and $i_n$ for each amplifier, as obtained from the manufacturer data sheets. The input capacitance, $C_{in}$, was 3 pF for each amplifier.

Multi-layer elements were modeled by inserting an ideal transformer with a turns ratio of 1:N adjacent to the array element. Previous studies have shown that this is a reasonable approximation to the behavior of multilayer transducers [7,9].

**IV. RESULTS**

**A. Design for Increased Transmit Signal**

The control configuration on transmit was a 50Ω source driving a single layer array element. The transmit output power can be increased by using a lower source impedance and matching the element impedance to the source using multiple layers.

We evaluated the array at its center frequency of 7.5 MHz. From KLM model computations, a single layer element had an impedance magnitude of $|Z_t| = 1.36\Omega$ and a resistance of $R_{m} = 930\Omega$ at 7.5 MHz. Fig. 4(a) shows the graph of $P_{out}$ vs. $N$ vs. $R_0$ from (1). All values were normalized to the control transducer configuration of $R_0 = 50\Omega$ and $N = 1$ layer. Maximum $P_{out}$ was obtained with $R_0 = 0\Omega$ and $N = \infty$ (a zero transducer impedance matched to a zero source impedance).

It is not practical to fabricate a transmitter with zero source impedance. Therefore, we determined that an impedance of 10 Ω would be used for this application. Fig. 4(b) shows $P_{out}$ vs. $N$ when $R_0 = 10\Omega$. Maximum output power occurs when the element impedance is matched to the source impedance, or
The Voltage and Current Noise for Each of the Preampifiers Tested.

<table>
<thead>
<tr>
<th>Preamplifier type</th>
<th>Manufacturer/part #</th>
<th>$\epsilon_n (nV/\sqrt{Hz})$</th>
<th>$i_n (pA/\sqrt{Hz})$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bipolar</td>
<td>Comlinear/CLC425</td>
<td>1.05</td>
<td>1.6</td>
</tr>
<tr>
<td>MOSFET</td>
<td>Custom design [15]</td>
<td>9.00</td>
<td>*</td>
</tr>
<tr>
<td>JFET</td>
<td>National Semiconductor/J310</td>
<td>2.20</td>
<td>*</td>
</tr>
</tbody>
</table>

*NA, < 1pA/\sqrt{Hz}*

![Graphs of noise spectrum vs. frequency for different preamplifiers](image)

Fig. 5. Simulated noise spectrum showing the transducer noise, $<v_t^2>$, amplifier noise voltage $<v_{en}^2>$, and amplifier noise current, $<v_{in}^2>$, for a single layer element driving each of the following preamplifiers: (a) bipolar transistor amplifiers located in the ultrasound scanner (coaxial cables connect each array element to the amplifiers); (b) bipolar transistor amplifiers located in the transducer handle; (c) MOSFET amplifiers located in the handle; (d) JFET amplifiers located in the handle.

$$N = \sqrt{\frac{Z_t}{R_0}} = 11\text{layers.} \quad (9)$$

From Fig. 4(b), it is apparent that with $N = 9$ layers, the relative transmit signal was 15 dB, which is within 1 dB of the peak transmit signal. We chose to use 9 layers instead of 11 to simplify transducer fabrication. Similar results were obtained with a broadband analysis using the KLM model.

B. Design for Maximum Receive SNR

To improve the receive SNR, we evaluated receive SNR vs. number of layers for the transducer driving several different loads: 2.4 m long coaxial cable, terminated by bipolar transistor preamplifiers in the ultrasound scanner (control configuration); bipolar transistor preamplifiers located in the transducer handle; MOSFET preamplifiers in the handle; and JFET preamplifiers in the handle.

For each configuration, the KLM model calculated the receive signal at the preamplifier input vs. frequency. From this data, the rms receive signal, $V_{sig}$, was computed by summing the signal across the transducer bandwidth:

$$V_{sig} = \sqrt{\sum_{BW} V_r^2} \quad (10)$$

where BW is the −6 dB fractional transducer bandwidth and $V_r$ is the receive signal at the preamplifier input (from
Fig. 6. Simulated SNR in receive as a function of number of layers, $N$, for the 1.5-D array element. The following configurations were modeled: bipolar transistor amplifiers located in the ultrasound scanner; bipolar transistor amplifiers located in the transducer handle; MOSFET amplifiers located in the handle; JFET amplifiers located in the handle.

The bandwidth was determined by the KLM model and varied from 65 to 80%, depending on the preamplifier configuration.

For determining the rms noise, the KLM model was used to compute the electrical impedance vs. frequency for a single layer element in each of the above configurations. Then, the noise contributions contributed by $\langle v_i^2 \rangle$, $\langle v_c^2 \rangle$, $\langle v_t^2 \rangle$ were calculated using (4)-(6). Fig. 5(a–d) show these noise spectra for a single layer receive element used in each of the four preamplifier configurations.

The total noise $\langle v_{noise}^2 \rangle$ was determined from (7) and the rms noise was computed using the following equation:

$$V_{total} = \sqrt{\frac{\sum_{BW} \langle v_{noise}^2 \rangle \cdot \Delta f}{\Delta f}}$$

using BW as the bandwidth (shown by thick vertical lines in the figures).

This process was repeated for each configuration while varying the number of layers from 1 to 11. The SNR was computed from the ratio of (10) and (11). Fig. 6 summarizes the results for receive SNR vs. $N$. Results are normalized to the control configuration of a single layer element driving a coaxial cable terminated by bipolar transistor preamplifiers in the scanner. For this preamplifier configuration, SNR was improved by 3.7 dB if a 3 layer receive element was used instead of a single layer element.

In the other preamplifier configurations, the peak SNR was obtained using a single layer element. For bipolar transistor amplifiers in the transducer handle, SNR was improved by 8.4 dB, and for MOSFET amplifiers, SNR was improved by 2.4 dB. The greatest receive SNR was obtained using a single layer element driving JFET amplifiers in the handle. The improvement in this case was 8.7 dB compared to the control.

Fig. 7. Simulated impedance magnitude and phase with a water load (KLM model): (a) single layer element; (b) MLC element.

C. Complex Transducer Impedance vs. Frequency

For single elements of the control and the MLC, the electrical impedance at the element was simulated using the KLM model. The impedance is shown in Fig. 7 with a water load on the front of each element. The single layer elements had an impedance of 1.1 kΩ at series resonance of 7.5 MHz (Fig. 7(a)), compared to 10Ω for the MLC elements (Fig. 7(b)).

D. One-way Impulse Response

Fig. 8 shows the KLM model simulations of the transmit impulse response for the single layer control element and
Fig. 8. Simulated transmit impulse response (KLM model): (a) 50Ω source exciting a single-layer element; (b) 10Ω source exciting an MLC element.

Fig. 9. Simulated receive impulse response (KLM model): (a) single-layer element driving a coaxial cable terminated by bipolar transistor amplifiers located in the ultrasound scanner; (b) single-layer element driving JFET amplifiers located in the handle.
the 9 layer element. For a 50 Ω source exciting a single
layer element, the -6 dB pulse length was 0.16 msec and
a -20 dB pulse length was 0.46 msec. For a 10 Ω source
exciting a multi-layer element, the -6 dB pulse length was
0.15 msec and the -20 dB pulse length was 0.25 msec. The
-6 dB bandwidth was 69% for the control and 80% for the
multi-layer transmit element.

The simulated receive impulse response is shown in Fig.
9. For a single layer element driving bipolar transistor am-
plifiers in the scanner, the -6 dB pulse length was 0.18
msec and a -20 dB pulse length was 0.48 msec. For a sin-
gle layer element driving JFET amplifiers in the handle, the
-6 dB pulse length was 0.15 msec and the -20 dB pulse
length was 0.61 msec. The -6 dB bandwidth was 75% for
the control and 71% for the JFET preamplifiers.

V. DISCUSSION

A. Comparison of Receive Configurations

The graphs in Fig. 6 compare the receive SNR for four
different preamplifier configurations. These graphs can be
understood from analyzing (2)−(6) and the resulting
graphs in Fig. 5, which were derived from the circuit
diagram in Fig. 2. As N increases, the element impedance
becomes smaller by a factor of 1/N².

In the control case, the element is driving the reactance
of the coaxial cable, approximately 100 Ω at series reso-
ance. As N increases, the receive signal increases as the
transducer drives this load more effectively. At larger N,
however, the receive signal starts to decrease as it varies
as 1/N (2). The receive noise is dominated by <v²
em> at
all N. As a result, the SNR peaks when N = 3.

In the other configurations, the preamplifiers are in the
handle so C_cable can be considered zero and the pream-
plifier input capacitance, C_in, is also small (3 pF). As a
result, the receive signal is maximum when N = 1 and it
varies as 1/N for increasing N. In addition, <v²
em> re-
mains approximately constant with increasing N (5) and
<v² > and <v²
in> vary approximately as 1/N² (4) and (6).
Therefore, as N increases, the receive noise becomes
dominated by the relatively constant <v²
em> and changes
in SNR are dominated by changes in receive signal.

These trends can be seen in Fig. 6. When the pream-
plifiers are located in the handle, the SNR is maximum
for a single layer transducer (N = 1). For this case, the
JFET amplifiers have the lowest total noise, resulting in
the greatest SNR of the configurations we tested. At higher
values of N, the SNR asymptotically approaches a 1/N
curve. Since the bipolar amplifiers have the lowest <v²
em> of all the amplifiers, they have the greatest SNR for large
N.

For single layer transducers, the SNR of the JFET am-
plifiers is similar to that of the bipolar transistor amplifi-
cers. However, the JFETs are still preferred because of
their lower power consumption, which leads to lower heat
dissipation. For a proposed 256 channel receive system,
JFET amplifiers would consume a total of 1.28 W com-
pared to 38.4 W for the bipolar transistor amplifiers. This
is a significant consideration since we propose to have these
preamplifiers in the small space of the transducer handle.

B. Improvement in SNR

Single layer 1.5-D arrays typically have an electrical
impedance of less than 1 kΩ; therefore, the mismatch to
a 50 Ω source is not severe. We have previously shown
that with 1.5-D arrays, a modest improvement in pulse-
echo SNR (i.e., an increase of 5 to 10 dB) is sufficient to
obtain improved in vivo ultrasound images [7]. This was
obtained using a 50 Ω source exciting an MLC element,
and the MLC element driving a coaxial cable load on re-
ceive.

In this work, several constraints were eliminated from
the previous design. A hybrid configuration was used which
optimized separately for maximum transmit output power
and maximum receive SNR. Table III summarizes our re-

results. According to KLM model and simplified circuit sim-
ulations, the hybrid array has 15 dB greater transmit out-

put power and 8.7 dB greater receive SNR than the control configuration. This provides a total of 23.7 dB increase in SNR.

Even greater improvements in SNR are possible with 2-D arrays that can focus and steer in both azimuth and elevation. Since a 2-D array element is smaller than a 1.5-D element, the problems of high electrical impedance and poor SNR are worse for 2-D arrays. We repeated all of the above computations for a proposed 40 x 40 hybrid array, with element dimensions of 0.315 x 0.315 x 0.485 mm and operating at 3.0 MHz [16]. The simulation results are shown in Fig. 10. The performance was compared to a single layer array in the same control configuration as above. In this case, transmit output power was increased by 25 dB for a 25-layer element. Receive SNR was increased by 11 dB when a single-layer element was driving a JFET preamplifier located in the transducer handle. In practice these elements may be difficult to fabricate.

C. Proposed Implementation

The fabrication technique for a multi-layer PZT array has been described previously [7]. For the hybrid ceramic array, the technique can be modified so that a single ceramic substrate contains both single-layer and multi-layer elements. The N layer hybrid array is fabricated from N layers of PZT “green sheets” as shown in Fig. 11. At the locations of the multi-layer elements, internal electrodes are screen printed onto each green sheet. A pair of vias provides interlaminar connections for each MLC element. In the regions of single-layer receive elements, internal electrodes would not be screen printed. During the high temperature firing, the N ceramic layers would fuse into a single layer of homogeneous PZT in these regions.

With dramatic improvements in pulse-echo SNR, new configurations are possible for the transducer and scanner using sparsely sampled 2-D arrays [10], [11]. The block diagram of Fig. 12 shows a conventional ultrasound scanner and transducer assembly. The sparsely sampled array contains q transmit elements and r receiver elements, which are connected to the scanner via q+r coaxial cables.
TABLE III
Summary of Results for Transmit Output Power and Receive SNR.

<table>
<thead>
<tr>
<th>Configuration</th>
<th>Transmit Output Power</th>
<th>Receive SNR</th>
</tr>
</thead>
<tbody>
<tr>
<td>50Ω source, single layer transmit element</td>
<td>0 dB</td>
<td></td>
</tr>
<tr>
<td>10Ω source, 9 layer, 10Ω transmit element</td>
<td>15 dB</td>
<td></td>
</tr>
</tbody>
</table>

On the Duke ultrasound scanner, currently 192 transmit elements and 64 receive elements are used [11]. The number of elements will continue to increase in future years to improve the SNR and resolution of such systems. The number of coaxial cables will rise accordingly. This will significantly increase the cost and complexity of manually attaching each cable to the transducer assembly. In addition, the bulk of the coaxial cables will make it difficult for the clinician to manipulate the transducer handle. This problem can be alleviated by relocating the transmitters to the transducer handle to reduce the number of coaxial cables.

Conventional ultrasound systems have a 100 V, 50 Ω transmitter driving a 50 Ω array element. In this configuration, the transmitter generates an instantaneous power of 100 W, or an average power of 0.4 W assuming a typical duty factor of 1/250. Half of this power is dissipated in the transmitter source impedance and half is radiated/dissipated by the transducer. For a 256 channel system, this corresponds to an average power of 51 W (256 channels * 0.2 W/channel) dissipated in the transmitter source impedance. As a result, the transmitter circuitry must be located in the ultrasound scanner, where it can dissipate the large amount of heat generated. On the other hand, if low voltage transmitters could be used, they could be dramatically reduced in size and located in the transducer assembly.

Fig. 13 shows the proposed implementation of the transmitter and receiver circuitry. The 10 V, 10 Ω transmitter circuits are located in the transducer assembly. When driving a 10 Ω array element, each transmitter generates an instantaneous power of 5 W. For 256 transmitters, the total average power dissipated in the source impedance is 2.6 W for the hybrid configuration, compared to 51 W for the control case described above.

Several IC's contain the q transmitters as well as the transmit control logic and memory. This circuitry is activated by a few digital control lines from the ultrasound scanner. In the receive mode, the high impedance preamplifiers are also manufactured on IC's, which are located in the transducer assembly. Each of the r preamplifiers drives a separate coaxial cable, which connects to the delay and summation circuitry in the ultrasound scanner. The total number of coaxial cables is r.

In Fig. 14, the proposed transducer assembly is shown with the IC transmitters and preamplifiers. The single and multi-layer hybrid ceramic is bonded to a polyimide flexible circuit connector [17]. This is used in place of conventional wire guides [14], and it is similar in design and concept to the thick film, multi-layer alumina connectors previously designed and tested at Duke [18]. The bare IC dies are bonded to the connector using advanced chip to package interconnection techniques, such as wire bonding or flip chip [19]. These IC's are connected to the array elements by traces in the flex connector. Additional traces also connect to pads at the back of the transducer assembly, which are connected to the transducer handle. The region behind the array elements is filled with a light epoxy backing.

Several new challenges must be overcome to implement this design. Our previous work indicates that fabrication of the multi-layer PZT is feasible [6],[7]. However, it has been noted that receiver preamplifiers are difficult to implement in the transducer handle because of space limitations and increased power dissipation [7]. These problems have recently been alleviated with the design of preamplifiers that have low power consumption [15].

Despite these difficulties, this configuration adds flexibility to the ultrasound scanner. The number of transmit elements can be increased without the cost of additional coaxial cables. As a result, the transducer resolution and SNR can be improved.

VI. SUMMARY

A 5 × 102, multi-layer PZT array has been designed and simulated with 9 layer transmit elements and single-layer receive elements operating at 7.5 MHz. The array is sparsely sampled so that the transmit and receive apertures contain no common elements. Therefore, the transmit SNR and receive SNR are optimized separately. For maximum pulse-echo signal, the following configuration is used: in the transmit mode, a 10 Ω source exciting a 9-layer array element; in the receive mode, a single-layer array element driving a high impedance JFET preamplifier located in the transducer assembly.
This hybrid array consists of single and multi-layer elements on the same ceramic substrate. Fabrication of this array is currently under study. In addition, a single layer control array was designed with similar parameters.

Performance of the MLC elements was compared to the single-layer elements using KLM model simulations. In the transmit mode, the MLC array elements were excited by a 10Ω source. The simulated transmit signal increased by 15 dB compared to the control. In the receive mode, the single layer array elements were driving a high impedance preamplifier. The measured receive signal improved by 8.7 dB compared to the control. Therefore, the pulse-echo SNR increased by 23.7 dB compared to the control.

With the implementation of this 1.5-D MLC array, new scanner configurations are possible. If a 100 V transmitter was used in the control configuration, then the same pulse-echo SNR is obtained with a 10 V source in the improved configuration. Therefore, low voltage transmitters could be used to excite the MLC elements. These transmitters could be fabricated on an integrated circuit and located in the transducer assembly. This would greatly reduce the system complexity by eliminating the coaxial cables connecting the transmit elements to the scanner.

REFERENCES


Richard L. Goldberg was born in Hackensack, NJ on February 5, 1966. He received the B.S. degree in electrical engineering in May 1988 from Bucknell University, Lewisburg, PA. In 1994, he completed the Ph.D. degree in biomedical engineering at Duke University, Durham, NC. For his dissertation, he designed, fabricated and tested ultrasound transducer arrays made of multilayer piezoelectric ceramic.

Currently, he studies echolocating and hearing in a living ultrasound system, the mustached bat. He is at the Department of Cell Biology and Anatomy, University of North Carolina at Chapel Hill.

Charles D. Emery (S’86) graduated from Washington University in St. Louis summa cum laude in 1992 with B.S. degrees in Electrical Engineering and Physics. He was an NSF-REU Summer Fellow in 1990 and 1991 at Hope College in Holland, Michigan where he studied dispersed fluorescence spectroscopy of formaldehyde in the Department of Chemistry. He is currently working towards a Ph.D. at Duke University in the Department of Biomedical Engineering with an emphasis in diagnostic ultrasound. Current areas of interest include optoelectronics, fiber optics, microelectronic fabrication, and multilayer ceramics. Mr. Emery is a member of IEEE.

Stephen W. Smith (M’91) was born in Covington, KY on July 27, 1947. He received the B.A. degree in physics in 1969 from Iowa State University, Ames, and the Ph.D. degree in biomedical engineering in 1975 from Duke University, Durham, NC. In 1969, he became a Commissioned Officer in the U.S. Public Health Service, assigned to the Food and Drug Administration, Center for Devices and Radiological Health, Rockville, MD, where he worked until 1990 in the study of medical imaging, particularly diagnostic ultrasound and in the development of performance standards for such equipment. In 1978, he became adjunct Assistant Professor of Radiology at Duke University Engineering and Radiology, and Director of Undergraduate Studies in Biomedical Engineering at Duke University. In 1996 he became Professor of Biomedical Engineering. He holds 14 patents in medical ultrasound and has authored 100 publications in the field. Dr. Smith has served on the education committee of the American Institute of Ultrasound in Medicine, the executive board of the American Registry of Diagnostic Medical Sonographers, and the editorial board of Ultrasonic Imaging. He was co-recipient of the American Institute of Ultrasound in Medicine Magna Cum Laude Award in 1988 and 1990 and co-recipient of the IEEE-UFFC Outstanding Paper Award in 1983 and 1994.