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High-Density Flexible Interconnect for Two-Dimensional Ultrasound Arrays

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***Abstract*—We present a method for fabricating flexible multilayer circuits for interconnection to 2-D array ultrasound transducers. In addition, we describe four 2-D arrays in which such flexible interconnect is implemented, including transthoracic arrays with 438 channels operating at up to 7 MHz and intracardiac catheter arrays with 70 channels operating at up to 7 MHz. We employ thin and thick film microfabrication techniques to batch produce the interconnect circuits with minimum dimensions of 12-μm lines, 40-μm vias, and 150-μm array pitch. The arrays show 50-Ω insertion loss of -60 to -84 dB and a fractional bandwidth of 27 to 67%. The arrays are used to obtain real time, in vivo volumetric scans.**

I. Introduction

Two-dimensional array ultrasound transducers have received much attention in recent years and offer numerous advantages for medical imaging over linear arrays [1], [2]. The literature reports 1.5-D arrays operating up to 3.5 MHz with -6 dB fractional bandwidths around 70% [3], [4], and 2-D arrays operating between 2.5 and 5.0 MHz with bandwidths ranging from 30 to 60% [5]–[9]. However, interconnection to the large number of individual channels in these arrays poses engineering challenges, particularly when the intended application, such as cardiac imaging, limits the overall size of the transducer assembly. Researchers describe a variety of interconnection schemes for 2-D arrays including hand-wiring [2], multilayer ceramics [10], and printed circuit technologies [11].

Multilayer flexible (MLF) circuits, constructed using standard electronics methods, have emerged as a promising solution to these engineering challenges. MLF interconnect is desirable for its acoustic properties and also because it is relatively inexpensive and suitable for volume manufacturing. Furthermore, as illustrated in our devices, flexible interconnect may be folded to facilitate connections in confined volumes. Single-layer flexible circuits have been routinely used for interconnect to 1-D arrays [12]. The interconnection of 2-D array elements using a proprietary polyimide-based electrical circuit was proposed in 1990 by L. S. Smith *et al.* [13]. In 1995, Davidsen and S. W. Smith first reported a two-layer, thin film, flex circuit interconnect [14]. This report included a functional 2-D array transducer with 120 active elements on a 400-μm pitch and a 2.5-MHz center frequency. Also in 1995, Tournois *et al.* [3] reported a 512-element, 1.5-D array based on a single-layer flexible circuit. Davidsen and Smith reported in 1998 a 256-element sparse array on 400-μm pitch constructed with a laminated multilayer flexible circuit [15].

Despite the benefits of MLF circuits, difficulties persist as higher density interconnections are required. Interelement spacing decreases as operational frequency increases, and arrays of hundreds of elements are desirable. In such cases, interconnect density may exceed that of standard constructions in the electronics industry.

Furthermore, the established method of isolating individual elements from a plate of piezoelectric ceramic is to separate the elements electrically and acoustically with a saw kerf. This method calls for a relatively thick material beneath the transducers, but this material must be pierced with an array of vias for electrical contact to each element. Thus, an additional constraint is imposed on the circuit geometry.

In this paper, we present a new method for building high density flexible interconnect for 2-D ultrasound arrays. In addition, we discuss the implementation of this type of interconnect in several transducer assemblies. We describe the fabrication of the MLF circuits and assembly of the transducer arrays, and we present test data along with images obtained from these devices.

Our method differs from previous efforts in that a high density of transducer elements is achieved with conventional microfabrication processes. The complete circuit is built on one surface of the flexible substrate. We are able to take advantage of the precision and small feature sizes that photolithography makes possible. Multiple MLF circuits can be fabricated in parallel and tested before assembly. Furthermore, features such as the ground plane are readily constructed with thin film techniques. Other reported methods generally supplement microcircuit techniques with mechanical alignment or drilled vias [13], [15]–[18].

Our circuits are fabricated using a combination of thin and thick film methods. The circuit is built on a sheet or cast film of polyimide. As shown in Fig. 1, two layers each of sputtered thin-film Au and spin-coated polyimide form signal and ground layers, and photo-defined, stepped vias route connections between layers. A laminated and photo-defined solder mask material is applied to form a thick backing to support the PZT elements. The uppermost layer consists of electroplated Cu solder pads and contact pads at the transducer elements. With this scheme, we achieve, with good yield, minimum dimensions of 12-μm lines and spaces at the signal layer, 40-μm square vias, and array spacing on a 150-μm grid.

We have constructed five 2-D array transducers incorporating the MLF circuits: 1) a 10×13, 5.0-MHz, 3.1-mm wide probe with 64 channels, which fits inside a 12-French cardiac catheter; 2) a 10×10, 7.0-MHz, 2.0-mm wide probe with 70 channels, which fits inside a 9-French cardiac catheter; 3) a 40×40, 3.5-MHz transducer with 438 channels in a 12-mm diameter aperture; 4) a 40×40, 5.0-MHz transducer with 438 channels in a 12-mm diameter aperture; and 5) a 40×40, 7.0-MHz transducer with 438 channels in a 12-mm diameter aperture.

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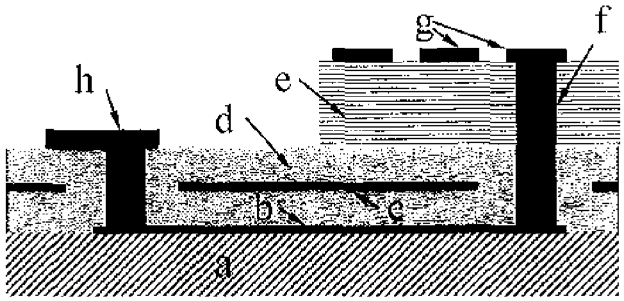


Fig. 1. Simplified cross section of circuit: a) substrate, b) trace layer, c) ground layer, d) dielectric, e) dry film dielectric, f) via, g) element pads, and h) solder pad for coax cable.

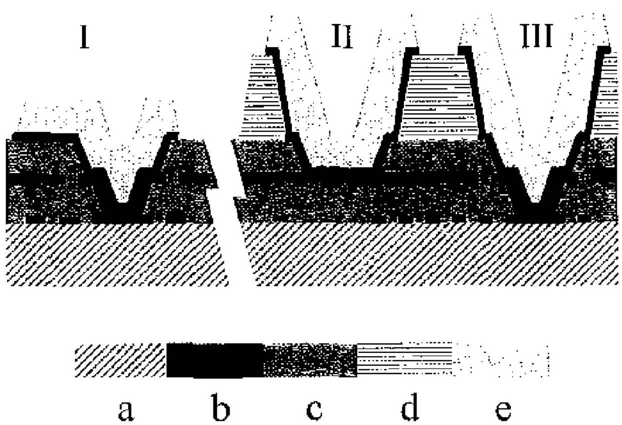


Fig. 2. Detailed schematic cross section. Materials are a) polyimide substrate, b) Cr/Au sputtered metal layer, c) spin-coated polyimide dielectric, d) dry film dielectric, and e) electroplated Cu pad. Typical features are I) solder pad, II) grounded (unused) element pad, and III) active element pad.

II. Methods

The detailed structure of the flex circuit is illustrated in Fig. 2. The circuit contains three metal layers and three dielectric layers in addition to the substrate. Each circuit consists of an array of pads corresponding to the elements of the transducer array, an array of larger solder pads compatible with the wiring at the transducer handle, and a buried network of lateral traces connecting the two sets of pads. Thin dielectric films separate and insulate the metal layers. Vias through the dielectric with metallized sidewalls provide contact between the pads and the traces. Interposed between the top layer of pads and the trace layer is a ground layer with a mesh design. Vias connect this ground layer to unused elements and ground bars on the top level. The pads at the top level are Cu-plated so that solder connections may be made. Directly beneath the element pads, the interlayer dielectric is augmented with a thicker material. This thickness provides sufficient tolerance to allow dicing through the PZT material without damaging the interconnect circuitry below.

The MLF circuits were fabricated in the Microelectronics Laboratory of North Carolina State University (Raleigh) using conventional tools for cleaning, photolithography, metal deposition, and metal etching with the addition of a vacuum laminator of the type used in printed circuit board fabrication.

Construction of the MLF interconnect proceeds as illustrated in Fig. 3. We prepare polyimide substrates, which are either unsupported DuPont Kapton VN sheets or spin-cast cured films of DuPont Pyralin. The cast films are processed on thermally oxidized Si wafers and may be released at the completion of circuit fabrication by immersion in an HF solution. The cast films on rigid backing are advantageous because they have a smooth surface, have greater dimensional stability, and can be handled with standard Si wafer fixtures.

The substrates are DC sputtered in a Denton Vacuum DV-602 system (Moorestown, NJ) with a seed layer of Cr followed by 0.5 μm of Au. The first metal level is patterned with photoresist and chemically etched to form the signal traces. After photoresist removal and cleaning, the wafers are spin coated with DuPont Pyralin PI2700 photosensitive polyimide. Exposure through the second photomask and developing of the polyimide produces tapered openings to the Au capture pads below (Fig. 3.1). Precision alignment of successive layers is accomplished with a Karl-Suss MA-6 alignment and exposure system. The polyimide coating is then cured.

We repeat these steps to deposit and pattern the second metal layer (the ground plane) and second polyimide dielectric. The openings in the second polyimide layer are made slightly larger than those in the first so that a stepped via profile is obtained (Fig. 3.2). With each subsequent metallization the via sidewalls are metallized and the metal at the capture pad accumulates.

Once the first four layers are complete, the circuits are laminated with DuPont Pyralux PC1000 photoimageable coverlay, which is a flexible acrylic, urethane, and imide composite. Originally developed as a soldermask, Pyralux PC has found other uses as a structural material in microfabrication [19]. This material meets the needs of our single-sided circuit design for several reasons. The coverlay is flexible, is photosensitive (and therefore can be patterned and aligned with high precision), can be metallized to serve as an interlayer dielectric, and is thick enough to support the PZT for dicing. A drawback is that its acoustic properties are not well characterized. However, insofar as it is a cured organic polymer, for this initial evaluation, we assume that it behaves approximately like polyimide.

The coverlay film is applied under heat and pressure with a DuPont DVL-20 drawer vacuum laminator. Exposure through the fifth-level photomask and subsequent development defines openings to the vias and also defines the boundary of the coverlay, which extends across the area beneath the transducer array but not in the region of the solder pads. The coverlay is oven-cured (Fig. 3.3).

A third metal film of Cr and Au is sputtered as a base layer for the Cu contact pads. The procedure for this level is sometimes called semi-additive patterning. Semi-additive patterning is useful for electroplating features in parallel when they are not electrically connected in the finished circuit. A laminated dry film photoresist, DuPont Riston, is applied over the Au and exposed through the sixth and final photomask. Development of the Riston produces openings in the photoresist at the vias and also at the areas for the solder pads (Fig. 3.4). The open areas may then be electroplated with Cu, with the underlying Au film supplying electrical contact to each of the isolated pads (Fig. 3.5).

The photoresist is stripped, and the Au and Cr base metal etched, leaving the Cu pads intact (Fig. 3.6). The principle

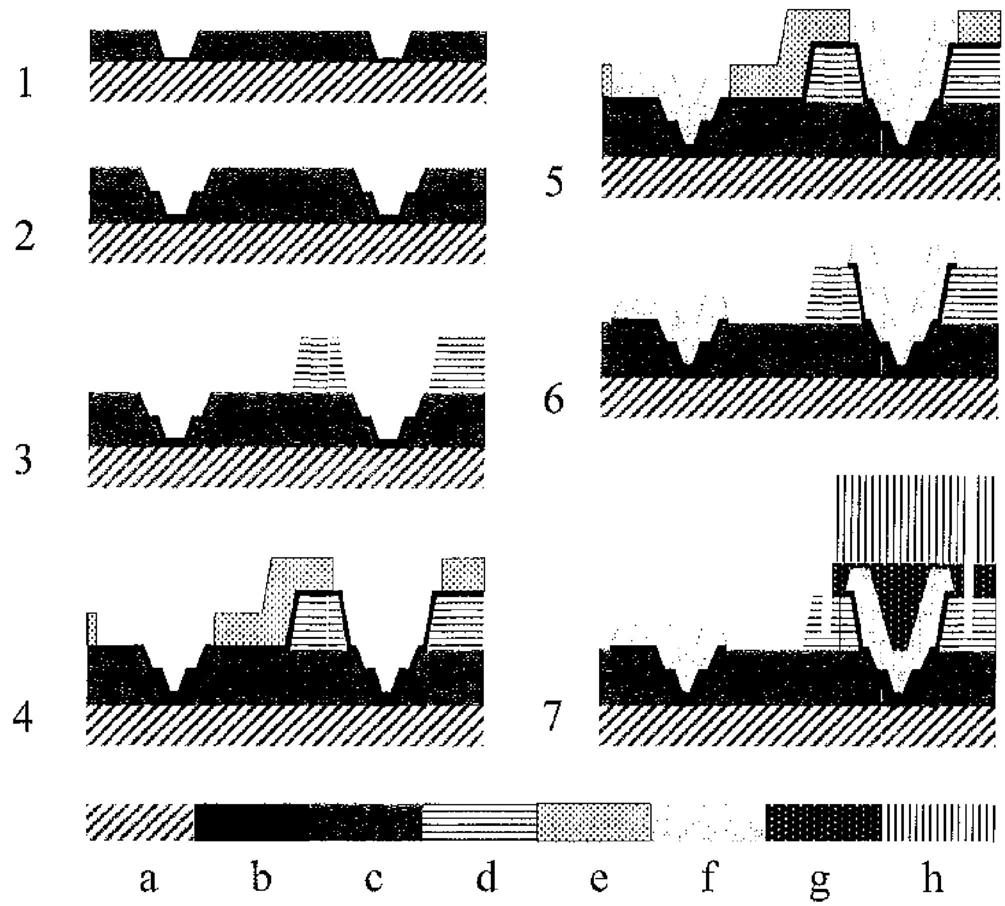


Fig. 3. Schematic fabrication sequence. Materials are a) polyimide substrate, b) Cr/Au sputtered metal layer, c) spin-coated polyimide dielectric, d) dry film dielectric, e) dry film photoresist, f) electroplated Cu pad, g) conductive Ag epoxy, and h) PZT and matching layers.

reason for a thick Cu metallization is to form an acceptable finish on the solder pads. An incidental benefit of this process is that the interior of each via is Cu-plated, helping to ensure reliable continuity among the multiple metal layers.

The final thickness of the circuit in the region of the transducers is summarized in Table I. Although the total flex thickness approaches one wavelength, at the transducer operating frequencies, the effect on acoustic propagation is minimal because there is a good acoustic match between the polymer flex circuit and the epoxy backing [15]. The metal layer thicknesses are less than 0.05 wavelength and can be ignored.

The completed circuits are cut out and prepared for application of the piezoelectric ceramic. We have described in detail elsewhere the procedure of bonding and dicing PZT, matching layer, and top ground plane [15]. In brief, a plate of PZT and matching layer, large enough to cover the array area, is bonded to the circuit with conductive silver epoxy. Saw cuts made with a dicing saw penetrate the thickness of the PZT and epoxy and extend partially into the coverlay material (Fig. 3.7).

The top ground foil is bonded to the top surface of the elements and connected to ground pads at the edge of the array. The circuit is designed such that any PZT material surrounding the active elements is grounded at both top and bottom electrodes to suppress spurious signals.

III. Results

We have used the procedure just described to fabricate 2-D array transducers of three designs. Two are narrow strips with an array at one end and rows of solder pads at the other end. The first is a 64-channel, 5.0-MHz transducer [Fig. 4a], and the second is 70-channel, 7.0-MHz transducer [Fig. 4b]. The transducers are inserted into a catheter to produce side-looking volumetric scans. Alternatively, the MLF circuits can be bent 90° just below the array to produce forward-looking volumetric scans. The third design is a 438-channel array [Fig. 4c]. The circuit extends laterally from the array region in two tabs, which fold behind the array for solder connections. We have used this third design to fabricate 3.5-, 5.0-, and 7.0-MHz transducers.

The dimensions of the catheter MLF circuits are given in Table II. We have described the array configuration previously [5]. The solder pads are arranged to accommodate coaxial cables manufactured by Precision Interconnect (Portland, OR). The cable sheathing is soldered to ground bars located between the active pads. The coaxial cables are bundled and soldered to cards for edge connection to the scanning electronics.

The 438-channel MLF transducers were fabricated with dimensions given in Table II. The signal layer is routed with a maximum density of six traces between via capture pads. The solder pads are positioned in seven double rows of 36 pads. The pad pitch matches specially fabricated double-sided flex

TABLE I
THICKNESS DIMENSIONS OF THE MULTILAYER FLEX CIRCUITS.

	3.5-, 5.0-, 7.0-MHz Transthoracic	5.0-MHz Intracardiac	7.0-MHz Intracardiac
	Nominal Thickness (μm)		
Polyimide substrate	75	75	12
Cr/Au traces	0.5	0.5	0.5
Polyimide dielectric	5	5	2
Cr/Au ground plane	0.2	0.2	0.2
Polyimide dielectric	5	5	2
PC 1000 dielectric	60	60	25
Cr/Au/Cu contact pads	5	5	5
Total thickness	151	151	47

TABLE II
CRITICAL DIMENSIONS OF THE THREE TYPES OF MLF CIRCUITS FABRICATED.

	3.5-, 5.0-, 7.0-MHz Transthoracic	5.0-MHz Intracardiac	7.0-MHz Intracardiac
Interelement spacing, mm	0.350	0.200	0.150
Aperture, mm	12	2.5	1.5
Array layout	40 × 40	11 × 13	10 × 10
Channel count	438	64	70
MLF length × width, mm	85.2 × 24.0	32.0 × 3.1	49.5 × 2.1
Minimum trace/space width, mm	0.028	0.022	0.012
Solder pad pitch, mm	0.635	0.380	0.380
Cable wire gauge	46	46	48
Catheter size (French)	N/A	12	9

TABLE III
PERFORMANCE MEASUREMENTS OF THE MLF TRANSDUCERS.

	Transthoracic			Intracardiac	
	3.5-MHz	5.0-MHz	7.0-MHz	5.0-MHz	7.0-MHz
Center frequency, MHz	3.5	4.5	6.7	5.2	6.7
-6 dB Fractional bandwidth, %	46	67	27	50	27
50-Ω insertion loss, dB	-68.4	-60.4	-72	-64	-84[1]
Cross talk, dB	-28	-31	-30	-26	-16
-6 dB angular response, °	54	46	Not available	36	Not available

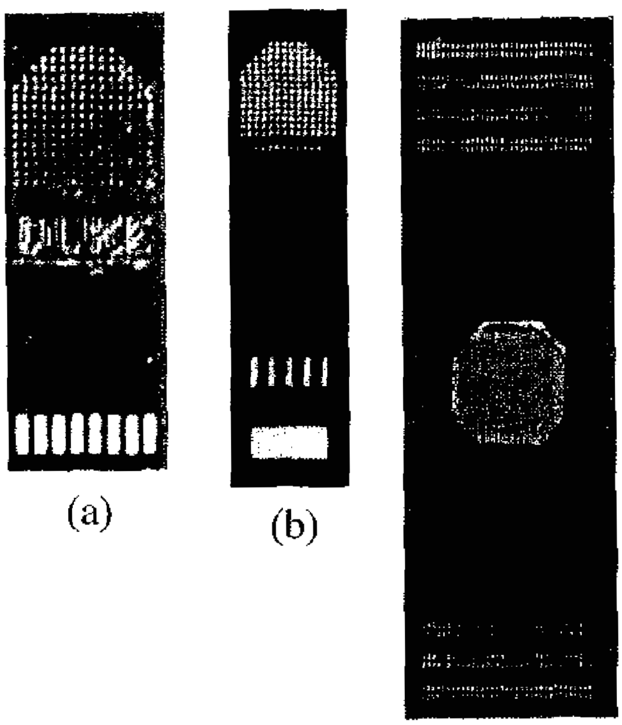
[1] Includes 1 m of cable.

circuit cards, which are soldered at right angles to the MLF. The free ends of these cards in turn fit into seven, 72-channel edge connectors (Precision Interconnect PAC connectors). The completed MLF circuit array is fitted into a compact housing by wrapping the outer tabs of the circuit around and behind the array on a backing of filled, lossy epoxy, which has been cast in the shape of a truncated square pyramid.

Electrical and acoustic measurements were made with an apparatus described previously [5]. The results from the transducer tests are summarized in Table III. The −6 dB fractional bandwidths vary from 27 to 67%. The range of insertion losses is from −60.4 to −84 dB, the angular response varies from 36° to 54°, and the cross talk is between −16 and −31 dB.

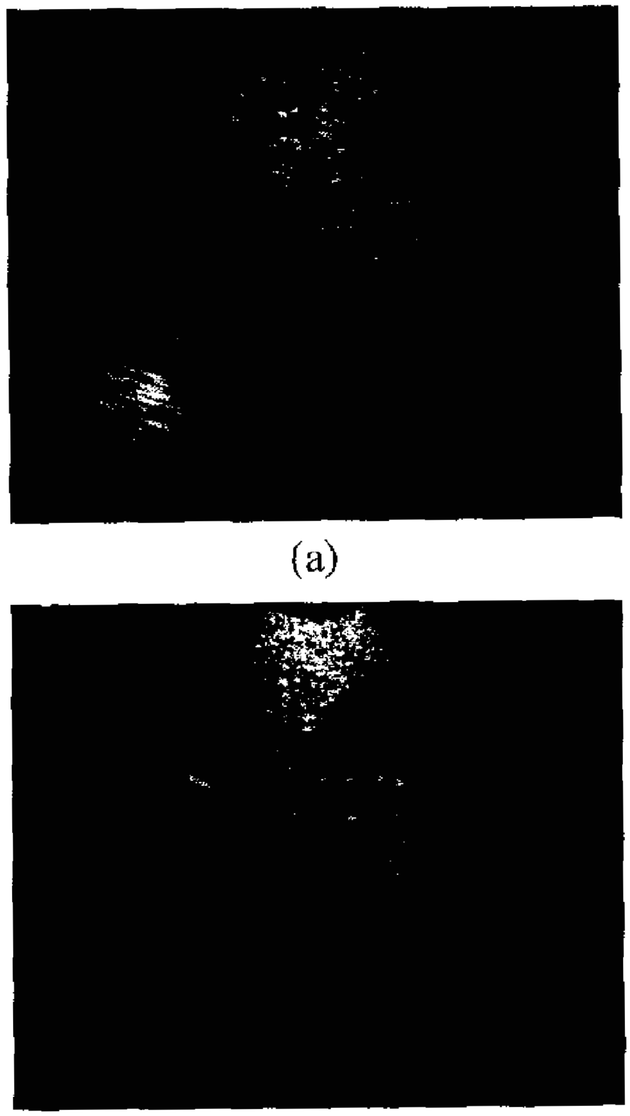
We have evaluated the imaging performance of the MLF transducers using the Duke Volumetric scanner [20]. Fig. 5 shows a typical B-scan through a phantom that includes a fluid-filled spherical cyst at the top and a solid lesion near the bottom. Each target is 20 mm in diameter, and the scan depth for each image is 12 cm. The phantom has approximately 0.5 dB/cm-MHz attenuation. Receive mode gain was adjusted to optimize each of the images at the time of scanning. Fig. 6 shows the corresponding real time C-scan through the near cyst in Fig. 5. Fig. 5(a) and 6(a) were taken with the 3.5-MHz array, and Fig. 5(b) and 6(b) were taken with the 5.0-MHz array. All of the images show good contrast resolution with little cyst fill in. The 5.0-MHz pictures show a reduced speckle size compared with the images from the 3.5-MHz transducer, but reduced penetration in the phantom. This is expected because of the frequency dependent attenuation.

Fig. 7 and 8 show typical images obtained with the 2-D array catheter transducers. Fig. 7 is made with the 5.0-MHz catheter and shows the pulmonary veins of an excised left atrium from



(a) (b) (c)

Fig. 4. Detail of MLF showing array and one row of solder pads for a) 5.0-MHz intracardiac catheter and b) 7.0-MHz intracardiac catheter. c) MLF for 40 × 40 transthoracic arrays.



(a)

(b)

Fig. 5. B-mode phantom images using a) the 3.5-MHz transthoracic array and b) the 5.0-MHz transthoracic array.

a sheep. We can clearly see the long axis of the veins in the B-scan, and the cross section of the orifice in the corresponding real time C-scan.

Fig. 8 shows an image of a wire phantom in a water tank made with the 7.0-MHz catheter transducer. We show a B-scan and two real time C-scans. The wires are spaced from 0 to 9 mm apart in the lateral dimension and 2 mm apart axially. We can start resolving the wires laterally at 4-mm separation at a depth of 2 cm. We see these 4-mm separated wires in the bottom real time C-scan on the right. The C-scan on the top is through the wires that are 6 mm apart. We can resolve the 2-mm axial separation of the wires throughout the phantom.

IV. Discussion

We have fabricated several 2-D array transducers operating from 3.5 to 7.0 MHz on multilayer polyimide circuits. The MLF interconnects offer some advantages over previous fabrication methods. The MLF processes lend themselves well to mass production methods. Once the MLF is prepared, it takes 3 d to build a transducer versus 3 wk with our hand-wired technique. In addition, because of the fine trace sizes and spaces, to our knowledge, these arrays have the smallest interelement spacing (0.15 mm) developed for a 2-D array. The major drawback to the MLF is the small thickness of the substrate. We would like to make a deep kerf to ensure good acoustic isolation, but we cannot cut deeper than our coverlay material without damaging the electrical traces. This leads to increased acoustic cross talk when compared with other fabrication methods [6], [9].

To our knowledge, the 7.0-MHz transducers are the highest frequency 2-D arrays yet fabricated. Because these are the first prototypes, no matching layers were used. As expected, this causes lower bandwidth and increased insertion loss over the other transducers. Table III shows that the MLF transducers have performance comparable with our previously published fabrication techniques. Fig. 5 and Fig. 6 show improved images for the 3.5- and 5.0-MHz MLF transducers over their hand-wired predecessors of the same frequency [8]. Other researchers have published measurement results for 2-D arrays operating between 2.3 and 3.2 MHz. Bandwidths ranging from 50 to 60%, single-element angular response from 36° to 54° and cross talk around −40 dB were reported [6], [9].

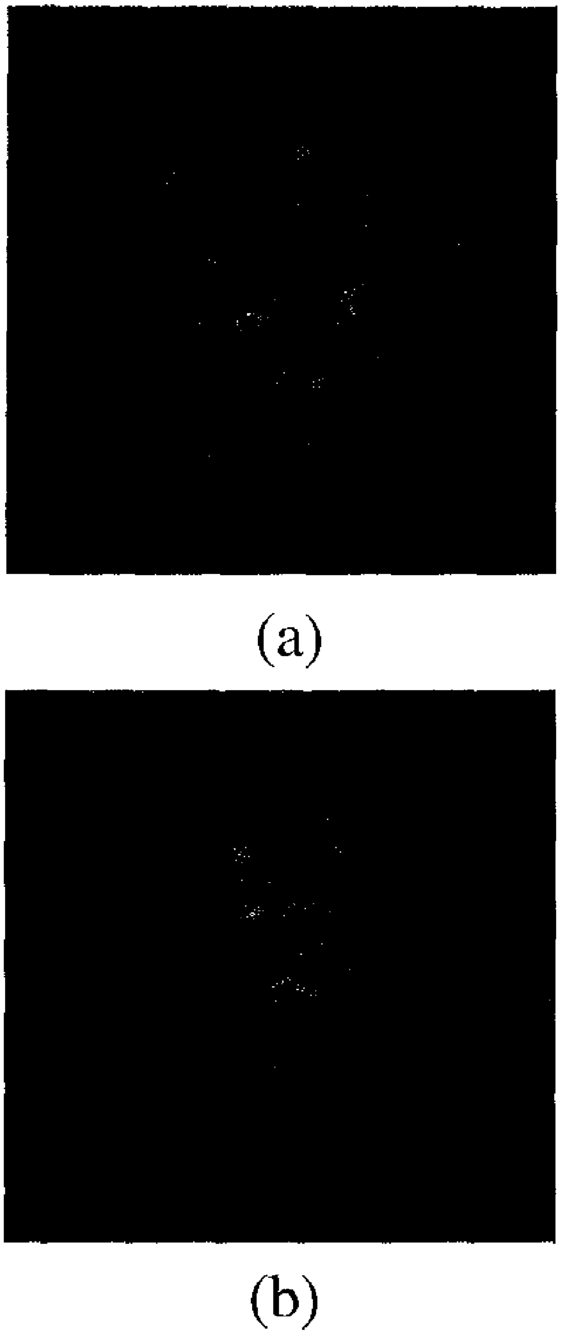


Fig. 6. Real time C-scans through the cyst in Fig. 5 using a) the 3.5-MHz transthoracic array and b) the 5.0-MHz transthoracic array.

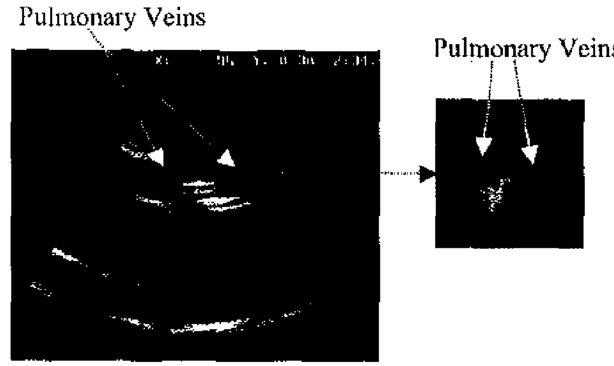


Fig. 7. B-mode and corresponding real time C-scan through an excised left ventricle of a sheep using the 5.0-MHz cardiac catheter array.

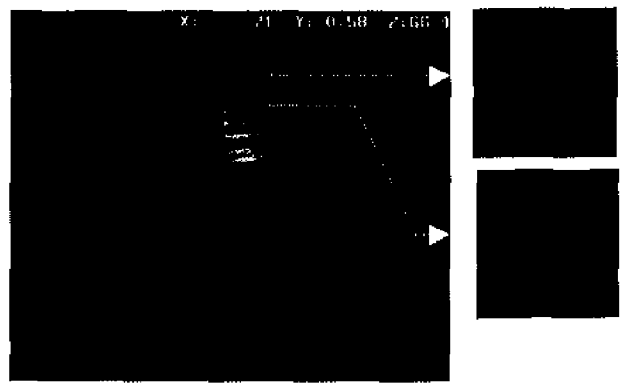


Fig. 8. B-mode and corresponding real time C-scan of a wire phantom using the 7.0-MHz cardiac catheter array.

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