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PRINCIPAL INVESTIGATOR: Mr. Timothy White

CONTRACTING ORGANIZATION: Lockheed Martin Infrared Imaging Systems
Lexington, Massachusetts 02421

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Mr. Timothy White

7. PERFORMING ORGANIZATION NAME(S) AND ADDRESS(ES)
Lockheed Martin Infrared Imaging Systems
Lexington, Massachusetts 02421

E-MAIL:
tim.white@lmco.com

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13. ABSTRACT *(Maximum 200 Words)*

This final report describes the accomplishments of a two phase, three year technology demonstration program whose goal was "to develop a 2-D acoustic array which incorporates VLSI technology and demonstrate it's 3-D imaging capability in a portable ultrasound imager".

During Phase I (year 1), Lockheed Martin demonstrated ultrasound imaging at 5 MHz using an electronically scanned, two-dimensional hybrid transducer array and an acoustical lens. The hybrid array was constructed by mating a readout integrated circuit (ROIC) originally designed for infrared imaging to a 42x64 element piezoelectric array.

In Phase II (years 2 & 3), a camera system suitable for medical imaging was designed and key components were demonstrated. This system included: a 128x128 element piezocomposite array mated to new ROICs specifically designed for medical imaging; supporting electronics and signal processing; and a high performance acoustical lens.

In-vitro real-time images were made of test objects and biological specimens. Demonstration of a portable imager was not realized.

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PI - Signature Date

THE BATTLEFIELD ULTRASONIC DIAGNOSTIC IMAGER (BUDI)

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Real-Time 3-D Ultrasonic Diagnostic Imager for Battlefield Application

Introduction

The goal of the Real-Time 3-D Ultrasonic Diagnostic Imager for Battlefield Application (BUDI) program was *"to develop a 2-D acoustic array which incorporates VLSI technology and demonstrate its 3-D imaging capability in a portable ultrasound imager"*.[1] The camera is the acoustical analog of a conventional video camera. It uses a large area, high pixel density, fully populated, two-dimensional (2D) transducer array to act as the sensing element in the focal plane of an acoustical lens. The third dimension is achieved by range gating the received ultrasound signals.

Background

The ultimate goal of the Real-Time 3-D Ultrasonic Diagnostic Imager for Battlefield Application (BUDI) program is the development of a manportable ultrasound imaging capability for rapid evaluation of battlefield injuries, particularly abdominal trauma. By moving a capability to monitor internal bleeding to forward echelons, triage decisions may be optimized in the "golden hour" so vital for survival. Compared to conventional B-scan ultrasound systems which only image a single plane at a time and require mechanical (typically motorized) scanning for 3D coverage, the ability to simply "point and shoot" with our acoustical camera will make imaging more reliable under the often chaotic conditions of the battlefield. This technology bridges the large gap between taking vital signs and making basic observations in the field and taking CT scans in a hospital.

Transmitting these images to a radiologist in a field hospital will allow the initial diagnosis to be made in a more benign environment. In many cases, additional imaging may not be needed when the individual reaches the field hospital and they may proceed rapidly to the next phase of therapy.

Our new technology enables three-dimensional imaging of radiolucent shrapnel, fluid collections and abdominal organs with:

- Excellent image quality
- Real-time frame rates
- Image manipulation to obtain any derived view of cross-section
- Image transmission for remote diagnosis and archiving
- Portability
- Low cost

This is true for both battlefield environments and civilian EMT/ambulance applications. Health care in remote locations where specialists and expensive imaging diagnostic equipment are not available can also benefit.

In 1999, the importance of 3-D imaging is well understood by the medical community and the ultrasound industry is in search of a technical solution to building high resolution, 2-D ultrasound arrays. Current B-scan ultrasound imaging limits the visualization of 3-D structures because only 2-D slices are imaged. These slices must then be reconstructed in

[1] Lockheed Martin (Loral) Proposal 9403-21 to BAA 94-14, 31 Mar. 1994.

the operator's mind to obtain a mental 3D "picture" of the target anatomy. Automatic movement of the transducer with off line computer reconstruction has recently been introduced by some manufacturers, but the systems are slower, more costly and harder to use than conventional B-scan systems. They are not well suited to battlefield use.

The Lockheed Martin IR Imaging Systems acoustical camera is the application of dual use technology derived from other DARPA sponsored programs. Dense 2D arrays are derived from infrared focal plane array (IRFPA) technology, using acoustical transducers in place of the IR detectors. In the DARPA/MTO flexible manufacturing program Lockheed Martin IR Imaging Systems has also demonstrated fabrication of ultrasound Transducer Hybrid Assemblies (THA's). Using the same IR hybrid fabrication line that supplies DOD and commercial IR imaging products, THAs have been made available to this program without significant startup expenses, providing maximum synergy of DARPA programs.

Conventional approaches to two-dimensional array fabrication have been constrained by the technique employed to interconnect the transducer and the initial stage of electronics. These approaches rely on micro-coaxial cable soldered to each array element. Although micro-coax technology has improved dramatically in the past decade, interconnecting thousands of array elements with separate wires remains a formidable and expensive challenge. In addition to this practical fabrication issue, the capacitance of a long coaxial cable (typically > 40 pF/m) is much larger than that of a typical 2D array element (< 1 pF). This creates a voltage divider that severely reduces the signal-to-noise ratio of the channel.

Phase I – Feasibility Demonstration (Fig. 1)

During Phase I (year 1, 1995) of the program, Lockheed Martin IR Imaging Systems demonstrated ultrasound imaging at 5 MHz with an electronically scanned two-dimensional transducer hybrid array (THA) and acoustical lens[2]. A receive-only integrated circuit (ROIC), originally designed for infrared imaging was hybridized to a 42x64 element piezoelectric array. The readout electronics originally used with an IR focal plane array were also modified by adding a range gate. Real-time diffraction limited images of high reflectance targets were obtained with the f/1.0 doublet acoustical lens.

The two-dimensional transducer array used a "flip-chip" or direct interconnection approach to join the transducer array to the ROIC. This direct interconnect technique allows array elements to be individually connected *en-masse*. Furthermore, the length of the direct interconnection is less than 0.02 mm, reducing interconnection capacitance to the level where it is no longer a dominant factor in the channel signal-to-noise ratio.

As described in the Phase I report, both specular and diffuse scattering targets were successfully imaged in reflection. An image of line targets buried within a medical ultrasound test object, taken by the Phase I transducer hybrid is shown in Fig. 2. The feasibility of real-time 3D imaging was thus demonstrated, however, there were many significant limitations to this experimental apparatus. Additional information regarding Phase I is in Reference 2 and there is no further discussion herein.

[2] Real-Time 3-D Ultrasonic Diagnostic Imager for Battlefield Application, 1995 Annual Report (Phase I).

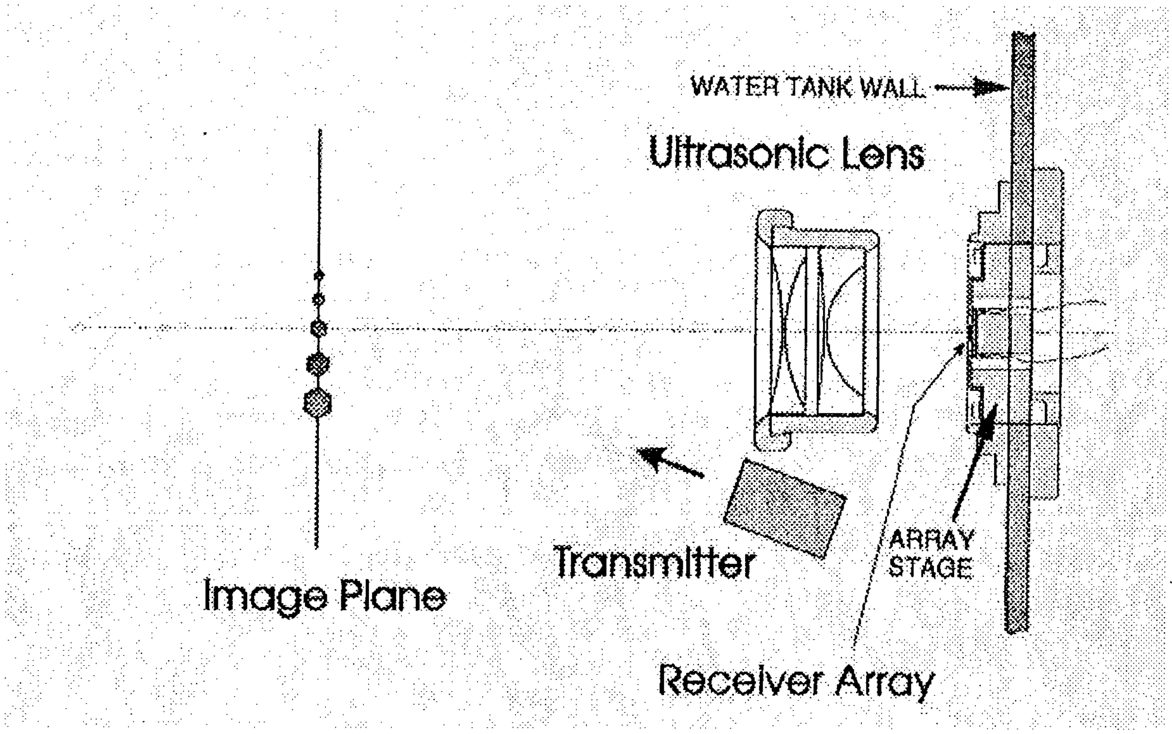

Fig. 1 Phase I Feasibility System

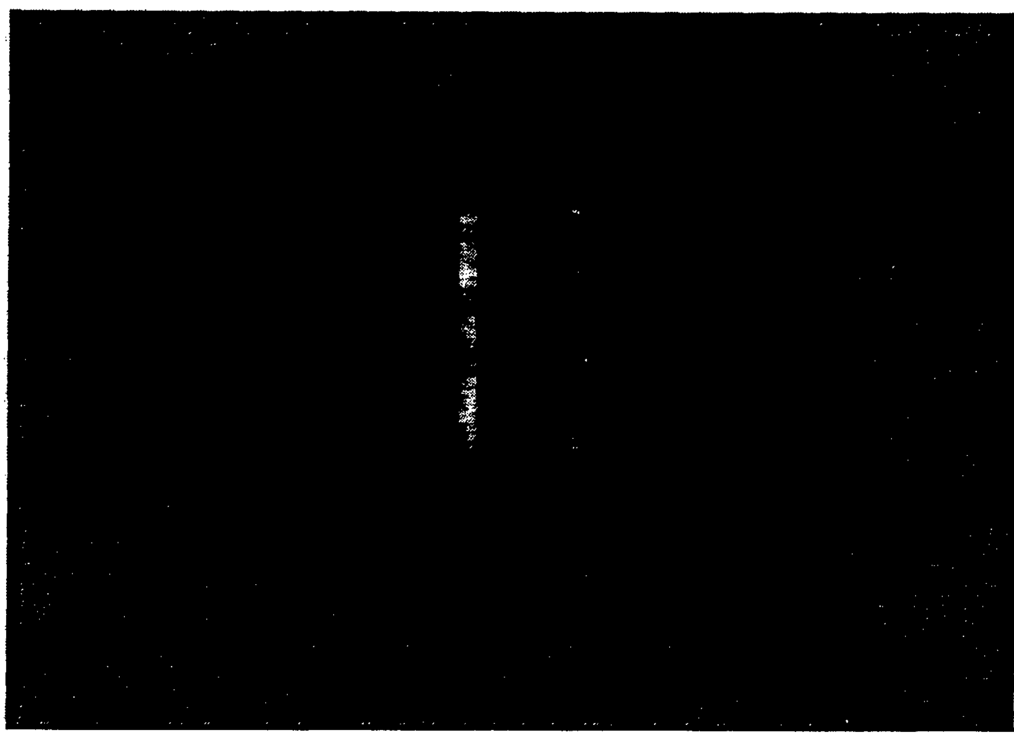

Fig. 2. Image of 1 mm Monofilament Nylon in Tissue Equivalent Test Object at a Depth of 1 cm.

PHASE II - LABORATORY DEMONSTRATION SYSTEM (LDS)

In Phase II (years 2 & 3), a camera system suitable for a laboratory demonstration of medical imaging was designed and key components were demonstrated. This system included: a 128x128 element piezocomposite array mated to new ROICs specifically designed for medical imaging; supporting electronics and signal processing; and a high performance acoustical lens.

In-vitro real-time images were made of test objects and biological specimens. Development and demonstration of a portable imager was not realized.

This final report summarizes the Phase II system design and presents test results.

System Requirements

The Battlefield Ultrasonic Diagnostic Imager Laboratory Demonstration System (BUDI-LDS) is intended for real-time 3D imaging of foreign objects, abdominal organs, and pooled fluids such as blood in the abdominal cavity with dimensions greater than 1 cm, and detecting hard foreign objects within the abdomen, e.g. plastic or metal fragments, larger than 1 mm^3. Table I summarizes target system performance requirements.

Table 1. Target System Performance Requirements

Objects to be Imaged:	Foreign Objects, Abdominal Organs and Pooled Fluids	
Frame Rate:	10 Hz - 30 Hz	
Resolution:	**BUDI**	**Present Systems**
Azimuth:	< 1 mm	> 2 mm
Elevation:	< 1 mm	> 6 mm
Range:	< 1 mm	< 0.5 mm
Operating Frequency:	5 MHz	
Range in Tissue: (Penetration Depth)	≥ 10 cm ≥ 15 cm for hard objects	
Field of View	8 x 8 cm	
Array Sensitivity:	≥ -220 dB re 1 V/μPa	
Noise Equivalent Pressure:	less than 1.0 Pa	
Dynamic Range:	≥ 90 dB total, >30 dB Instantaneous	
Contrast Resolution:	resolves 3 dB difference from background	
Fixed Pattern Noise:	less than 1 LSB @ 8 bits or -48 dB	

The BUDI-LDS was designed to acquire data for subsequent three-dimensional image reconstruction and rendering. This data can take the form of either a sequence of frames focused at succeeding depths for direct volume rendering, or an amplitude and phase image from which three-dimensional data can be holographically reconstructed and subsequently rendered.

Predicted LDS System Performance

Figure 3 is a graph of the predicted LDS system signal levels from targets in the human body as a function of target depth from the skin surface (penetration). It is based on the average value of attenuation in the body of 1 dB/cm/MHz where the distance is the depth into the body (i.e. ½ the total path length of sound propagation). At 5 MHz, the attenuation is thus 5 dB/cm. The graph is referenced to 0 dB which is the reflection from a perfect reflector at the depth.

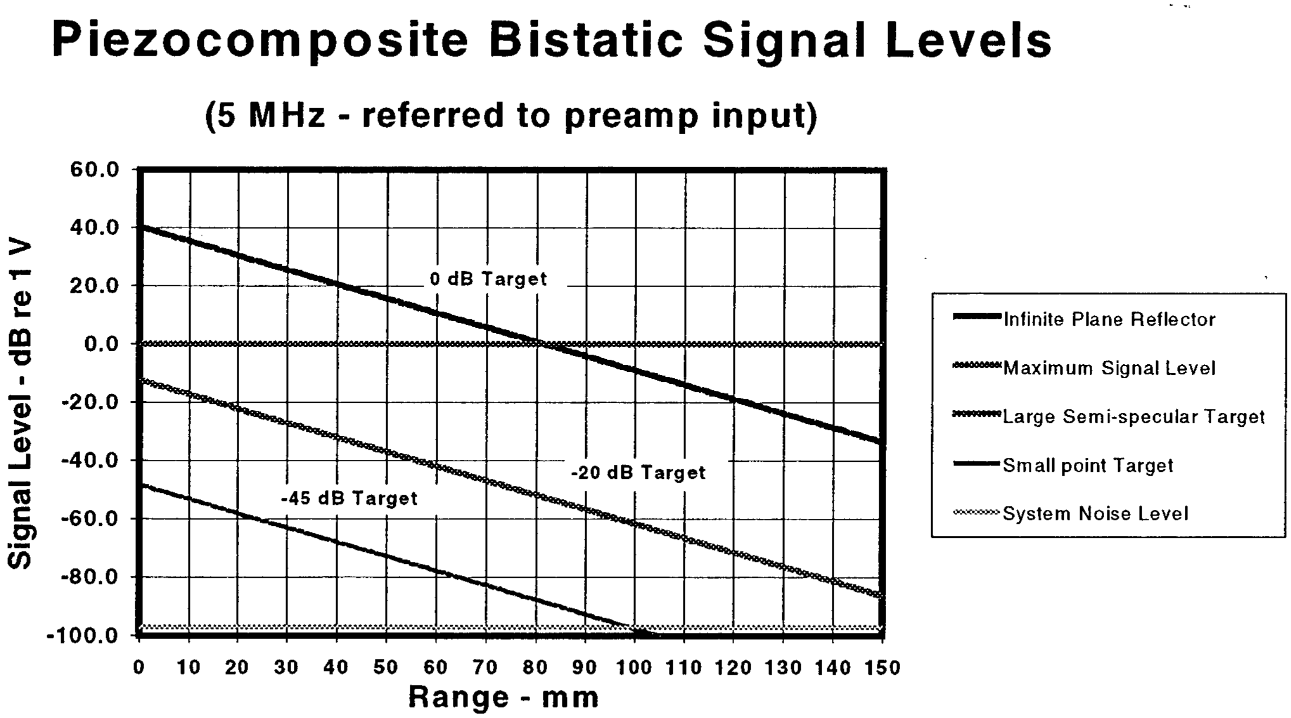

Fig. 3 Signal Levels vs. Depth in the Human Body

A perfect reflector at the skin surface would thus gives a signal level of 40 dB re 1V or 100 Volts. If this same reflector is located at a target depth of 20 mm, the signal level is reduced by 10 dB (2 cm *depth* x 5 dB/cm). At 4 cm, this value is -20 dB, etc.

Objects of interest in the abdomen are generally not perfect reflectors, but have lower reflectance. Typically, large specular reflections such as the boundaries between fat and muscle are -20 dB re a perfect reflector. Non-specular targets such as blood vessels, tumors, small foreign bodies, tissue parenchyma, blood, fluid filled cysts, etc. range in scattering power from nearly in the specular range to extremely small values depending on their size *and orientation.* In the absence of blood vessels, cartilage and connecting tissue, a large relatively uniform tissue such as liver parenchyma is generally considered to be a "volumetric" scatterer, i.e., it is composed of a statistical ensemble of unresolved scatterers.

Echo levels in the body may have a large amplitude range, for example, 60 dB. Any imaging system attempting to image inside the body at 5 MHz, thus must contend with a huge dynamic range, typically on the order of 100 dB.

There are two limits to the ability of the imaging system to detect such targets. The first is the electronic noise level as shown in Fig. 3. Clearly, with ordinary techniques, signals lower in amplitude than this noise level will not be seen. The electronics noise limit is generally the limiting factor at a certain depth (beyond 100 mm for small targets). Targets with higher scattering can be seen out to greater depths (150 mm).

Increasing the input acoustic levels can boost echo signals above the electronics noise level and make it possible to see deeper into the body. Practical limits in the transmitter, presumed safety limits and/or governmental regulatory limits do not permit acoustic input levels to the human body to be raised beyond a certain point. The acoustical power levels in the BUDI system as designed are well within current FDA Guidelines for Diagnostic Ultrasound Systems.[3]

The other limit is the target clutter/contrast ratio. This limit is determined both by the imaging system itself and by the scattering properties of the tissue. This is a topic of great contemporary interest and research in the medical ultrasound field, where it is often discussed in terms of aberration correction. Unlike the case of the electronic noise level limitation, it is not possible to improve target contrast *ratio* by increasing input power levels[4].

> "Wavefronts distort when ultrasound propagates through an inhomogeneous medium, lowering target contrast in the image and introducing image artifacts. Isotropic scattering reduces the contrast."
>
> When target strength is stronger than background scattering, the principal scattering mechanism is primary (i.e. simple, single or nonmultiple) scattering."
>
> Scattering produces a halo in the image centered on each scatter source or target. Its width is the order of 2-6 deg at 3 MHz and (*its angular extent*) is proportional to wavelength." (italicized comment added).

---

[3] "Information for Manufacturers Seeking Marketing Clearance of Diagnostic Ultrasound Systems and Transducers", U.S. Department of Health and Human Services, Food and Drug Administration, Center for Devices and Radiological Health, September 30, 1997

[4] "Scattering from a Multiple Random Phase Screen Model of a Random Inhomogeneous Medium" Bernard Steinberg, U. of Pennsylvania, Personal Communication.

"Target Contrast Ratio (TCR) within the halo is linear with 2-D array area, inversely proportional to time delay variance per unit distance and propagation length and decreases as the square of the frequency and correlation distance. The formula is

$$TCR \approx \frac{L^2}{D\omega^2 \left(\pi \cdot x_{om}\right)^{(2)} \left(\sigma_{\tau m}^2 / D_m \right)} \qquad \text{(Eq. 1)}$$

where
L = array size
D = propagation path length or tissue thickness
ω = 2πf is the angular frequency
$\pi \cdot x_{0m} \approx$ a - typical scatterer size
$\sigma_{\tau m}^2$ = time delay variance
D_m = tissue thickness in which $\sigma_{\tau m}^2$ is measured."

"The numerator parameter and first two denominator parameters are design parameters. The next two denominator parameters are measured properties of the tissue. The last is the length of the tissue sample in which tissue parameter measurements are made. Thus, time delay variance and correlation distance are the primary tissue measurements required for prediction of TCR in a purely scattering medium."

"lateral resolution was found to remain λ/L, the width of the diffraction pattern, and unaffected by the scatter field so long as the scattered energy does not obscure the main image lobe".

Thus both resolution and tissue contrast ratio are functions of the aperture L. The lens aperture of the acoustical camera is 78 mm, which is 3 to 5 times larger than any conventional B-scan system. This provides a theoretical resolution of 1.1 mm, which should provide a substantial improvement in image quality, independent of other sidelobe issues.

Laboratory Demonstration System (LDS) Top-Level Description (Fig. 4)

The Battlefield Ultrasonic Diagnostic Imager Laboratory Demonstration System (BUDI-LDS) consists of two major components: the camera and the operator's console.

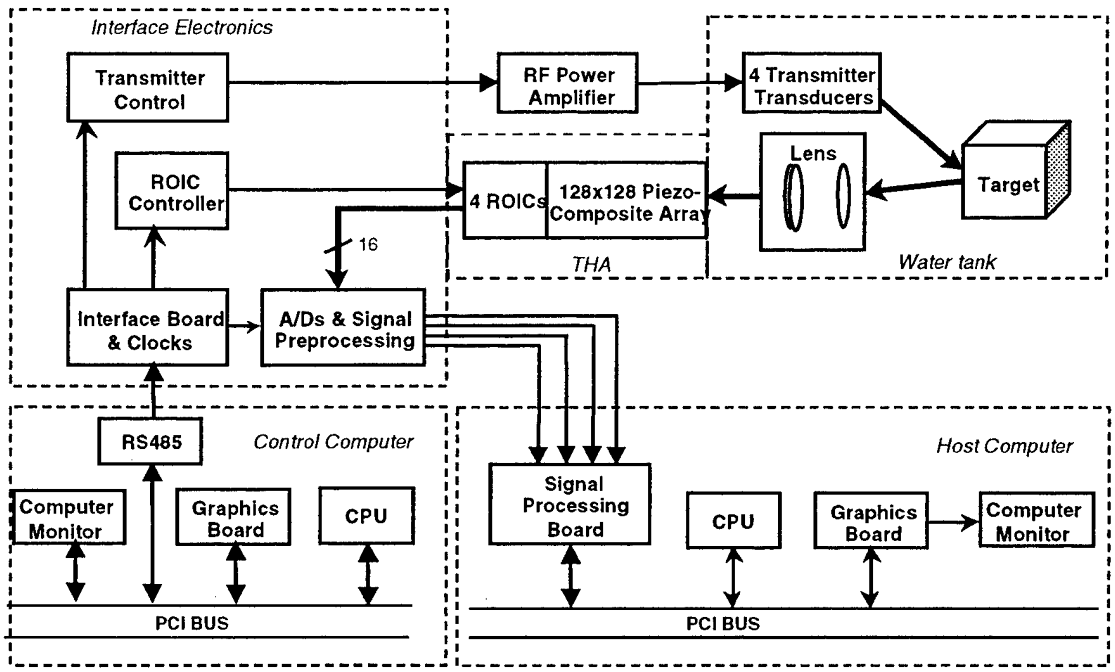

Fig. 4 BUDI-LDS Block Diagram

Acoustical pulses are generated in the camera by four 5 MHz transmit transducer. This allows the system to improve contrast by averaging the four different speckle patterns. The transmitter pulse propagates into the body through a water path and acoustically transparent membrane or in the water tank for in-vitro imaging.

The camera acoustical lens collects and images the reflected ultrasonic energy from the targets. The camera electronics consists of two subassemblies: the transducer hybrid assembly (THA) and the interface electronics (IE). The THA performs the acoustical to electronic image conversion. Located in the focal plane of the acoustic lens, it consists of a 128x128 element, two-dimensional, piezocomposite array, electrically interconnected to a custom, silicon readout-integrated-circuit (ROIC). The IE operates the THA, providing power and logic control signals and image preprocessing and then transfers the image across four high-speed serial links to the signal processing board in the host computer for further processing. The IE also controls the transmission of the acoustic energy pulse.

The host computer provides overall system control, image processing via the signal processing board, image display, the user interface and system support. Its primary elements are an IBM PC compatible microcomputer, a monochrome monitor, a RS-485 serial link for communication to the IE. An external power amplifier and power conditioning electronics were used in the LDS.

The signal processing board runs embedded firmware to process 3D data from the camera electronics into a two-dimensional image. A standard graphics board displays the image via analog video (RS-170) on a standard monitor.

The RF power amplifier supplies the required high power (up to 2500 W peak, into 50 ohms), high frequency pulses required for acoustic interrogation. It is controlled by the camera interface electronics and drives the transmit transducers.

System master control originates at the host CPU running under Windows 3.11, DOS and LabView 4.0 on a conventional color monitor. The user selects a command such as START imaging from the User Interface screen using a mouse. This initiates setup information, which is downloaded across an RS-485 serial port into the Interface Electronics (IE) in the camera.

The Interface Electronics interpret this command and begin image generation and acquisition. A 5 MHz gated sinewave signal and a synch signal are sent out to an external RF power amplifier. The high power RF pulse returns to the camera and IE, where it is distributed to one of four transmitter transducers. A transmitter transducer converts the electrical pulse into an acoustic pulse. This acoustic pulse propagates through an internal fluid path in the camera and exits into the patient through a low loss, low reflection acoustic window.

The I/E electronics provide the control signals, clocks and power for the ROICs. Each acoustic element of the array is 0.2 x 0.2 mm square and is bonded directly to a similarly sized region on the integrated circuit containing a preamplifier, analog sampling, signal processing and storage. The in-phase (I) and quadrature (Q) components of the ultrasonic image signals incident on the array from the single transmitted pulse are simultaneously sampled and stored on all 16,384 ROIC cells. Up to 5 ranges are sampled in the THA.

The sampled data stored in the four ROICs are then rastered out through 16 parallel, 10 MHz data channels into the I/E electronics, where they are digitized to 12 bits, ordered and multiplexed through four parallel high speed serial ports leading to the console board in the host CPU.

In the console board, data from the serial ports are converted back to 12 bit words, further reordered and magnitude and phase are computed from the sampled I and Q components of the image.

These data are then sent over hardwired ports to an image processor DSP board in the PC, where gain correction, median filtering, bilinear interpolation to 256x256, gray scale manipulation, and finally RS-170 formatting occurs. At this point, the video image is sent to a black and white monitor for operator viewing.

This completes the process for a single pulse of sound. Depending on the imaging configuration selected by the operator, several additional operations may be performed.

The maximum transmitted pulse rate is determined by the time it takes the pulse to travel from the transmitter, pass through the acoustic window, enter and leave the body and propagate back through the acoustic lens to the THA. The velocity of sound in the various media and the path lengths traveled determines this time. For a 15 cm depth in the body, a maximum acoustical pulse rate of 240 Hz rate was achieved. Since a video monitor updates its information at 30 Hz, this means that up to 8 acoustic frames can be averaged to improve image quality. For this averaging to be effective, any noise in the images must be statistically independent. One of the major image noise sources expected is ultrasonic speckle arising from the coherent nature of the ultrasound and the properties of the human body. To create images with statistically independent speckle noise, four spatially separated transmitting transducers, driven at multiple frequencies are employed. These acoustic frames may then be averaged in the console board.

LDS Components

The following sections describe the design of components of the camera and console in detail. Discussed are design considerations, requirements and implementation and performance, where completed.

Camera Housing (Fig. 5)

The camera housing provides the mechanical support for the various subsystems including the transmitter, the water-filled acoustic lens, the patient/camera interface, the acoustic imaging module (AIM) and the depth selection mechanism. Figure 5 also shows the camera's depth and focus drives. The focus drive is used to allow optimal adjustment of the fixed-focus lens, while the depth drive allows the selection of the desired object plane. An external pump fills the additional fluid volume to provide a continuous image path. The interface between the acoustic lens and the THA is designed to provide both optimum acoustic matching and a robust water seal. It should be noted that the camera housing and drive mechanisms were not fabricated.

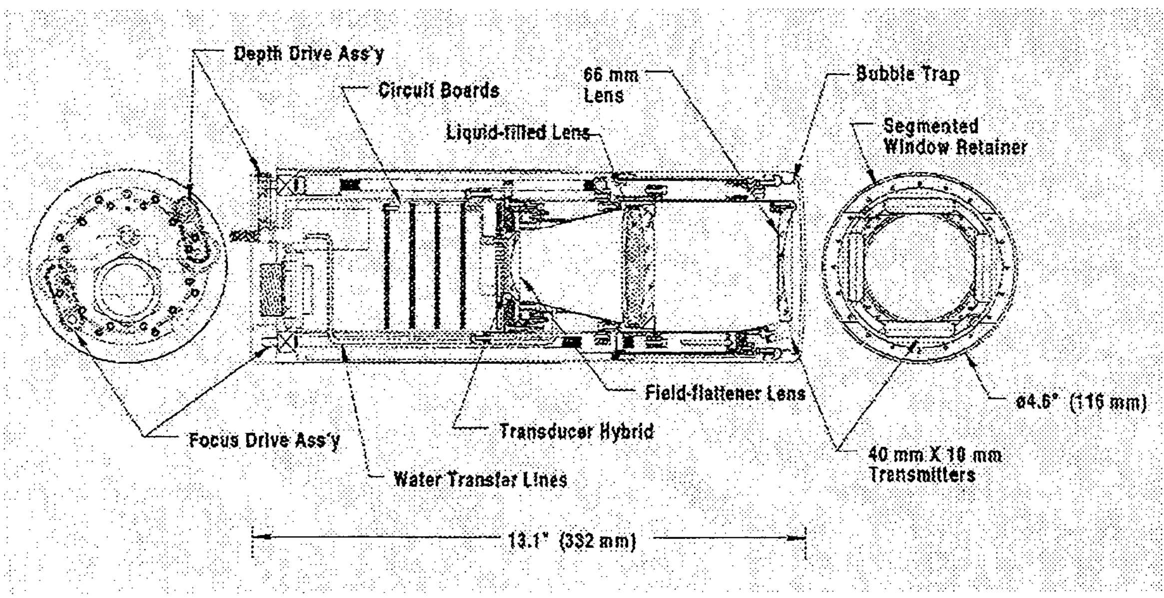

Fig. 5 Camera Housing (as designed)

Figure 6 depicts the camera at the extremes of the image depth range. Since the BUDI-LDS employs a fixed focused lens for simplicity, the camera shroud extends to image shallow depths. The AIM is composed of the Transducer Hybrid Assembly and Interface Electronics boards.

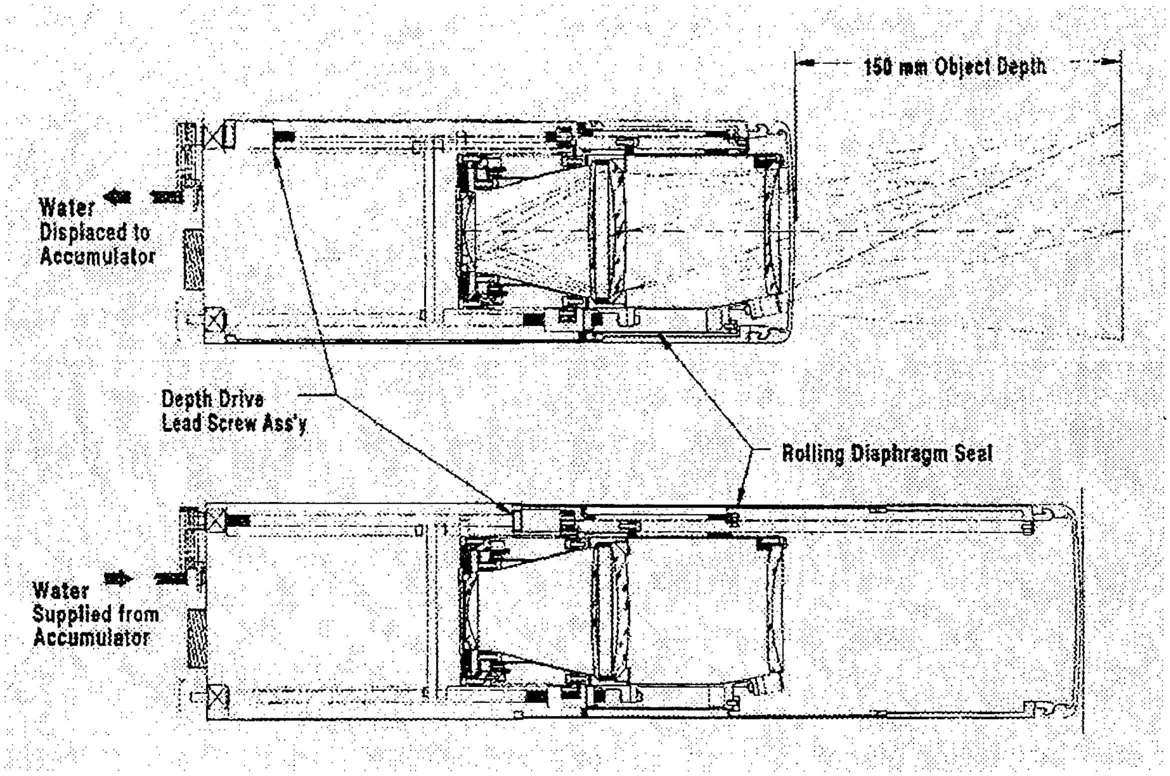

Fig. 6 Camera at the two extremes of the Image Depth Range

Transmitter Subsystem (See Fig. 4)

The interface electronics (IE) controls transmitter operation during imaging, setting the transmitted frequency, synchronizing the output pulse with the operation of the THA and selecting the transducer that will transmit during a given frame.

A pulse is initiated when the IE generates a gated, sinusoidal voltage pulse at the frequency of interest. The drive circuit receives this pulse, controls its amplitude and passes it to the RF amplifier, where it receives a fixed gain. This increases the total power of the pulse to a maximum of 1000 W peak. The high voltage pulse returns to the IE, where a high speed relay bank routes it to the selected transmitter transducer. The transducer is designed to fill the entire 80 mm x 80 mm field of view of the acoustic lens.

Acoustical Lens (Figs. 7 & 8)

The four-element wide-angle lens was designed using standard optical lens design software in routine use at Lockheed Martin. A ray trace and predicted performance parameters of the uncoated f/1.3, 76 mm diameter polystyrene lens is shown in Fig. 7.

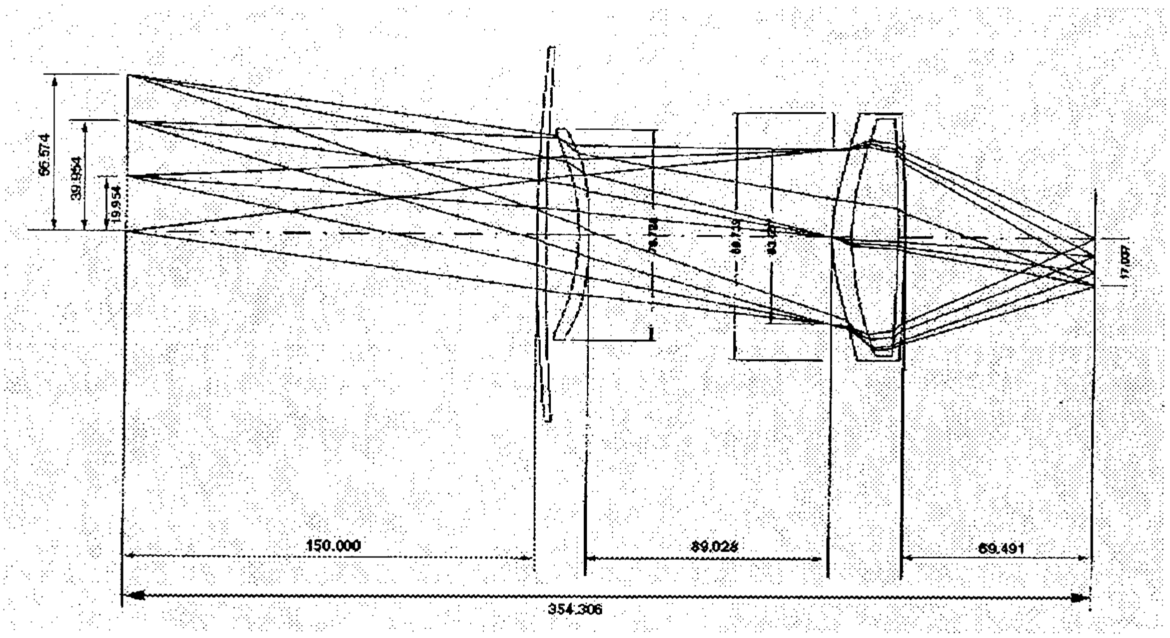

Fig. 7 Acoustical Lens Ray Trace

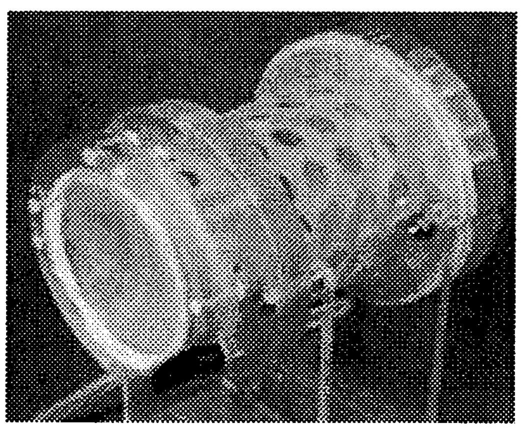

Fig. 8 Acoustical Lens Photograph

Transducer Hybrid Assembly (THA) (Fig. 9)

The Transducer Hybrid Assembly is the receiver transducer for the BUDI-LDS. It consists of a 128 x 128 (16,384) element, two-dimensional array of piezocomposite transducers electrically integrated to a 128 x 128 element array of amplification and processing electronics. Figure 9 is a photograph of the THA. The piezoelectric array is attached to four 64 x 64 element readout integrated circuits using an indium bump bonding technique developed for infrared focal plane assembly. The resulting package is 1.8" square and provides a 25.6 mm square active area.

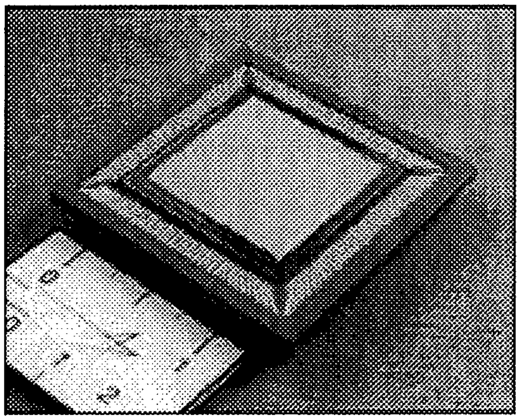

Fig. 9 Photograph of the Transducer Hybrid Assembly (THA)

Figure 10 shows a cross sectional schematic of the THA, while Fig. 11 shows a plan view schematic . Note that the array active surface rises above the package top surface to mate with the membrane that forms the acoustic interface with the lens.

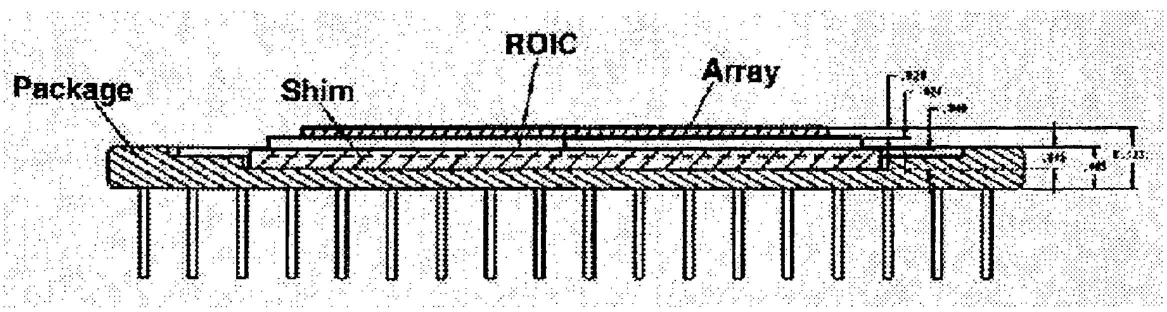

Fig. 10 Cross-section of the Transducer Hybrid Assembly

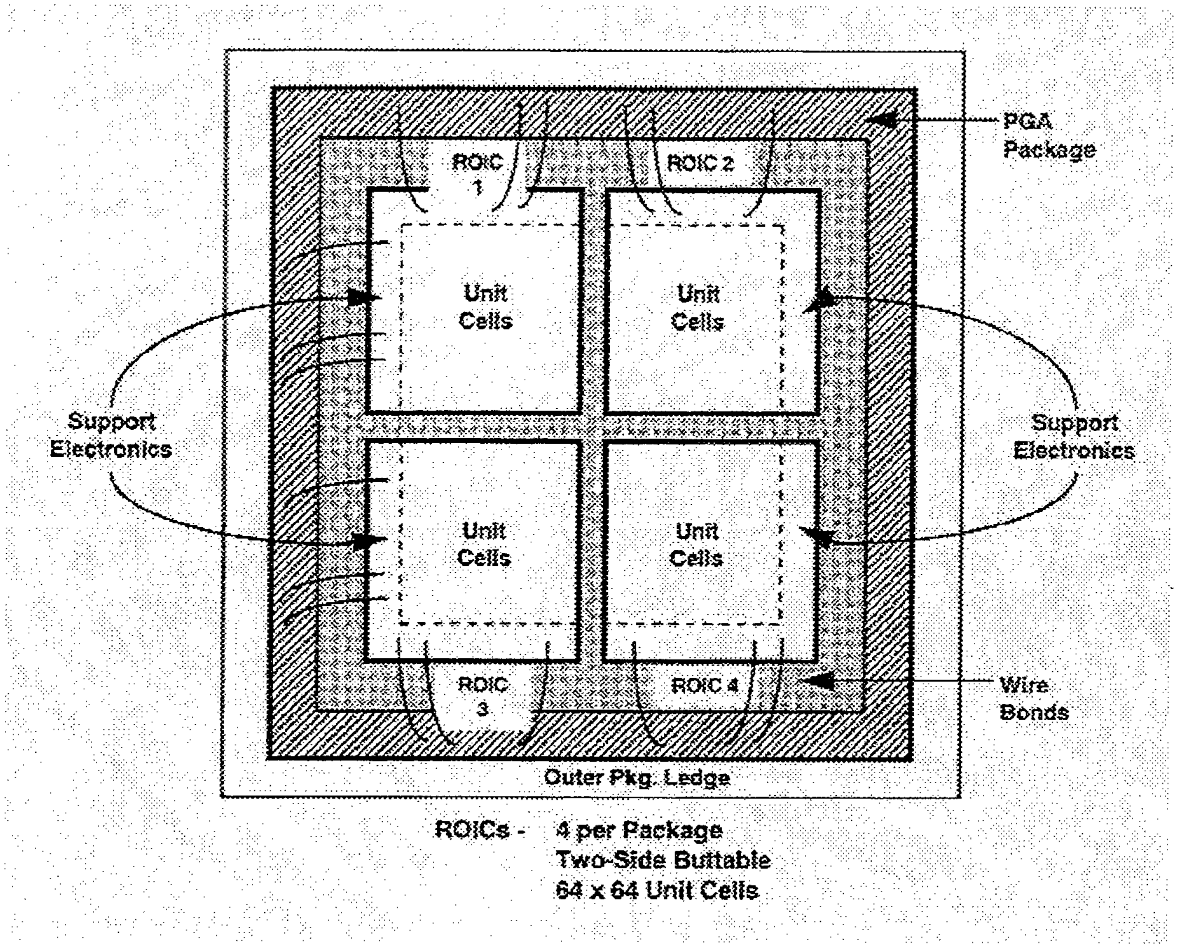

Fig. 10b.

Fig. 11 Schematic diagram of the Transducer Hybrid Assembly -(Plan view)

A 16x16 element 3.0 MHz test array was hybridized to a single ROIC for initial testing. The average measured sensitivity was -194 (±2) dB re 1V/μPA with a broadband electronic noise level of 84 μV. Using relatively uniform direct insonification, Fig. 12 demonstrates the range resolution of the THA by placing a short acoustical pulse in only one range plane out of 5.

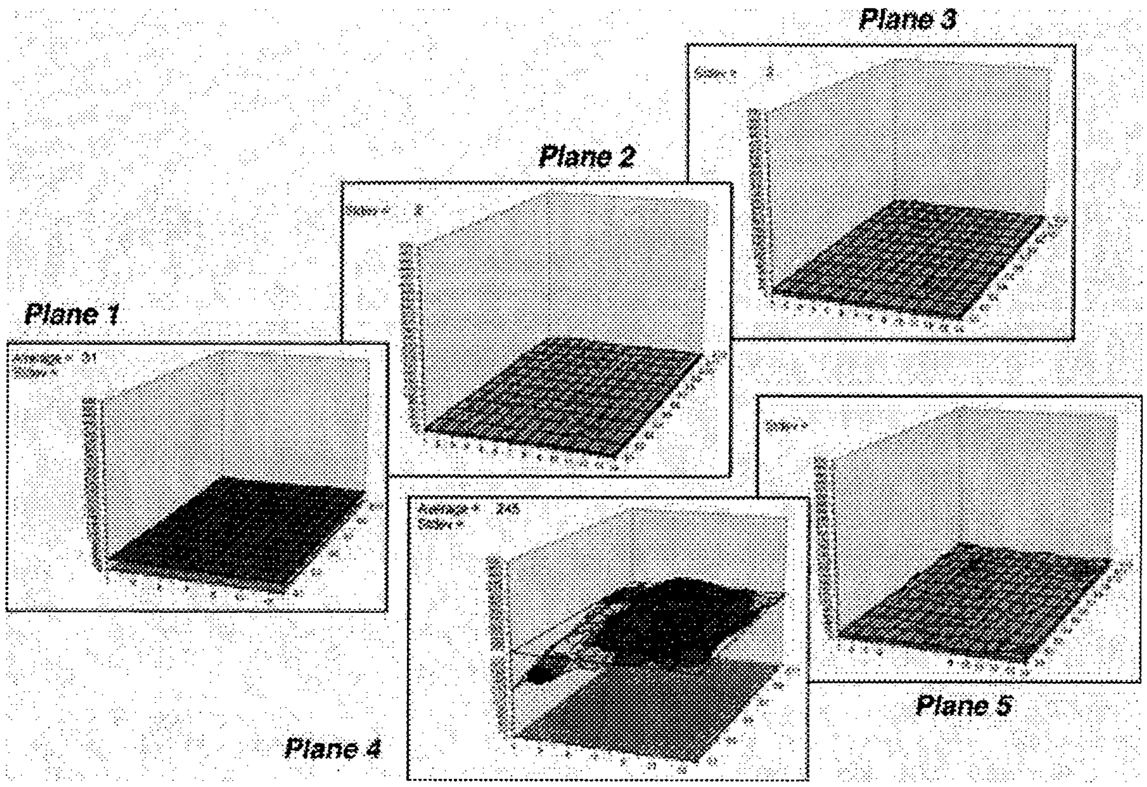

Fig. 12 Range resolution of the Transducer Hybrid Assembly

Many 128x128 THA's have been fabricated, however, the reliability of these devices never achieved an adequate level in this program, despite much process development. Poor reliability was finally traced to differential thermal expansion between the piezocomposite array and the silicon ROIC. With a temperature change of less than 10 degrees C, the array detaches from the ROIC at the indium bumps. The thermal dissipation in the ROIC's is about 4 watts and was only a partial contributor to the thermal expansion failure mode. Late in the program, it was discovered that several devices in the interface electronics were injecting substantial amounts of heat into the THA and were the primary cause of the failure. Greatly increased heat sinking of the IE helped to improve the reliability problem. This was the major issue in the later stages of the program and was responsible for major schedule variances that were reflected in added cost and therefore the reduced scope achieved.

Piezocomposite Array

A 1-3 composite piezoelectric array using PZT-5H and an epoxy resin was designed, fabricated and tested. A 50% piezoceramic volume fraction was chosen for the ultrasound array. This yields elements of improved sensitivity, with a typical capacitance of several hundred femtofarads (fF), and remains within the practical limits of present fabrication techniques. The center frequency of the array elements is 5.0 MHz with a center-to-center spacing of 0.2 mm was chosen to achieve spatial sampling of approximately one wavelength. Sawing through one electrode layer delineated the elements. A common ground layer was used on the matching layer side. 16x16 and 32x32 test arrays and 128x128 arrays were fabricated and tested.

Readout Integrated Circuit (ROIC) (Fig. 13)

Figure 13 is a block diagram of the readout integrated circuit (ROIC), a 64 x 64 (4096) element (or unit cell) array of amplification and processing electronics. The ROIC also provides the required analog multiplexing and support electronics to readout this array at a rate of up 240 Hz using four outputs, each capable of providing up to 10 million analog samples per second.

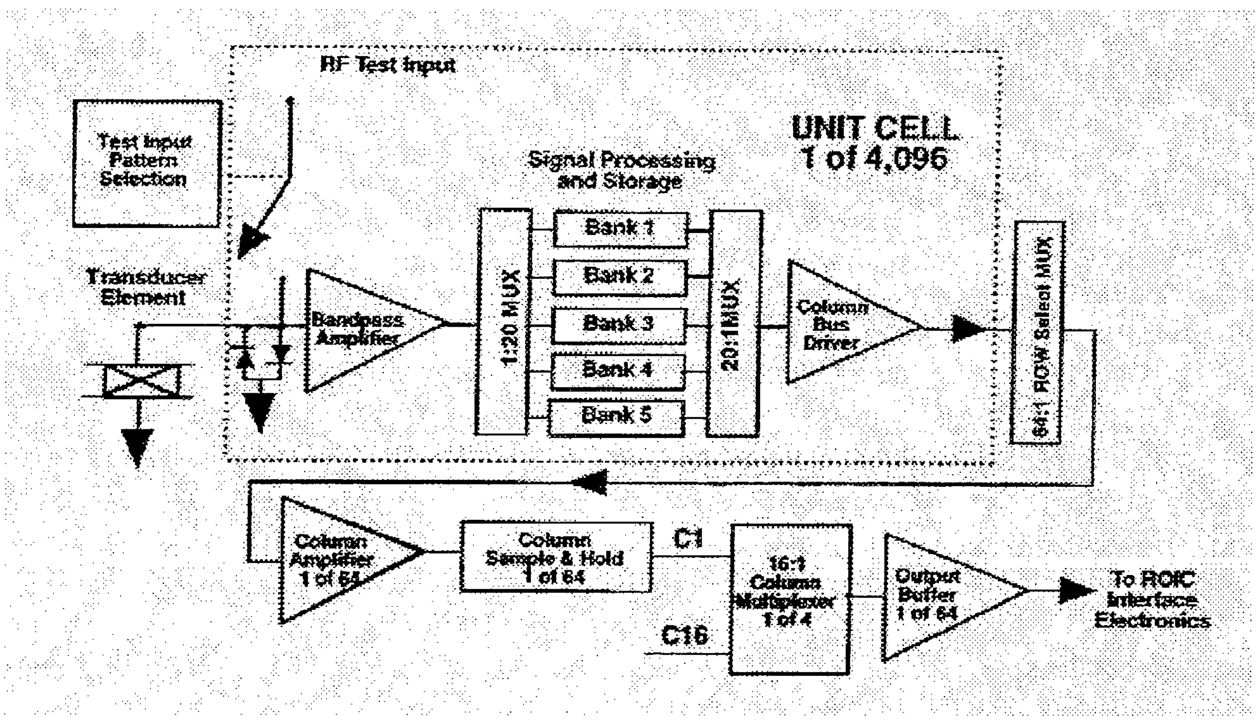

Fig. 13 Readout Integrated Circuit (ROIC) Block Diagram

The ROIC is capable of two data collection modes: in-phase and quadrature (I/Q) data from the temporal voltage signal and sample/hold of the received waveform. This data is later converted to a magnitude and phase measurement of the voltage generated by each transducer element during the period of interest defined by the range gate width. The ROIC can store up to five separate I/Q data sets, using the five storage cell banks resident in each unit cell. This allows the BUDI-LDS to capture a three-dimensional data set containing image data from up to five separate depths within the body using a single acoustic pulse.

In the sample and hold mode the voltage waveform sampled up to 20 samples per transmit pulse. These samples can be used to derive a number of different values, e.g. peak value or rms average, appropriate for use in image display.

IMAGING RESULTS

Figure 14 shows a photograph of a low contrast plastic test target together with an image taken with this lens and a 64x64 version of the array. Although this image was made at 2.8 MHz, with a substantially longer wavelength, the resolution and detail in the image demonstrates the excellent diffraction limited performance of the lens. At 5 MHz, this lens will provide the 1 mm resolution to meet one of the key goals of the program.

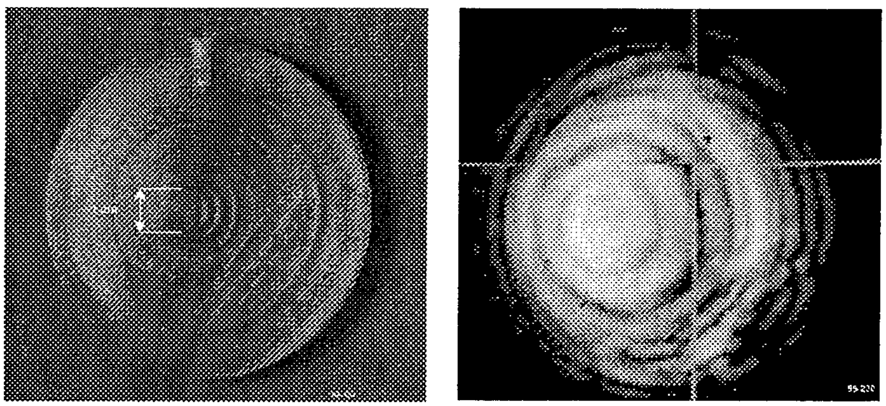

Fig. 14 Image of the surface of a low-contrast test object in water. The image is one frame of a real-time display on a video monitor. The artifactual lines in the image result from a timing error in the readout electronics. Resolution better than 2 mm is achieved.

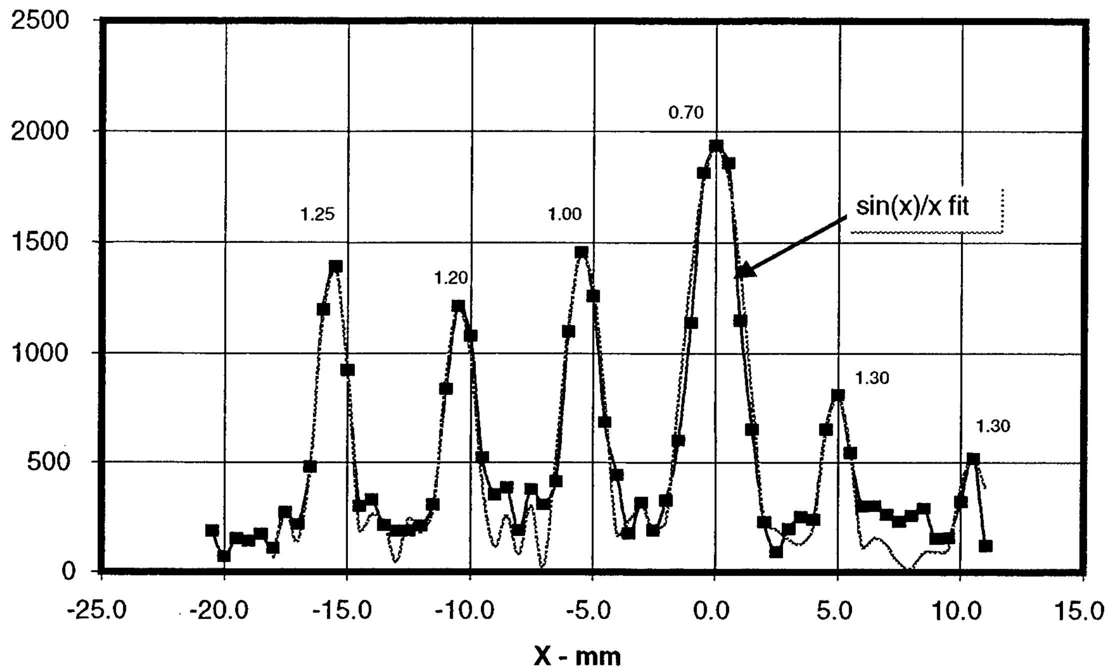

Fig. 15 Measured resolution of line targets in a *tissue equivalent medium.* made at 5 MHz with a 128x128 array. (Vertical scale is in relative units). A focus-parameter is shown at each peak: a smaller number indicates better focus.

The measured resolution is calculated from Fig. 15, a set of data from an image of a standard tissue equivalent test object. The lines in the test object are 5 mm apart. Using the data from the best-focused element (0.7 focus parameter) as the best focused line, the system resolution using the Rayleigh resolution criterion is 2 mm.

KEY RESEARCH ACCOMPLISHMENTS

There are significant scientific and technical advances demonstrated in this work:

1. A real-time 3D ultrasound imaging capability;
2. A 128x128 (16,384 total channels) dense area array with custom integrated circuit (the number of channels is at least a factor of 10 higher than any medical imaging system);
3. A large aperture, diffraction-limited acoustical lens;

REPORTABLE OUTCOMES

Manuscripts, abstracts, presentations:

1997 Presentations

Acoustical Imaging Symposium
IEEE Engineering for Medicine and Biology (EMBS)
Advanced Technology for Combat Casualty Care (ATACCC)
ONR Transducer Conference @ Penn State

1998 Presentations

Acoustical Imaging Symposium
Advanced Technology for Combat Casualty Care (ATACCC)
ONR Transducer Conference @ Penn State
IEEE UFFC Invited Paper

Patents and licenses applied for/and or issued: None

Degrees obtained that are supported by this award:

Daniel F. Lohmeyer, MSEE, Massachusetts Institute of Technology, 1998. Thesis Title: "Signal Processing in a Real-Time Three Dimensional Acoustical Imaging System".

Development of cell lines, tissue or serum repositories: N/A

Informatics such as databases and animal models, etc.: N/A

Funding applied for based on work supported by this award: (See next item)

Employment or research opportunities applied for and/or received on experiences/training supported by this award:

In 1997, a contract was awarded from NAVEODTECHDIV for a 64 x 64 3 MHz camera system using much of the BUDI design including the ROIC. More recently this worked was extended into the current DARPA funded Sonoelectronics program, also for a diver camera.

CONCLUSIONS

Lockheed Martin IR Imaging Systems has demonstrated the basic functionality of the acoustical camera design and shown that the laboratory demonstration system meets the design goals. Basic designs of all other components of the complete camera system were completed.

Although completion of the BUDI medical imaging system awaits further funding, the feasibility shown in this program has enabled development of additional cameras for Navy diver applications.

We feel confident that the BUDI camera as designed has a place in medical ultrasound practice, both in the field in military applications as well as in civilian medicine. It is one of the few system concepts that can provide real-time, high-resolution 3D images.

REFERENCES None

APPENDICES None

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LIST OF PERSONNEL

Abban, MP
Ainley, John J
Barnett, Judith M
Beaghen, PL
Boucher, S
Breton, MJ
Bures, JA
Burnham, HL
Burnham, S
Campoli, PM
Cardarelli, Andrea M
Charles, Eddy A
Clark, Kristen E
Desrochers, EA
Dunn, JA
Elwell, RM
Erikson, Kenneth R
Figucia, Frank
George, SA
Hairston, AW
Kokoszka, J
Lewis, George K
Magee, PW
Malacaria, Andrew J
Marciniec, JW
Marques, G
Marsh, CR
McDonald, PA
McPhie, Robert M
Morris, JC
Nadeau, Pauline
Nicoli, AM
Niles, T
Owen, RC Jr
Paine, MF
Peters, S
Phillips, FD
Poirier, DF
Reitz, G
Rich, GC
Righini, A
Robinson, Charles F
Schneider, RJ
Siegel, AM
Smolinsky, ES
Stockwell, Jason T
Stone, BM
Strunk, EL
Surprenant, DR
Wall, EL
Walter, Leslie
White, Timothy E